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Safety of children in cars: A review of biomechanical aspects and human body models

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ABSTRACT

The protection of children in motor vehicle crashes has improved since the introduction of child restraint systems. However, motor vehicle crashes remain one of the top leading causes of death for children. Today, computer-aided engineering is an essential part of vehicle development and it is anticipated that safety assessments will increasingly rely on simulations. Therefore, this study presents a review of important biomechanical aspects for the safety of children in cars, including child human body models, for scenarios ranging from on-road driving, emergency maneuvers, and pre-crash events to crash loading. The review is divided into four parts: Crash safety, On-road driving for forward facing children, Numerical whole body models, and Discussion and future outlook.

The first two parts provide ample references and a state-of-the-art description of important biomechanical aspects for the safety of children in cars. That children are not small adults has been known for decades and has been considered during the development of current restraints that protect the child in the crash phase. The head, neck, thorax, and pelvis are body areas where development with age changes the biomechanics and the interaction with restraint systems. The rear facing child seat distributes the crash load over a large area of the body and has proved to be a very efficient means of reducing child injuries and fatalities. Children up to age 4 years need to be seated rearward facing for optimal protection, mainly because of the proportionally large head, neck anthropometry and cartilaginous pelvis. Children aged 4 up to 12 years should use a belt positioning booster together with the vehicle seat belt to ensure good protection, as the pelvis is not fully developed and because of the smaller size of these children compared to adults. On-road driving studies have illustrated that children frequently change seated posture and may choose slouched positions that are poor for lap belt interaction if seated directly on the rear seat. Emergency maneuvers with volunteers illustrate that pre-crash loading forces forward-facing children into involuntary postures with large head displacements, having potential influence on the risk of head impact. Children, similar to adults, benefit from the safety systems offered in the vehicle. By providing child adaptability of the vehicle, such as integrated booster cushions, the child-restraint interaction can be further optimized. An example of this is the significant reduction of lap belt misuse when using integrated boosters, due to the simplified and natural positioning of the lap belt in close contact with the pelvis. The research presented in this review illustrates that there is a need for enhanced tools, such as child human body models, to take into account the requirements of children of different ages and sizes in the development of countermeasures.

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To study how children interact with restraints during on-road driving and during pre- and in-crash events, numerical child models implementing age-specific anthropometric features will be essential. The review of human whole body models covers multi body models (age 1.5 to 15 years) and finite element models (ages 3, 6, and 10 years). All reviewed child models are developed for crash scenarios. The only finite element models to implement age dependent anthropometry details for the spine and pelvis were a 3 year-old model and an upcoming 10 year-old model. One ongoing project is implementing active muscles response in a 6 year-old multi body model to study pre-crash scenarios. These active models are suitable for the next important step in providing the automotive industry with adequate tools for development and assessment of future restraint systems in the full sequence of events from pre- to in-crash.

Contents

1. Introduction

The protection of children in motor vehicle crashes has improved since the introduction of child restraint systems. However, car crashes are the second leading cause of death for children between 5 and 14 years old [1]. Different parts of the world face different challenges with regards to protecting children in traffic. In countries like India and China, child pedestrians face considerable risks of injury and fatality, and child car occupants are rarely protected with restraints developed specifically for children. In, for example, North America, Australia, and Europe, child restraints are mandatory for child occupants and still 32% of European road traffic fatalities for children up to 14 years of age involved car occupants [2]. Hence there is a need to continue research, education and policy activities to enhance the safety of children in all parts of our world.

Child occupant fatalities and injuries occur mostly in frontal and side impacts for children seated in passenger vehicles. The head is the most frequently injured body region for forward-facing children regardless of crash direction [3–5]. To understand how children are injured, Bohman et al. [6] studied the causation scenarios of head injuries in frontal impacts for rear-seated, forward-facing, restrained children and concluded that contact with the car interior, like the back of the front seat, the door panel, or the window was the principal cause of head injuries. They also found that emergency maneuvers such as braking, steering or a combination of both, influenced the kinematics of children before the impact and affected the child's interaction with the restraint systems. Consequently, to reduce head injuries further, child protection and restraint systems need to be evaluated for the whole crash sequence, including the pre-crash phase and how the child interacts with restraints in everyday on-road driving. This example illustrates the necessity to have a thorough understanding of the biomechanics of children in a range of loading situations from gravity to high severity crash loading in order to improve the safety for children today. Children are not small adults [7] and especially the head, spine, thorax, and pelvis have implications for vehicle safety. Among others, a textbook [8] from 2013 describes thoroughly pediatric biomechanics in crash loading while Stockman [9] focuses on emergency events. To the best knowledge of the authors there are no publications reviewing child kinematics for the range of scenarios from on-road driving, emergency maneuvers, pre-crash events, and through to crash loading.

Traditionally, Anthropomorphic Test Devices (ATDs) are used to evaluate safety systems in crash loading. Four main child sized ATD series are available: the Hybrid III, CRABI, the P and Q series. The Hybrid III series is composed of 3, 6 and 10 year-old children. The Q series represent 0, 1, 1.5, 3, 6 and 10 year-old children. For each ATD, the anatomical representation of body regions with regards to size and weight is based on child anthropometry databases. For instance for the Q series, the CANDAT database [10] with anthropometry data collected in the US, Europe and Japan, was used. However, in recent year numerical Human Body Models (HBMs) are becoming increasingly popular and are used to simulate both pre-crash and in-crash occupant responses. As computer capacity is increasing, safety development and assessment of vehicles are relying more and more on simulation tools. The HBMs have the potential advantage over ATDs to simulate anatomical details and predict biofidelic kinematics and injuries at tissue level. HBMs are either Finite Element (FE) or Multi Body (MB) models. FE models are usually models with detailed geometry intended for crash simulations and have strong contact definitions which are of importance to model the interaction between the occupant and the restraint systems. MB models usually have fast computational times and are relatively simple and easy to use, but contact definition is complex, and the output is mainly kinematic. While most ATDs are developed for impacts in one direction, and are therefore limited to either frontal, lateral or rear impacts, detailed HBMs can represent the human response to omnidirectional impacts if sufficiently validated. In addition, there are adult models intended for low g-loading that implement the active muscle response [11]. Hence, biofidelic child HBMs can enable optimization of safety systems based on real world data and can increase the understanding of child specific injury mechanisms. However, such models need thorough validation and should capture age-specific anatomical changes. Whole body models of children are rare and have many limitations. Still, they are important for future research and much research is currently ongoing. Therefore, an up to date overview of published whole body child HBMs would be beneficial.

This aim of this paper is to present a review of important biomechanical aspects for the safety of children in cars, including child HBMs, with a focus on the head, spine, thorax, and pelvis. The review is divided into four parts: Crash safety, On-road driving, Numerical whole body models, and Discussion and future outlook.

2. Crash safety

It is known that children are not small adults. Already in 1969, Burdi et al. [7] published a study on the structural differences between

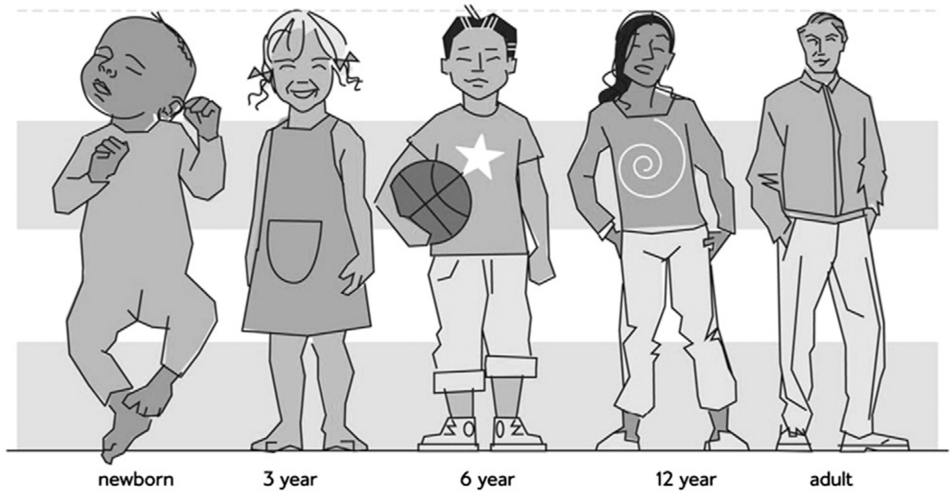

Fig. 1. The proportional changes in body segments with age as reported by [7]. Copyright Volvo Cars.

children of different ages and adults. Fig. 1 illustrates the changes in the overall body dimensions from newborn to adult, as reported in [7]. The head is proportionally larger and heavier in a child compared with an adult, and the face-brain proportions are different resulting in a higher center of gravity for a child [12]. At 3 years of age the head mass is 80% of the adult mass, and at 10 years of age it is approximately 95%. Similar anthropometrical trends have been presented by later studies [13–16].

The child neck is a slender structure and the vertebrae have large portions of cartilage where adults have fused bones [7,17]. The neck facet joints have more horizontal angles than in adults, which give less structural resistance to shear perpendicular to the spine longitudinal axis. Kasai [18] reports facet joint angles (defined between the facet surface and the spine longitudinal axis) for children between 1 and 18 years of age (values given for C7 through C3); 42 to 66° at age 1, 29 to 45° at age 6. Modest variation is seen after age 10, when the angles are 27 to 41°. The study shows that there is a positive correlation between facet joint angle and sliding distance (relative x movement between two adjacent vertebrae).

The proportionally larger head is one of the most important implications for traffic safety, firstly since it shifts the body center of gravity upward for children compared to adults and secondly because of the head inertia loading. For infants and small children, the spinal muscle strength is usually not enough to control head motion in impact situations because of the large head inertia. Due to the high center of gravity, the interaction of the child's body with the car's three-point seat belt will be different compared with an adult, and needs to be considered as it may cause extensive flexion in a frontal impact. To avoid head inertia loading of the slender neck and to distribute the energy of the impact on a greater area of the body, Aldman [19] presented the first rear facing child prototype seat in 1964 (see Fig. 2) inspired by space travel. The injury risk reduction for infants and toddlers associated with rear facing child seats (see Fig. 3) has been shown in many studies, for example [12,20–22]. In Sweden, the recommendation is to use rear facing child seats up to the age of 4 years, while the recommendation in several other European countries is until the age of 9 to 15 months and in the US is until 2 years.

Further, the structural stiffness of the rib cage, formed by the ribs, the sternum and the thoracic vertebral column, varies with age. There are several factors contributing to this variation: the geometry of the ribs and their orientation, the constitution of cartilage and bone within the ribs and the sternum, and the bone and cartilage material properties [23]. For example, the infant rib cage is mostly cartilaginous; the shafts of the ribs are ossified early, while the ends are cartilaginous until after puberty. The cartilage is most elastic in youth and becomes stiffer as the cartilage calcifies with age. Together, these factors results in a structural stiffness that gradually increases from youth to middle age [24]. Volunteer experiments in 4-g sled tests have demonstrated the combined effect of the differences in spinal and thoracic geometrical and tissue properties resulting in greater head forward and downward motion for children than adults [25]. The subjects in that study were restrained by a three-point seat belt. The majority of the spine flexion occurred at the base of the neck and the magnitude of flexion was greatest for the youngest subjects. Additional flexion occurred in the

Fig. 2. World's first rearward facing child seat, a prototype from 1964 [19], crash tested in a Volvo PV544. Copyright Volvo Cars.

Fig. 3. Examples of rearward facing child seats, from left to right: a modern infant seat (up to 13 kg); infant seat attached with an ISOFIX-base; rearward facing child seat (up to 25 kg) attached with the vehicle seat belt. Copyright Volvo Cars.

thoracic spine as well. This kinematic response may partially explain the occurrence of head injury due to impact also among children of short stature identified in [6].

Another important anatomical and physiological difference concerns the pelvis [23]. At birth, the pelvis is mainly cartilaginous. The ossification occurs gradually in three separate areas until the age of 8 years. The fusion of these three areas is required so that the pelvis becomes a stable ring structure, capable of being loaded, and does not occur until puberty. At puberty, the iliac spines are fully developed. The iliac spines are the bony processes at the upper front edges of the pelvis and are considered to be very important for proper interaction with the lap portion of the seat belt. Before the pelvis is fully developed, the concept of the three-point adult seat belt does not work on its own. Belt-positioning booster cushions were introduced in the late 1970s [26] (see Fig. 4). Today, there are three main belt-positioning boosters; booster cushions, highback booster seats (including seat backs) and integrated (built-in) booster cushions. The systems are used with the adult seat belt which restrains both the child and the booster. The integrated boosters were developed in order to simplify usage and to minimize misuse [27]. The booster allows the geometry of the adult seat belt to function in a better way with respect to the child occupant (see Fig. 5). The booster raises the child, so that the lap part of the adult seat belt can be positioned over the thighs, which reduces the risk of the abdomen interacting with the belt [4]. User studies have shown that current designs of boosters can lead to misuse. Incorrectly routed lap belts over the belt guides, incorrectly routed shoulder belt position under the arm or behind the back and loose belt are examples of such misuse [28,29]. Consequently, the belt routing will then not optimally interact with the child's body and the protective effect is reduced [30].

Fig. 4. World's first booster cushion from 1978 [26]. Copyright Volvo Cars.

The adult seat belt in combination with a booster is recommended for ages 4 to 12 years. For smaller children, the rearward facing seat offers the best protection for the pelvis during a frontal impact by transferring the loads to the pelvis by the seat back. The child is secured in the rearward facing seat by a harness including a crotch strap that keeps the lap belt down on the pelvis (Fig. 3).

Hence, from a biomechanical point of view it is clear how best to protect children in crashes; rearward facing seats up to 4 years of age (Figs. 2–3) and thereafter the vehicle seat belt together with a belt positioning booster (Figs. 4–5). Forward facing child seats where the child is restrained by a child harness, which are used in several countries for children aged 1 to 4 years, are not to be recommended for children below 4 years since a forward facing seat will not provide the distributed load needed to optimally protect the head and neck of small children.

3. On-road driving for forward facing children

Child occupants do not always sit like ATDs that are seated in a standardized posture according to protocols for crash tests. The real-world seated posture can either be a self-selected posture or an involuntary posture due to rapid motion of the vehicle [9]. The self-selected postures are influenced by children's activities and distractions, as well as perceived discomfort, including child restraint systems and seat belt positions. As an example Osvalder et al. [31] showed that playing with a smart phone often resulted in a forward flexed seated posture with the head leaned forward (see Fig. 6). This seated posture poses other challenges in a crash than when seated upright.

Several studies have explored seated postures in children of different ages during on-road driving [31–36] by analyzing video recordings or photos of the children. Several factors such as age, restraint design and seat belt geometry, discomfort, activity, and road environment affect seated postures (see Fig. 6). Sitting upright was the most common lateral seated posture [31,32,34,35]. However, children often sought support by leaning to the side of the side supports and resting their head against the side supports of the child restraint system or leaning the lower arm and/or the head towards the door panel. Substantial inboard or forward tilting often occurred periodically as the child was reaching for something, interacting with the parent in the driver's seat, or looking out through the windows [31,35]. The seated posture was also influenced by possibilities for finding support for the feet. Children occasionally stretched out one of their legs to find support against the back of the front seat or by putting the right foot on the sill and the left foot on the center panel in front of the middle seat position in the rear seat [31,32].

Highback booster seats with side supports help restrict lateral motion and provide head and torso support for children when sleeping [36]. However, another study showed that for children who were not

Fig. 5. Illustration of belt fit for children using, from left to right: belt positioning booster cushion, highback booster seat (Copyright Volvo Cars), and integrated booster cushion (Copyright Volvo Cars).

sleeping, larger head side supports increased the likelihood of a more forward seated posture without head and shoulder contact with the booster's back, as compared to smaller or no head side supports [31, 34]. Possible reasons for this behavior were discomfort, looking out through the side window, and talking to the driver. Children aged 7 to 9 years when seated on a highback booster seat spent less time seated with the upper back and shoulders in contact with the backrest compared with children on integrated booster cushions [31]. Also, several

Fig. 6. Examples of voluntary seated postures in [31,35], from top left to bottom right: forward leaning posture when the child is playing with a smart phone; rotation of the upper torso to play with a smart phone when seated in a highback booster seat; slouching to comfortably accommodate bent knees when seated directly on the rear seat; child seated on an integrated booster.

Fig. 7. Examples of slouch in [35] for a child seated directly on the rear seat (left) and a child of the same age seated on a belt positioning booster cushion (right).

children were not able to sit comfortably with their entire back and arms within the side supports, especially not when they were using a smart phone or similar devices. Furthermore, individual variation in seated posture among children when seated on the highback booster seat was greater than when seated on the integrated booster cushion [31]. During long periods of on-road driving, children showed noticeable signs of tiredness or discomfort by changing posture more often compared to at the beginning of the ride, and choosing either an upright sagittal posture without seat back contact or a forward leaning torso posture to support the upper body with their arms [34].

Jakobsson et al. [35] compared the seated postures of children 8 to 13 year-old when seated on a booster cushion compared to when seated directly on the seat. It was observed that half of the children assumed a slouched position when seated directly on the car seat without the booster cushion, while slouching was rare when using booster cushions. One child was slouching throughout the majority of the ride and it was obvious that his pelvis rotated due to discomfort as his thigh was shorter than the rear seat. Even in slouched seated postures, the booster helped to guide the lap belt on the pelvis and below the abdomen (see Fig. 7). For all children, the booster cushion helped to keep a more stable lateral seated posture and the absence of the booster resulted in a greater variation of lateral tilting postures. The children seated directly on the seat more often had neck contact with the seat belt and in order to avoid the neck discomfort, the children leaned inboard relative to the vehicle or rotated their upper body.

Hence, the voluntary seated postures of children are affected by several factors. Large side supports of highback booster seats can result in a more forward seated posture, seat belt contact with the neck is avoided by leaning or rotating the upper body away from the belt, and activities such as using a smart phone device or interacting with the driver result in a flexed head position or inboard leaning. The findings in the available on-road driving studies provide increased knowledge on how the design of restraint systems, perceived discomfort, activity and riding conditions affect seated posture for children of different ages; findings that are of importance to understand how the biomechanics of the child will influence the restraint design.

In a maneuver study comprising child volunteers [37,38], vehicle and video data was recorded simultaneously to enable a direct relation between child kinematic responses, vehicle interior and vehicle movement. Sixteen children aged 4 to 10 years-old were exposed to evasive braking and steering events when seated in different restraint systems. During braking events short children moved forward and downward with a greater flexion motion of the head compared with the tall children who had a more upright forward motion. The head trajectories were influenced by the size of the child and the restraint system used. The backrest of the highback booster seat affected the initial position of the child relative to the vehicle and thus resulted in a more forward posture at maximum displacement. Furthermore, the maximum forward head position was outside the highback booster's side supports [38]. In the steering maneuvers, the seat belt slipped off the shoulder

Fig. 8. Example of a short child in an involuntary seated posture during the emergency steering event in [37].

in 20% of the maneuvers, varying by age of the child and the restraint system used. Among the short children, shoulder belt slip-off occurred in almost 67% of the trials when seated on a booster cushion (see Fig. 8) while for tall children belt slip-off did not occur. The differences in shoulder belt slip-off between the short and the tall children may be explained by the fact that the stature of the tall children allowed the belt to have a grabbing effect on the shoulder while the short children slipped out of the belt immediately. Also, tall children have wider shoulders compared with short children. Another finding was a difference in how the lower part of the shoulder belt contacted the lower torso. For the majority of the short children the shoulder belt position was low on the abdomen's side with a gap between the belt and the body while for the tall children the position of the belt was higher and the belt was tighter. The high abdominal position of the shoulder belt portion may restrict the lateral movement by supporting the lower torso [37].

The child's seated posture when potentially exposed to a crash is influenced by self-selected, as well as involuntary factors. Irrespective of the reason why, to optimally protect a child there is a need to understand how the interaction between the child and the restraint system changes due to different seated postures as well as the variety of crash scenarios. Moreover, the growing child needs to be taken into account and factors such as initial seated posture, how the child is restrained by the seat belt, and the effect of the booster design must be further understood.

4. Numerical whole body models

The following chapter provides an overview of child HBMs. The MB models span children of the age from 1.5 to 15 year-old and the FE models cover 3, 6, and 10 year-old. To the best knowledge of the authors, there are no published FE HBMs for other age groups. The project Child Advanced Safety Project for European Roads (CASPER) has developed detailed FE models of the ages 6 months, 1, 3 and 6 year-old [39]. To date, there is no publication on the development or validation of these complete models. The Digital Child Project proposed by the Southern Consortium for Injury Biomechanics started in 2006 and has modeled 3, 6 and 10 year-old children with highly detailed anthropometry based on medical imaging data sets [40]. However, neither validation nor applications of the whole body models have been published yet.

Frequently used FE codes are LS-DYNA (Livermore Software Technology Corporation, Livermore, California, US), RADIOSS (Altair Engineering Inc, Troy, Michigan, US), and PAM-CRASH (ESI Group, Paris, France) and for MB and coupled MB and FE simulations the MADYMO (TASS, Rijswijk, The Netherlands) code is used.

4.1. Finite element models

In 2005, Koizumi et al. [41] developed a 3 year-old occupant FE model in the MADYMO code, with kinematic joints for the skeletal articulations. The geometry was obtained by scaling a 50th percentile male occupant model developed for MADYMO [42] in the HUMOS project to a child geometry based on anthropometry data from the CANDAT database [10] using the probabilistic Kriging technique to fit the adult FE surfaces to the 3 year-old anthropometry. The head and neck dimensions were close to the database, less than 3.5% for the measures: head breadth, depth, height, circumference, and neck breath and circumference. Hence, the head inertia and relative slender neck is captured with this scaling, while anatomical differences in spinal development of a 3 year-old are not implemented. Also, it should be noted that the original adult model does not have a normal cervical spine seated lordosis due to the anthropometry of the cadaveric specimen it is based on and it is not clear if this was considered in the child model. The joint ranges of motions were assumed to be equal to the adult ranges of motion, which for the spine is a severe limitation. For material properties, only bone properties were scaled down using the Irwin and Mertz method [43] while the other material properties were kept equal to those of the adult model. The model was validated by comparing its response with corridors used for the 3 year-old ATDs certification: frontal and lateral thoracic impact tests (pendulum test, velocity: 4.3 m/s and 6.7 m/s for fontal tests and 4.3 m/s for lateral tests). Then, the model was used to reproduce the UNECE-R44 regulation fontal sled test (specific acceleration level and velocity are not specified). Its response was compared with the response of the Hybrid III 3-year-old which showed that the head displacement was greater for the HBM.

In 2005, Mizuno et al. [44] developed a 3 year-old occupant model in the LS-DYNA code which has been widely used, and successively improved to increase its biofidelity. The geometry was obtained by scaling the 50th percentile male Total Human Model for Safety (THUMS) version AM50 [45,46] based on anthropometry data from [10] using linear scaling. The same scale factor was used in the x, y, and z direction to maintain the bone shapes and was determined for each body part separately. The head scale factor was 0.879 and the neck scale factor was 0.557 illustrating that some aspects of the head-neck anthropometry were captured. However, the head circumference was still smaller than the data in [10] and anatomical differences in spinal development of a 3 year-old are not implemented. The material properties of bones were scaled from adult data using the Irwin and Mertz method [43]. The skin was made softer while the mechanical characteristics of other soft tissues were kept the same as that of the THUMS adult model. Validation of the model was focused on the neck, spine and torso. This was achieved by comparing the model response with corridors on child volunteers for the abdomen (lap belt loading) and with the Hybrid III 3 year-old physical ATD response in calibration tests for the neck (pendulum test on the thorax, acceleration: 230 m/s^2 during 20 ms), thorax (pendulum test on the thorax, velocity: 6 m/s) and spine (flexion test at 45°). The model response was then compared to the response of the physical Hybrid III 3 year-old ATD in a frontal sled test according to UNECE-R44 regulation (acceleration: 24.5 g, velocity: 50 km/h). A qualitative analysis showed higher spine flexibility for the model and thus different kinematics response between the model and the ATD, leading to a more human-like response for the HBM compared to the ATD model. Nevertheless, the main limitations of the model were the lack of experimental data for validation, the lack of material properties for child tissues and the lack of anatomical shape of child bones, especially for the cervical vertebrae, rib cage and pelvis.

A year later, Mizuno et al. [47] presented an improvement to the pelvis of the HBM in [44]. The shape of the pelvis iliac crest was modified and cartilage was added based on medical imaging of a 5 year-old child, thus capturing the age related differences between the child and adult pelvis that are important for lap belt interaction. The material properties of bone and cartilage were the same as in [44]. Validation of the model was obtained by simulating a pendulum side impact test according to the Q3 ATD certification protocol (impactor velocity: 5.2 m/s). Compared to the previous pelvis model, the pelvis model with anatomical child shape and cartilage presented less stress in the bones.

Further improvement of the head and neck complex was presented in 2009 by Zhang et al. [48], for the 3 year-old occupant model in [47]. The mechanical properties of the ligaments, intervertebral disks and joints of the cervical spine were adjusted using a trial and error process where the apparent stiffness distribution and energy of the head and neck model were compared with the response of reported pediatric cadaver head and neck complexes [49]. Nine specimen from subjects aged less than 8 year-old were tested in flexion–extension bending (moment applied to T1 vertebra: $\pm$2.4 Nm in 100 ms) and tensile testing (velocity applied to head center of mass: 50 cm/s for 40 ms). Hence, even though the age specific vertebral and facet geometry is not implemented the spinal model has increased biofidelity and highlights the influence of soft tissues on the head and neck kinematics. Li et al. [50] compared the response of the HBM without any improvement to the neck [47] and the HBM with head and neck improvements [48] in near side and far side impacts (acceleration: 22 g) when properly and incorrectly restrained. They concluded that because of the increased biofidelic mechanical properties, the HBM in [48] exhibited higher head accelerations than the HBM in [47] for similar chest accelerations.

The response of all three versions of this particular 3 year-old child HBM has been compared to the response of FE ATD models of corresponding age (Hybrid III 3-year-old, Q3 and Q3S) developed by Humanetics Ltd (former First Technology Safety Systems) in frontal and lateral sled tests, in accordance with UNECE regulations [44,47,48]. Also, the HBM in [44,47,48] and 3 year-old child ATD models were used to evaluate child restraint systems in frontal impacts [51–55] and side impacts [50,53,56,57], as well as to evaluate injury mitigation systems in side impacts [58]. The main result of these studies was the lack of spine flexibility in the ATD models compared to the child model. Among others, Zhang et al. [55] compared the head trajectory of HBM in [47] with the response of a child cadaver and the numerical Hybrid III 3 year-old in a frontal sled test with peak acceleration of 17 g [59]. The child model response was closer to the cadaver response than to the ATD response. This result was explained by the increased neck flexibility of the child model. Moreover, the head rotation around the y-axis was 14° higher for the HBM than for the ATD model. As a result, contact between the chin and the chest was observed for the HBM but not for the ATD.

In 2003, Okamoto et al. [60] developed a 6 year-old pedestrian model in the PAM-CRASH code with detailed lower limbs and only skin surface for the pelvis and above. Geometry of the lower limb model was based on magnetic resonance imaging of one child volunteer and included high anatomical detail such as growth plates and age specific anthropometry geometry of the cartilaginous structures at the femoral head, knee, and ankle. The upper body was rigid and its geometry was scaled down from adult data. Regarding the mechanical properties, no information was given by the authors. Neither validation nor application of this model has been found in the literature. However, it is worth noting that this is the only model where the age specific anthropometry of the child was taken directly from medical imaging of a child, instead of relying on data scaled from adults.

In 2007, Iwamoto et al. [61] presented a 6 year-old pedestrian model in the LS-DYNA code. The model was scaled down from the adult THUMS, which is not described in [59]. The model has not been validated.

In 2012, Mao et al. [62] presented the project to create a 10 year-old occupant model in the LS-DYNA code based on computer tomography and medical imaging of one child close to the height of an average 10 year-old. This approach can by default take all age-related anthropometry into account. Growth plates were modeled as cartilaginous tissue. Material properties for bone were scaled from adult data based on the density in the computer tomography images and soft tissue properties were assumed by down scaling from adult data. So far, the detailed models of the neck and thorax have been published [63–65] with neck validation compared to cadaveric data in tensile testing and flexion/extension in [63,64], thoracic stiffness compared to cardiopulmonary resuscitation data in [65] and preliminary cadaveric belt and hub loading in [62]. This model has the potential to be a very strong tool for development and evaluation of safety devices in crash loading when the complete whole body model and validation is published.

4.2. Multi body models

MB models of children were first developed to model child pedestrians. In 2002, Liu and Yang [66] developed child pedestrian models representing children of ages 3, 6, 9 and 15 year-old in the MADYMO code by scaling a 50th percentile male pedestrian model [67] composed of 15 ellipsoids linked together with 14 three-dimensional joints. Characteristic body dimensions for children and adults by [14] were used to determine x-, y-, and z-dimensional scale factors for each body segment. The joint properties were scaled from adult data using a simplified model based on experimental data of a lumbar vertebral unit [68] and knee joint [69]. Validation of the child pedestrian models was achieved by reconstruction of two road accidents with a 7 and a 9 year-old pedestrian. The overall trajectories and head impact locations agreed well with the accident data.

In 2003, Van Hoof et al. [70] developed child pedestrian models representing 3 and 6 year-old children in the MADYMO code by scaling a 50th percentile male pedestrian model [71] based on the specification of the Q child dummies [10]. Each model was composed of 64 ellipsoids and 52 kinematic joints. Mechanical properties were scaled down similarly to [66]. Validation of the child pedestrian models was not provided directly; only the 50th percentile male pedestrian model was validated by comparing different impact tests of the lower extremities, the pelvis, the abdomen, the thorax and the shoulder with post mortem human subjects. The 6 year-old pedestrian model was used to reconstruct and evaluate pedestrian kinematics in a car to pedestrian collision (40 km/h, brake deceleration of 0.7 g) [72]. Also, the 6 year-old pedestrian model with modified head contact characteristics was used to reproduce falls of children from a playground climbing frame [73]. Another study [74] developed optimization techniques to simulate a child running in front of a vehicle, with the same model. For instance, the limitations highlighted in these studies were the lack of accurate material properties in the literature to describe the response of the head contact characteristics and the lack of muscular activity that could potentially affect the kinematics of the child falling in the playground.

In 2005, occupant MB facet models were presented by van Rooij et al. [32] for a 1.5 and a 3 year-old child, developed in the MADYMO code based on the 50th percentile male facet model by [75] and subsequently extended to 6 and 10 year-old child HBMs [76]. The anthropometry was obtained by scaling the adult model to the specifications of the Q ATDs, similarly to the development of the pedestrian models [70]. The facet models consist of 92 bodies. The main differences compared to the ellipsoid models are the presence of a meshed skin, a deformable torso and a greater number of joints to model the spine. Each spinal segment is represented by a joint to increase the flexibility of the spine. Mechanical properties were scaled down from adult data. Only the validation of the 6 year-old model has been presented. The response of the model was compared with scaled corridors obtained from dynamic hub impactor tests [77]. The impactor mass and diameter were scaled according to the Irwin and Mertz [43] scaling method which resulted in an impactor mass of 5.3 kg. The impactor speed was 4.3 m/s and 6.7 m/s. The 6 year-old model was also validated against frontal thoracic pendulum tests (impactor mass: 3.5 kg, impactor speed: 6 m/s) performed on pediatric PMHSs [78], abdominal belt loading tests on porcine specimens [79], and quasi-static neck tension tests on pediatric PMHSs [49].

5. Discussion and future outlook

Today, the age group 4 to 12 year-old is not included in standardized dynamic vehicle testing either in regulatory or consumer rating programs. These children have been “forgotten” in the legal process of certifying vehicles [80]. Currently, most child restraints are developed as independent systems and not integrated as a part of a vehicle, since testing and certification is generally based on sled tests using a generic buck rather than full vehicle crash tests [81]. Tylko and Bussières [82,83] showed that since booster seats rely on the vehicle seat belt to provide restraint, booster seats should be designed and tested dynamically together with the vehicle to ensure good safety performance.

In general, improved vehicle structures in modern vehicles ensure reduced intrusion into the vehicle compartment in motor vehicle crashes. To some extent this has resulted in more severe crash pulses [84,85]. Energy management of front seat occupant protection, through the use of airbags, belt pretensioners and load limiters, for example, has not been addressed to the same level for rear seat occupant protection [86]. Recent studies have shown that pretensioners and load limiters are beneficial for crash test dummies and adult post-mortem human subjects in the rear seat, both in terms of improved kinematics and reduced risk of injury to the neck and thorax in frontal impacts [87–89]. This illustrates the potential for pretensioners and load limiters being beneficial for children in the real world as well. The ongoing update of EuroNCAP will encourage energy management in the rear seat.

Children, much like adults, benefit from the safety systems offered inside vehicles. Besides seat belt load limiters and pretensioners, some vehicles provide integrated (built-in) booster cushions that have been shown to be very easy to handle [90] and almost eliminate lap belt misuse [26]. In addition, the majority of vehicles today offer inflatable curtains as part of side impact protection. These systems should take children as well as adults into account when being developed. It is recommended that vehicle design and in-vehicle restraint systems include the protection of children above the age of 4 years to limit the need for additional aftermarket child restraint systems to those children requiring rearward-facing seats only. In particular, countries lacking child restraint recommendations and regulations have the opportunity to implement these safety aspects while regulations are under development. They would thereby be given access to less expensive aftermarket products and have the potential to reduce the misuse rate of child restraint systems.

Today, the choice of an optimal child restraint system is complex due to the fact that seated postures and the behavior of children are highly affected by the design of the restraint system and factors such as the time of the day and the road environment. For example, tired children will most likely benefit from side supports to lean on, as will very active children whose lateral motion may be restricted by the side supports; whereas a child engaged in an activity would be more comfortable seated in an upright position without the backrest and would have the option of looking out the window without leaning forward. Built-in child restraint systems can help address these aspects and take the biomechanics of active children into account in a more complete manner.

The on-road driving chapter highlights the challenge of keeping children properly restrained in a variety of situations, ensuring that they achieve a biomechanically optimal restraint interaction during both self-selected and involuntary seated postures during normal on-road driving as well as during pre-crash events. Some possible countermeasures to help keep the shoulder belt on the shoulder in a pre-crash maneuver have been investigated, albeit mainly on adults [91–95]. There is a need to further assess these countermeasures for children, taking into consideration the biomechanics of children and the specific challenges associated with child occupants highlighted in the current study. This emphasizes the urgency to enhance the tools, such as pediatric ATDs as well as child human body models, to take into account the needs of children of different ages and sizes during the development of these countermeasures.

Currently available tools are not sufficient to study how children interact with restraints during on-road driving in pre-crash and in-crash events. All reviewed child HBMs have been developed for crash scenarios and the limited validation focuses on crash loading. The only FE models to implement age dependent anthropometry details for the spine and pelvis were the 3 year-old model in [48] and the coming 10 year-old model in [62]. For the purpose of crash simulations, there is a need to at least publish FE models of the infant where the only potential model was designed in [39]. In addition, it needs to be stressed that all child HBMs require further validation for crash loading. The challenge is, as always, the lack of experimental data. In recent years, many efforts have been made to provide additional cadaveric data [8], volunteer data for low-severity crashes, for example in [25,92,96–98], and thoracic stiffness data collected during cardiopulmonary resuscitation [24]. To study on-road driving and involuntary child seated postures during emergency and pre-crash events, it is necessary to implement muscle activity in the HBM. Child models can be enhanced similarly to the adult models that have been implemented with active muscle response [11] to analyze pre-crash events. An ongoing project is implementing active muscle response in the 6 year-old MB model [76] and validating the kinematic performance to experimental child volunteer on-road driving data and low g sled test [99]. These active HBMs are suitable for the next important step in providing the automotive industry with adequate tools for development and assessment of future restraint systems in the full sequence of events from pre- to in-crash.

In conclusion, to improve the safety of children in cars there is a need to: improve the tools used for assessment and development of child safety; further promote vehicle built-in child restraint systems that accommodate the specific needs of children and their biomechanical responses; understand the biomechanics of children throughout the crash sequence; and how the pre-crash response influences the outcome of a crash.

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