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Safety Analysis of Battery-Powered Adaptive Ride-on Toys for Children with Disabilities

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Safety Analysis of Battery-Powered Adaptive Ride-on Toys for Children with Disabilities

By

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3. Abstract

Modified battery-powered ride-on toy cars, or adaptive ride-on toys, represent novel rehabilitation tools and developmental aids for children with disabilities. Studies have shown that children are benefiting socially and developmentally from their use. However, the use of these toys by children with disabilities potentially poses a risk of injury and it is vitally important to ensure the safe use of these toys, particularly for the benefit of those with developmental challenges.

Within this context, the purpose of the first study was to determine whether modifications to ride-on toys are sufficient to prevent common modes of injury such as falls, passenger excursion, and impact with the interior of the vehicle using an average six-year-old test dummy. Because the population of children with disabilities who are receiving adaptive ride-on toys ha a wide range of mobility impairments and may suffer from a wide range of musculoskeletal disorders, those with both decreased and increased muscle stiffness were considered in the second study. In both studies, safety modifications sufficiently reduced risk of primary injury mechanisms with little-to-no added risk.

These studies are significant due to lack of research in the field of safety of pediatric rehabilitative devices, specifically adaptive ride-on toys. The proven success of these rehabilitative programs further shows these studies are a valuable tool intended to better equip pediatric care providers with knowledge on the safety of car modifications. Furthermore, the findings of these studies support the growth of adaptive ride-on toy programs to increase rehabilitation opportunities for children with disabilities.

4. Introduction

4.1. Background

Play and independent mobility are critical to the development of a child. Furthermore, environmental factors heavily influence a child's neurodevelopment and exposure to neural stimuli is critical for children at a young age. However, children with neuromusculoskeletal impairments have significantly decreased mobility and therefore limited opportunity to explore their physical and social environments leading to impaired development[1]. Assistive technology exists to provide independent mobility but is accompanied with various limitations. Of these limitations, a primary issue is that children often do not receive assistive technology until after the age of five. This presents an issue for the neurodevelopment of children with disabilities because the years prior to age five are the most critical.

Modified battery-powered ride-on toy cars, or adaptive ride-on toys, represent novel rehabilitation tools and developmental aids for children with disabilities – primarily those under the age of five. Studies have shown that children are benefiting socially and developmentally from their use[2]. However, the use of these toys by children with disabilities potentially poses a risk of injury as these ride-on toys were not originally intended for this population by commercial manufacturers[3]. It is important to ensure the safe use of these toys, particularly for the benefit of those with developmental challenges, as they enable them to play and be independent.

4.1.1. Neurodevelopment

Biologically, neurodevelopment is the process of creating and refining billions of complex chemical transactions to produce the human brain and nervous system [4]. The mature, adult,

human brain contains 100 billion neurons (specialized nerve cells that transmit electrical signals throughout the body) and countless more neuroglial cells (interstitial, supportive tissue of the nervous system that maintains the ionic milieu of nerve cells, modulate the rate of nerve signal propagation, modulate synaptic action, and aid in recovery from neural injury)[5] . Trillions of specialized connections, or junctions, known as synapses, connect these cells. Synapses allows communication to take place between the cells in a process known as neurotransmission[6]. There are eight key stages, or processes, of neurodevelopment: neurogenesis, migration, differentiation, apoptosis, arborization, synaptogenesis, synaptic sculpting, and myelination. At each of these key processes, genetic and environmental factors play critical roles. During the first few stages, the brain grows rapidly until synapse formation reaches a maximum at the age of approximately 3 years, shown in Figure 1[7].

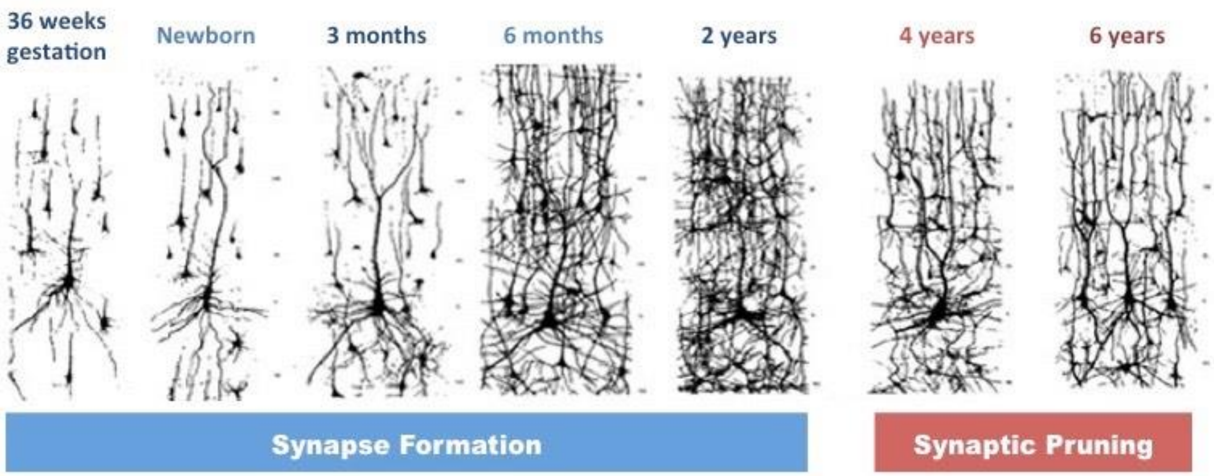


Figure 1: While infant and adult brains typically contain the same number of neurons, infant brains form vastly more synaptic connections (approximately twice the number of an adult brain). Pruning begins at approximately the age of three-year-old. [7]

During neurodevelopment, the brain is constantly undergoing physical and chemical changes as it responds to its environment, indicating enormous plasticity [8] . Plasticity reaches a peak at three years of age, before the process of apoptosis occurs (between ages four and six). The purpose of apoptosis, or pruning, is to reinforce complex wiring patterns associated with learning behavior and is therefore predominantly influenced by environmental factors. This process is heavily activity-dependent, and redundant neurons or those with little activity resorb (thus the phrase "use it or lose it")[6]. Therefore, it is critical for children to be exposed to neural stimuli very early on or their development will be severely stunted. The consequences of neglect can be seen in Figure 2.

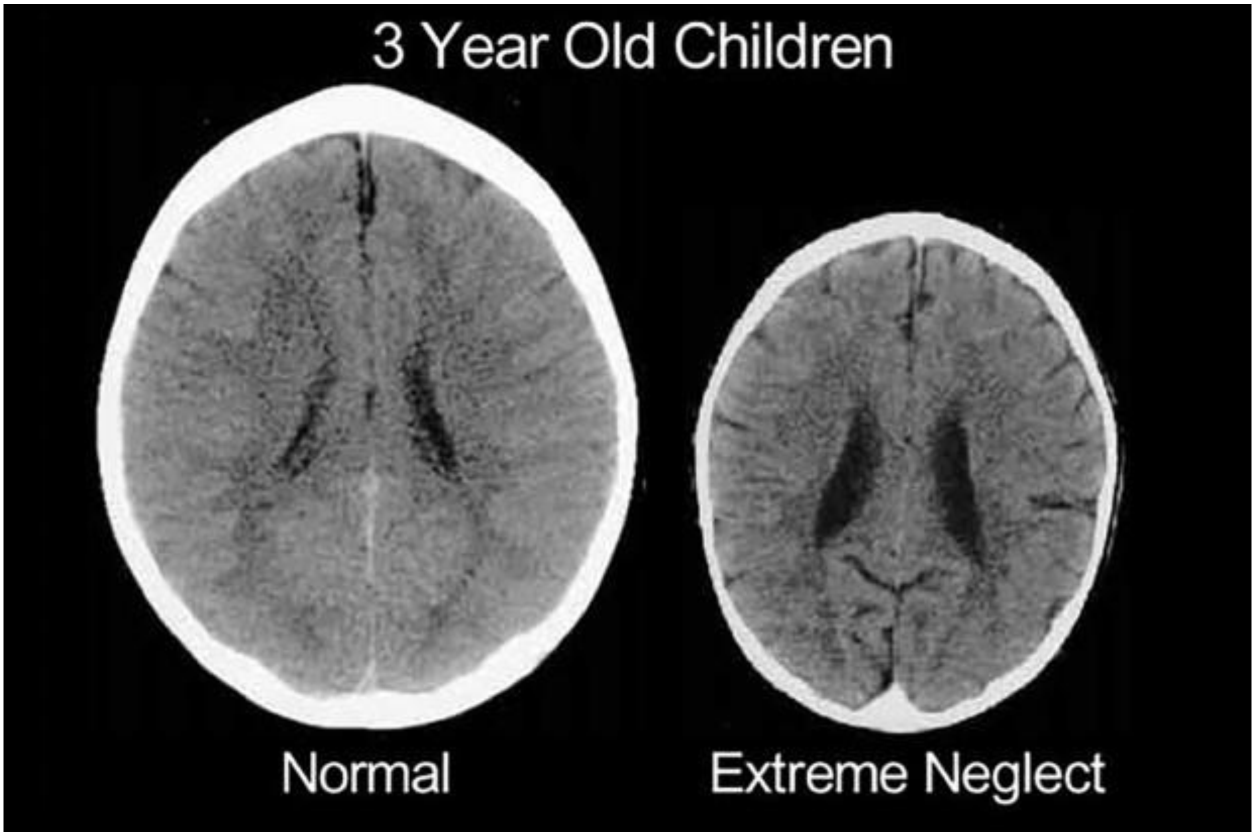


Figure 2: Abnormal brain development following sensory neglect in early childhood. The CT scan on the left depicts the brain of a healthy three-year-old with an average head size (50th percentile). The image on the right depicts a three-year-old suffering from severe sensory-deprivation neglect. The size of the child's brain is significantly smaller (3rd percentile) and has enlarged ventricles and cortical atrophy. [6]

Studies have been conducted to determine the implications of varying kinds of environments on brain growth for decades and suggest that interaction with enriched environments resulted in brain growth, while interaction with impoverished environments did not[9]-[12]. There are recorded increases in IQ of over 40 points in children following removal from neglectful environments and placement in nurturing and enriching environments [6] . Another study emphasized the role the environment plays in brain development and how necessary it is to highlight the serious consequences of 'play-deprivation' [12]. A key aspect of an enriching environment is play as the predominant outlet of learning for children.

4.1.2. Play

It is difficult to consider neurodevelopment without also considering play, as children spend more time playing than any other activity [13]. Therefore, play holds the largest role in a child's neural development and assists in the actualization of brain potential [8]. Play is so critical to the development of a child that the United Nations High Commission for Human Rights recognizes play as a right of every child[14]. In 2005, the Playwork Principles were established by the Playwork Principles Scrutiny Group and are eight principles that provide the professional and ethical framework for playwork. These principles are widely recognized and are a part of The National Occupational Standards for Playwork. Play is defined by The Playwork Principles as freely chosen, personally directed, and intrinsically motivated[15]. Playwork is an approach to working with children that aims to facilitate the play process through creating and maintaining spaces for children to play [16] . Environment is a vital factor in Playwork, but it has also been shown to play a role in brain growth and plasticity as well[17]. In this way, a relationship between play, brain growth and environment begin to materialize.

A key aspect of play is increased social participation, which facilitates social development [18], [19] . Often children move independently to play and participate socially. Independent mobility has been proven to improve motor, social, cognitive, and perceptual development[20]-[22]. For typically developing children, independent mobility begins as early as they can turn their head by themselves and only increases from there. However, for children with disabilities, mobility is not as simple and easy. For children with developmental delays, self-directed mobility is defined as movement initiated by that individual and may include independent motion such as walking with the use of mobility technology (e.g. gait trainers), standers, and powered mobility devices (e.g. wheelchairs, ride-on toys). Independent mobility

enables these children to choose to interact with parents, peers and even toys, which leads to increased quality of play and thus, development[23], [24].

Because play is characterized as self-motivated and directed, the lack of independent mobility for children is detrimental. If a child does not possess the freedom of mobility, they are at risk of secondary impairments such as cognitive delay and atypical social function[25]-[27]. Children with disabilities progress through the same sequence of play development but play for these children differs in ways such as: limited play repertoires, less time spent in play, reduced language during play, sedentary or passive play, and limited selection of toys. Children with disabilities still have a desire to play, but due to mobility impairments are often bored and distressed when circumstances severely limit their opportunities[28]. Due to the importance of play, it is critical to provide methods of independent movement to children with disabilities.

4.1.3. Adaptive Toys

Assistive technology (AT) is a general term for any technology designed to improve the quality of life for a person with disabilities. Specifically concerning children with disabilities, the Individuals with Disabilities Education Act (IDEA) – American legislation ensuring children with a disability are provided with education tailored to their needs – defines AT as "…any item, piece of equipment, or product system, whether acquired commercially off the shelf, modified, or customized, that is used to increase, maintain, or improve the functional capabilities of children with disabilities."[29] Because self-produced locomotion plays a crucial role in cognitive and psychosocial development, studies have found the benefits of power mobility devices to assist children with disabilities include learning cause and effect, increased environment activity, and improvements in mobility and social functions[30]-[35]. Powered Mobility Devices (PMD), or powered wheelchairs, are a common form of AT and

provide enhanced development of spatial and cognitive functions, increased self-initiated movement during play, and social participation. Unfortunately, pediatric power chairs have limited functional use due to their size, weight, cost, accessibility, ease of transportation, maintenance requirements, and social acceptance. In addition, powered mobility is not often available for children under the age of three due to high costs and limited availability. For these reasons, physical therapists began to look for other options as rehabilitative devices. Modified ride-on toy cars emerged as an option for assistive devices for young children with mobility impairments [1] .

Adaptive ride-on toys were initially introduced by organizations such as Go Baby Go[36] with the intention of encouraging self-directed mobility and social participation[1], [37]. These toy cars were a favorable option due to their cost, accessibility, aesthetics, and adjustability[1], [2], [35]. Most ride-on toys cost less than $400 and are lightweight, small, and easily transported. They are child-friendly with colorful toy designs and various toy functions causing them to be more acceptable to peers. Finally, these toy cars are simple electromechanical devices that can be modified quickly and easily with a range of child specific customizations. For these reasons, adaptive ride-on toy programs have been incredibly successful since their beginning in the early 2010s [1], [38] .

Typically, every toy car requires electrical and mechanical modifications for use by children with disabilities. Modifications can be permanent or temporary based on the child's changing capabilities and family goals. An initial decision must be made, typically by the physical therapist, regarding which car type and size fits the child currently, while also considering future body size and capabilities. Each toy car's original design contains features that inherently fit certain children. Initial basic changes made to the cars include driving and

steering assistance, safety devices, increased support, and functional adaptations. Therefore, from choosing the car, to the modifications made, each decision made is specific to that child. For example, larger vehicles with two seats may be more appropriate, providing more room for children with additional equipment such as an oxygen tank.

The purpose of most driving assistance devices is to help with directional steering and the drive system. Modifications of the steering and drive system can be made separately but are often modified together. In some cases, only the drive system is adapted by the installation of a large, easy-to-press activation switch placed on the steering wheel (see Figure 3). Increased surface area and sensitivity to touch allows for easier activation by the child. However, this requires coordination of the child to press the button and steer simultaneously.

Figure 3: Adapted ride-on toy cars with additional back and arm support, a seatbelt system, and a large, easy-to-press activation switch[39].

Joystick control is one option that allows the child to steer and propel the car at the same time. The steering wheel is removed, and a joystick installed in an appropriate place for the child (see Figure 4). Requirements of a joystick include proper ramp up speed, equal response in all directions (or restricted directional response based on the child's needs) and must have an immediate response to movement of joystick, teaching cause and effect. Sensitivity and directionality of the joystick and speed of the vehicle are determined based on the child's needs. In cases in which a child cannot steer or propel the car, alternative methods are considered such as line following technology. Line following technology allows a vehicle to be directed by a line taped or drawn on the ground. This low-tech method is seen in "magic" toy cars for less than twenty dollars from major superstores and is easily implemented. Therefore, this is a reasonable alternative steering system for children who are unable to control the direction of a vehicle but still benefit from semi-independent mobility[40].

Figure 4: Modified ride-on toy car with driving system replaced with joystick control, foam board to improve back support, and added seatbelt system[41]..

The intent of each modification is to increase safety and functionality. A 5-point seatbelt is a standard addition, as commercially available ride-on toys come with either just a lap belt or no seatbelt at all. The 5-point seatbelt is installed to increase safety and add trunk support. Each car is fitted with a custom seating system intended to augment a child's postural control. Seat modifications use materials such as cushions, swimming kickboards, foam, pool noodles, and PVC pipe to add trunk and head supports that can be adjusted, removed, or replaced. Often removable seat covers are included that can be removed and washed.

4.2. Safety

Considering the success of these adaptive toy programs, there is a reasonable desire by pediatric care providers to maximize the scope of these programs to increase rehabilitation opportunities for children with disabilities. As the reach of these programs extends to a broader pediatric population, it is necessary and morally/ethically responsible to ensure modifications are acting as intended without introducing additional risk of injury. Additionally, while the adaptations made to create an adaptive ride-on toy are intended to provide a safer experience for children with mobility impairments, the aftermarket modification of any commercially available device calls safety into question. As such, if safety concerns exist with the use of unmodified, ride-on toys, this same concern can be applied to the use of these vehicles by children with disabilities in a rehabilitative setting. A review of published literature was conducted to determine if any investigations into ride-on toy safety have been previously pursued.

4.3. Literature Review

Pediatric safety had garnered more attention as of late in the scientific community (e.g. car seat safety, modified ride on toys for children with disabilities, etc.). Despite this recent interest, limitations exist in the knowledge of pediatric safety testing and tolerance thresholds due to a limited amount of test data. Children are unable to volunteer as test subjects and child cadavers are not readily available for research. Often, research is done using anthropomorphic test dummies that model the average child. This presents a problem however because it does not account for children with disabilities. Moreover, in recent years, children with disabilities have seen an increase in opportunities for transportation due to power mobility technologies

and modified ride-on toys. These modified vehicles provide children with disabilities the chance to play and move in their environment. This is a significant advancement for children, because toddlers and preschoolers require independent exploration of their environment for their brain cells and neural connections to develop properly [42]. Around six months of age children will seek to move toward items that capture their interest and engage in independent, self-directed exploratory play [42]. Toys help children participate in these activities and develop the necessary cognitive, social, and motor skills to manage a fulfilling life [43]. This is true, and especially important, for children whose disabilities often keep them from engaging in normal play. As part of these independent, exploratory play activities, accidents are bound to happen and expected as part of their development. Pediatric safety research and regulation concentrates on protecting children from hazards such as toy-related injuries. It has been reported that the annual injury rate of toy-related injuries in children has increased from 1990 to 2011 by 39.9%. This has been highly correlated to injuries caused by ride-on toys, as they account for 34.9% of all injuries [44]. Most injuries caused by ride-on toys are due to falls and inertial impacts. Furthermore, the number and rate of injuries have been reported to peak at two years of age -an age when children are still beginning to learn about their environment through movement and play. In recent years, researchers have engaged in the development of adaptive technology in the form of modified ride-on toys to provide children with disabilities independent self-directed mobility [45]. It is noted that literature review, a ride-on toy was defined as any rideable toy used for the purpose of play. Also, limited work has been done in studying the safety of such devices, and how modifications can worsen or improve the safety of a device in relation to a disability. Most studies have been conducted reviewing toy-related injuries in children without disabilities. One study comprehensively investigated toy-related

injuries from a nationally representative data set collected from 1990 to 2011. This was the first time such a study was conducted. Mechanisms of injury were defined and separated into categories. Children were divided by age as well: five years of age or younger, and age five to 17. Most injuries were found to be caused by ride-on toys and this number increased by 73.7% from 1990 to 2011. It was hypothesized this is due to the increased popularity and accessibility of ride-on toys. These injuries spiked in 2000 and 2001. However, due to increased safety regulations, a noticeable decline in ride-on toy related injuries was observed. This displays the importance and impact of safer design and increased regulation. A few other studies have investigated the susceptibility of children with disabilities to injury. One pooled data from the years 1997–2005 from the National Health Interview Survey, which is a multipurpose health survey completed annually by the United States Census Bureau, for the National Center for Health Statistics. To compare prevalence of injury between children with and without disabilities, a child with disabilities was matched to a healthy child of the same gender and age. It was determined that socioeconomic variables were insignificant. It was also found that children with disabilities experience a higher rate of injury (3.8% vs 2.5%; $P < .01$). It was found that the risk of injury varied by the type of disability such that the more severe the disability, the higher the rate of injury. Another study utilized the China Disabled Persons' Federation (CDPF) to conduct a study on all children with disabilities ages 1–14 [46]. The CDPF maintains a registry database that monitors the number of persons with disabilities and tracks the medical and rehabilitation services provided by the government. Each child with a disability was matched with a healthy child of same gender, age, and living in the same neighborhood for the study. Disabilities were categorized as vision, hearing, speech, physical disabilities, intellectual disabilities, and mental health disorders. Children were also organized

by four levels of disabilities, defined as level four being most severe and nonfunctional, level three being less severe and minimal functionality, level two being semi-functional and level one being the mildest degree of disability with the best functionality. Sociodemographic variables were taken into consideration, including: gender and age of the child, parent's education, family income, single-parent family status, time of being supervised by an adult each day, and total number of family members [46]. Injuries were assessed when an injured child sought medical care at a hospital or community clinic. The rate of injury among healthy children was found to be 4.4% but increased to 9.6% for children with a single disability, and 11.2% for children with multiple disabilities. The level of disability that was injured the most was level two(11.5%), followed by level three (10.4%), level four (10.3%), and level one (8.1%) [47]. Emerging evidence indicates that individuals with disabilities face a significantly higher risk of injury than those without disability.

4.3.1. Systematic literature search and data extraction

Two independent researchers performed a systematic literature search to identify all relevant studies pertaining to pediatric safety and toy-related injuries for children with and without disabilities. Due to the limited number of studies on toy safety for children with disabilities, the search was divided into two sections. Section 1 focused on toy-related injuries. Section 2 focused on inertial impact and injury risk in vehicular collisions. This information is relevant due to its relation to inertial impacts caused by ride-on toys. The following search terms were used: pediatric OR children OR child OR infant AND injury OR injuries OR accident OR trauma AND disability OR disabilities OR disabled OR handicap for the first section. Search terms safety AND inertia impact OR crash OR "car crash" AND ATDs OR cadavers OR "computer simulations" AND "low-speed" AND scaling AND scaling techniques

AND crash analysis were added to the first search terms for the second section. All titles that were relevant to the criteria went through a subsequent screening based on their abstract, and full text articles were reviewed once they were determined appropriate for this study. Studies were excluded when: high velocity impact injuries were surveyed, there was insufficient data concerning children with disabilities, adult participants were included. The figure below describes the process of search and screening.

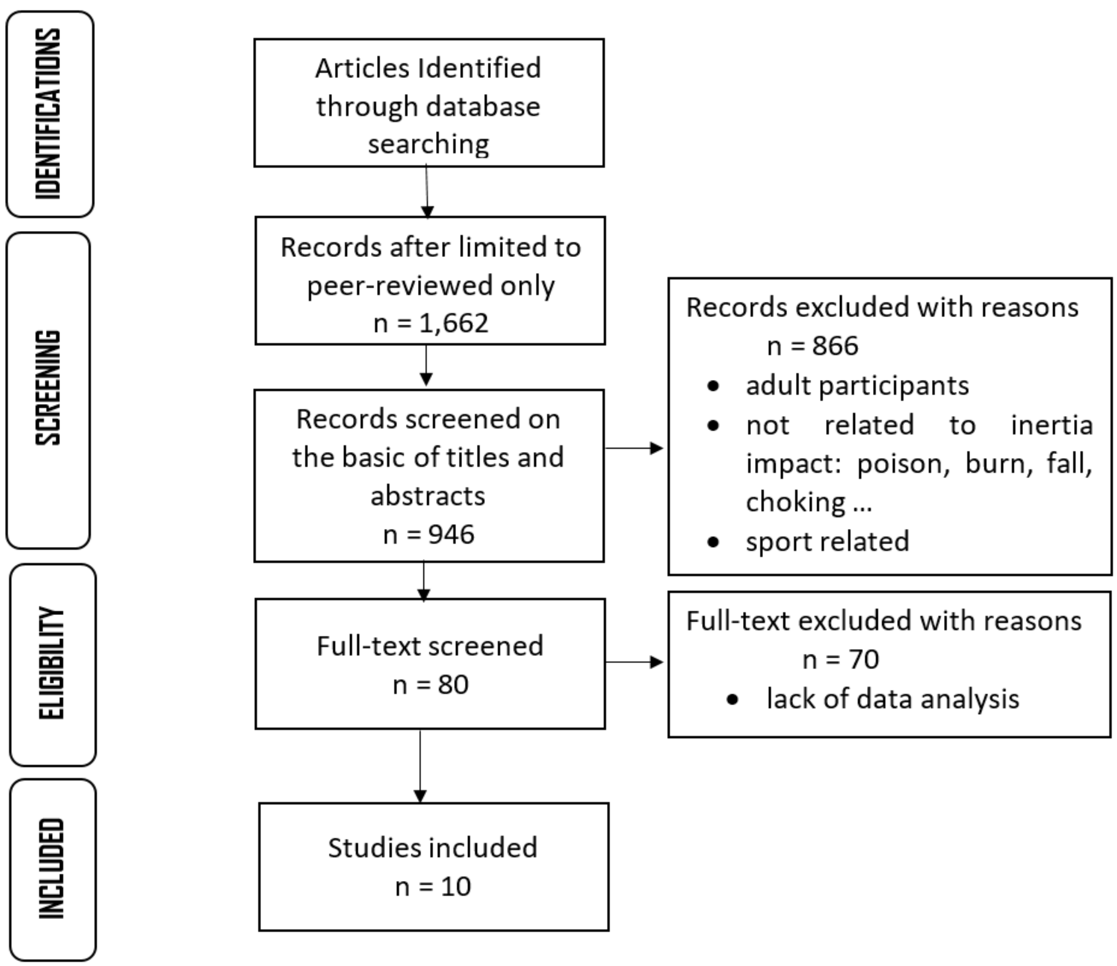


Figure 5: Search process and identification of relevant studies.

4.3.2. Results: Methodology employed in the studies

1.1.1.1. Safety testing by anthropometric test devices (ATDs)

Two studies investigated child safety in motor vehicle impacts by conducting a series of sled tests with 6 and 10-year-old anthropometric test devices (ATDs) using 48–50 km/hr. frontal crash pulse. One study investigated injury risk of a pediatric occupant with a disability sitting in a wheelchair while being transported [48]. The other studied the possibilities of better protection when children were not using booster seats. Methods of increasing protection that were studied include various cushion lengths and varying lap belt geometry [49]. Table 1 shows an overview of the studies. Data related to head, chest, and pelvis acceleration, femur and neck forces, chest compression, chest deflection, and moment were measured during each test. The results from the wheelchair occupant study were then compared with the kinematic limitations and injury criteria of the Federal Motor Vehicle Safety Standards (FMVSS). Comparison to FMVSS 213 (Safety standard that must be met for children car seats to be sold for use. Includes requirements such as the child restraint system must pass a 30 miles per hour frontal sled test that simulates a crash, padding requirements, flammability standards, and buckle release pressure.) and FMVSS 208 (Safety standard for occupant crash protection that establishes performance requirements for passenger vehicles [50]) was used to determine the injury risk of the pediatric wheelchair occupant in a motor vehicle crash [48]. The study that investigated alternative seat belt protection used head excursion, peak knee excursion, the difference between peak head and peak knee excursion, and maximum torso angle to determine whether sitting with a shorter cushion and mid or forward angle lap belt would be better for safety when children are not using booster seats.

Author	Objective & Population	Impact Conditions	Methodology	Results & Recommendations
Gaw,C. Abraham, V. M., Gaw, C. E., Chounthirath, T., & Smith, G. A. (Park & Yoo, 2009)	Comprehensively investigate toy-related injuries among children	N/A	Patients were separated into 2 age categories, younger than 5 years, and 5 to 17 years of age. Mechanism of injury was divided into categories such as falls, collisions and foreign body involvement.	Ride-on toys were 3.19 times more likely to be associated with a fracture or dislocation compared with other toy products. Patients younger than 5 years were more likely to injure their head or neck and face than patients aged 5 to 17. 34.9% of toy-related injuries were associated with ride-on toys.
Sinclair, S. A., & Xiang, H. (Blankenburg et al., 2018)	Verify reports from many researchers that report that disabled children are at a higher risk of injury than non-disabled children. The epidemiology of injury among children with disabilities hasn't been adequately studied.	N/A	Data was pooled from the 1997–2005 National Health Interview Survey (NHIS) The prevalence of injuries in children who had a single disability were compared to children without a disability by gender, age, parent's education, poverty status, and family size. An injury episode was defined as a traumatic event in which the person	It was found that the risk of injury was significantly higher among children with a single disability than among non-disabled children. (3.8%; 95% CI = 3.4, 4.1 vs. 2.5%; 95% CI = 2.5, 2.6, respectively; $P < 0.001$). However, the risk of injury differs by type of disability. The most frequent causes of injury episodes for both test groups were falls. The disability with the greatest probability of injury was

			was injured 1 or more times from an external cause.	children who had a bone, joint, or muscle problem.
Zhu, H., Xiang, H., Xia, X., Yang, X., Li, D., Stallones, L., & Du, Y. Â (AlemdaroÄŸlu, 2017)	Children with disabilities may have a reduced ability to handle environmental hazards because of physical limitations, impairments in mental processing, or in their ability to adjust to their environment.	N/A	The China Disabled Persons' Federation was utilized to survey all registered, disabled children ages 1–14 years. For every disabled child, a non-disabled child living in the same neighborhood and with the same gender and age was matched. Disabilities were organized into categories of vision, hearing, speech, physical, and mental health disorders. Children with multiple disabilities were also taken into consideration. A scale of four varying levels of disabilities was used. Socio demographic variables were also considered.	Rates of injuries among children with a single disability (9.6%) and multiple disabilities (11.2%) were significantly higher than that among children without disabilities (4.4%). It was found that age of the child, children in single parent households; children whose parent's highest education was middle school or less; children with less than 30% of time per day supervised by an adult; and children whose family income per month was less than 1000 RMB has little to no change on rate of injury. Level 2 of disability was injured the most (11.5%), followed by level 3 (10.4%), then level 4 (10.3%), and lastly level 1 (8.1%).

Ha D. (Ginsburg, 2007)	6-year-old children with disabilities 3 pediatric manual wheelchairs	20 g/48 km/h front crash pulse Wheelchair impact	Sled test using a seated Hybrid III 6-year-old ATD Head acceleration, chest acceleration, pelvis acceleration, femur forces, chest deflection, neck forces, and moment were measured.	Test results were compared with kinematic limitation and injury criteria that listed in the ANSI/RESNA WC-19, FMVSS 213 and FMVSS 208 standards.
Klinich KD (Kleinberger et al., n.d.)	6-year-old and 10-year-old healthy children Not using belt-positioning booster	Sled pulse delta velocity is 28.8 km/hr.	Sled test using a seated Hybrid III 6-year-old and 10-year-old ATD Vehicle seats Cushion length of 450 mm Cushion length of 350 mm Belt Geometry Lap belt angles tested: 23° (rear), 50° (mid),and 70° (forward)	Compared kinematic outcomes between long and short cushion length and increasing lap belt angles.
Park D (Zhu et al., 2014)	Healthy 3-year-old children Existing three-point belt-type child seat	50 km/h front crash pulse Children car seat impact	Combined sled test and computer based simulations a Standard crash test: velocity increased to 50 km/h then suddenly decelerated Simulation:	Compared sled test and computer simulation results to validate data collected. Developed an advanced new type of a child seat based on the results (six-point belt-type)

			Geometric modelling: LS-DYNA, CATIA Preprocessor: FEMB, and postprocessor: LS-POST Finite elements	
Isabelle S. (Sinclair & Xiang, 2008)	Compare the kinematic response of children and child anthropomorphic test devices (ATDs) during emergency braking events in different restraint configurations in a passenger vehicle 16 healthy children aged 4 to 12 Q3, Hybrid III (HIII) 3-year-old, 6-year-old, and 10-year-old ATDs	2 braking events Vehicle brakes as fast as possible to a full stop while traveling at a velocity of 70 km/h The maximum deceleration of all analyzed braking events was 1.2 g. The peak mean deceleration was 1.0 g with a standard deviation of 0.08 g and duration of 1.8 s The duration of the entire deceleration period was 2.4 s 2 sharp turns to the right in each restraint system	Child volunteer and ATDs test Short children (stature 107–123 cm) and the Q3, HIII 3-year-old, and 6-year-old were restrained on booster cushions as well as high-back booster seats. Tall children (stature 135–150 cm) and HIII 10-year-old were restrained on booster cushions or restrained by 3-point belts directly on the car seat. Restrained on the right rear seat of a modern passenger vehicle. Four small video cameras (Monacor TVCCD-30, lens focal length 3.6 mm, Monacor International, Bremen, Germany) were affixed inside the vehicle providing a front view of	40 trials were analyzed Child volunteers had greater maximum forward displacement of the head and greater head rotation compared to the ATDs. The average maximum displacement for children ranged from 165 to 210 mm and 155 to 195 mm for the forehead and ear target, respectively. Corresponding values for the ATDs were 55 to 165 mm and 50 to 160 mm. The change in head angle was greater for short children than for tall children. Shoulder belt force was within the same range for short children when restrained on booster cushions or high-back booster seats. For tall children, the shoulder belt force was greater when restrained on booster cushions compared to

			the child, a perpendicular lateral view, and 2 oblique views of the children volunteer. The recording rate was 12.5 frames per second. Data collected included vehicle velocity, acceleration in longitudinal and lateral directions, and brake pressure. MATLAB was used to analyze data.	being restrained by seat belts directly on the car seat
Jingwen H. (Gaw et al., 2015)	Analyses of crash injury data have shown that injury risk increases when children transition from belt-positioning boosters to the vehicle seat belt alone. Investigate how to improve the restraint environment for these children. Healthy children aged 6 to 12 years old	Frontal crash test	Used a parametric child ATD MADYMO model To scale the baseline child ATD model into different body sizes, custom software was developed by combining MADYMO Scaler and a program written by Scilab V5.2.2 (Scilab Enterprises, France) An automated computer program was developed using a combination of MADYMO (TASS, The	The maximum head and knee excursions in this parametric study were 639 and 833 mm, respectively. Both were below the limits defined in FMVSS No. 213, in which head excursion should be less than 720 mm and knee excursion should be less than 915 mm. Lower and more rearward D-rings (upper belt anchorages), higher and more forward lap belt anchorages, and shorter, stiffer, and thinner seat cushions were associated with

			Netherlands), Scilab, and ModeFRONTIER (ESTECO, Italy) to integrate the parametric child ATD model, ATD positioning procedure, automatic belt fitting algorithm, and other crash conditions together. A 200 N force was applied to the 3 belt anchorages	improved restraint performance. Children with smaller body sizes require more-forward D-rings, inboard anchors, and outboard anchor locations to avoid submarining. However, these anchorage locations increase head excursions relative to more-rearward anchorages.
Florian F. (Logan et al., 2016)	Case study: a serious accident involving two passenger cars took place in Austria in which three children seated in the rear were fatally injured in a frontal collision. The study was performed to gain a better understanding of rear occupants injury mechanisms and potential improvements to rear-seat restraint system 3 children: 5 years	Frontal collision EBS (equivalent barrier speed) and EES (energy equivalent speed) is 62 km/h Approaching angle of 85â—¦ The approaching velocity of the VW was calculated to be 63 km/h For the Nissan, a velocity of 69 km/h was determined	An HIII (hybrid III) six-year-old dummy (hereafter HIII 6yo) was used for simulating the youngest child, aged five, seated behind the driver. The eight-year-old child, who was seated in the middle, was simulated by a TNO P10 dummy (hereafter TNO P10) For the eldest child, aged 10, an HIII 5th percentile dummy (hereafter HIII 5th) was used An HIII 50th percentile dummy was seated in the driver's seat.	Results: The HIII 5th seated in the rear showed a considerable chest (52 mm chest deflection, 66 g chest acceleration) and head load (HIC [head injury criterion] = 1047 and acceleration exceeding during a cumulative time interval of 3 ms [cum3ms] = 96 g). The shoulder belt forces reached almost 9 kN ◦ The chest deflection in the HIII 6yo and HIII 5th only slightly exceeded the threshold values of 40 mm and 52 mm In contrast, the loads on the HIII 5th seated in the front seat

	old, 8 years old and 10 years old		An HIII 5th percentile dummy was situated on the front occupant's seat to enable direct comparison of the restraining effect between the front and the rear compartments A crash test was used for validating a numerical model of the rear compartment, programmed with the multi-body (MB) simulation code MADYMOR. The MADYMOR model was used for a set of parametric variations	were consistently lower compared with those on the rear-seated HIII: head acceleration was 25% lower, neck forces and torques were considerably lower (by 25–40%), chest deflection was 25% lower, shoulder-belt forces were 12% lower and chest acceleration was 15% lower. Furthermore, the shoulder belt in the front seat had a 50% greater pullout (100 mm) Recommendations Provision of mandatory seatbacks with side wings to protect against lateral impact. Provision of a mandatory guide for shoulder belts. Mandatory introduction of anti-rotation devices, e.g., top tether and outrigger. Definition of maximum size of not-ISOFIX seat (geometry envelope). Identification of CRS, including the weight, size and age of the child for which the specific model is designed.

Table 1:Overview about the studies' objectives, impact conditions, methodologies of reviewed pediatric safety studies

1.1.1.2. Safety testing by ATDs and computer simulations

One study used the combination of sled testing and computer simulations to develop an advanced child restraint system (CRS)[51]. A sled test was first performed using a 3-year-old ATD in an existing three-point seat belt CRS with the objective of achieving head and chest accelerations within safety limits. The crash test was designed to exert accelerations according to national standards, and increased velocity to 50 km/h and then suddenly decelerated. A dynamic simulation was then conducted using a commercial LS-DYNA® program developed by Livermore software Technology Corporation. LS-DYNA® was used for contact and collisions, and the computer-aided three-dimensional interactive application (CATIA™) program was used for geometric modeling. Once the sled test and computer simulation results were matched, a new type of child seat was developed. An optimization sequence was applied to determine the thickness of each part to decrease the weight. A new six-point CRS was then developed using LS-DYNA®. Once the final result was obtained from the computer simulation of the new design, the sled test was carried out with the developed prototype of a six-point child seat [51]. An overview is shown in Table 1.

4.3.3. Results: Main findings of the studies

1.1.1.3. Safety testing by anthropometric test devices (ATDs)

Sled tests were conducted under 48 km/h and 20 g average impact conditions on children riding in a motor vehicle while seated in a wheelchair. This study showed that a 6-year-old seated in a wheelchair may be at risk of neck injury during a frontal car crash and concluded that variations in the shoulder belt anchor point led to variance in restraint effectiveness. All tests conducted in this study exceeded the tension extension limit. All tests complied with the

requirement that the wheelchair not load the ATD. None of the tests exceeded the limit which evaluates the integrity of seat surface and seat attachment hardware and none of the tests exceeded the maximum chest acceleration limit of 60 g. The results of the safety testing satisfied the head injury criterion (HIC) of 700 which measures the amount of damage to the head. Chest deflection for two iterations of the test were at the limit of 40 mm specified in regulation. The first and second sled test exceeded the peak neck tension force limit of 1490 N. No tests exceeded the independent compressive neck force limits [48]. When cushion size and lap belt angle were tested, increased cushion length and bigger lap belt angles improved children safety in a motor crash. However, seat boosters still have the best child safety performance than simply increasing the cushion length and lap belt angle.

1.1.1.4. Safety testing by ATDs and computer simulations

While the study failed to match the sled test results to simulations results exactly, the collected data and magnitudes at the peak value were comparable. Based on the resulting similar trends, it was concluded that the simulation sequence was suitable to develop a new child seat. The design of a six-point belt-type child seat was carried out resulting in a lightweight design to save material and manufacturing cost. However, such a lightweight design compromises the safety of the seat. Simulations varying the thickness of the material was carried out using ANSYS (A computer simulation software) to explore proper thickness vs safety tradeoffs. Results yielded a final design having 64.5% of its original volume. Once computer simulations were performed for the new six-point CRS, the sled test was carried out indicating that a six-point CRS provides a lower impact force due to the force being distributed over an increased area[51].

4.3.4. Discussion

1.1.1.5. Safety testing by anthropometric test devices (ATDs)

Children with disabilities often differ anatomically from children without disabilities, and therefore are often required to remain seated in their wheelchair while being transported in a motor vehicle. Injury risk of a child, seated in a manual pediatric wheelchair, was analyzed in this study using frontal impact sled testing. A 6HybridIII ATD was used, which models a non-disabled 6-year-old child with normal muscle tone and balance. Therefore, a child with disabilities may be more susceptible to severe and fatal injuries in circumstances where a child without disabilities would not be injured[48]. This study showed that a 6-year-old seated in a wheelchair may be at risk of neck injury during a frontal car crash. The study also concluded that variation in the shoulder belt anchor point led to variance in restraint effectiveness. It was hypothesized that chest deflections would have been higher if the shoulder belt had been at a more optimum anchorage point [48].

1.1.1.6. Safety testing by ATDs and computer simulations

Dynamic simulations of a child seat were carried out using LS-DYNA® to develop an advanced CRS design. Simulation results for a six-point belt-type child seat were compared to sled testing concluding that LS-DYNA® is a suitable alternative to replace sled testing, reducing cost and time for new product development [51]. However, it is acknowledged in this study that precise material properties are needed for accurate results.

4.3.5. Conclusion

Studies have been carried out on the multiple aspects of toy-related injuries and the susceptibility of children with disabilities to injury. However, a gap in the literature occurs concerning the susceptibility of children with disabilities to toy-related injuries, specifically in

relation to ride-on toys and the repercussions surrounding such injuries. It is theorized that such lack of data is due to the difficulty and costs associated with experimental validation. Hence, it is recommended that computer simulations be used to provide preliminary data analysis. Various aspects of small inertial impacts on a child with disabilities could be drawn from these studies. Furthermore, safety recommendations for ride-on toy modifications could be derived from such simulations and these could be correlated to specific disabilities. Ultimately the goal of such work would be to draw specific guidelines regarding modifications of ride-on toys and children with disabilities

4.4. First Study Intro

Within this context, the purpose of the first study was to determine whether modifications to ride-on toys are sufficient to prevent common modes of injury such as falls, passenger excursion, and impact with the interior of the vehicle as well as how these modifications influence kinematic and kinetic injury metrics. Specifically, we seek to evaluate the effects of various seatbelt configurations (no belt, lap belt only, and 5-point harness), and determine how increased seat back height affects neck forces. These modifications were chosen because they are a standard modification made to all adapted ride-on toy cars. This evaluation will provide physical therapists and engineers with the data to make informed decisions when adapting ride-on toys for the benefit of children with disabilities.

Options for safety testing methods concerning the pediatric population are limited because children, especially those with disabilities, are unable to volunteer as test subjects. Furthermore, testing with live occupants, cadavers, and pediatric anthropomorphic test dummies (ATDs) all have significant limitations. Live occupants present ethical safety

concerns, especially when testing with children as test subjects. Pediatric cadavers are not readily available [3]. ATD's are costly and only represent the average population. Therefore, computer simulations represent the most viable alternative and were chosen for this study.

4.5. Second Study/Joint stiffness Intro

In the previous study, the effects of adding a seatbelt and back/neck support on safety were examined. However, this study did not take into consideration physical and developmental variations that exist across pediatric populations, particularly as can be seen in children with disabilities. Rather, this initial study investigated the effects of vehicle interior and safety modifications using only a 50^{th} percentile 6-year-old (6YO) Hybrid III ATD model. Because this model only considers the average 6YO, it does not necessarily represent the response behavior of a child with disabilities (e.g. size, geometry, and muscle development and spasticity). It is important to take the intended population into consideration if we are to properly examine the safety and effectiveness of modifications made to ride-on toys for children with disabilities.

4.5.1. Joint Stiffness Section

Lesions to the central nervous system (seen in conditions such as cerebral palsy, multiple sclerosis, motor neurone disease, etc.) can cause both positive and/ or negative upper motor neurone syndrome (UMNS) features (e.g. increased tone, spastic dystonia, released reflexes, motor weakness, loss of dexterity and motor control.)[52]-[54] Therefore, children who suffer from these musculoskeletal disorders can have impaired voluntary muscle function, abnormal muscle tone and increased spasticity.

Individuals with spasticity experience a velocity-dependent increase in muscle tone to passive movement, causing an inability to stretch muscles or coordinate movements effectively [53]. Studies indicate that spastic muscle has shown a higher stiffness than in control, non-spastic, muscle. In addition, this study found change to the extracellular matrix of muscle contributes to increased stiffness of muscle contracture. The extracellular matrix (ECM) is an intricate network of macromolecules linking together to form the structure of muscles and contributes to the mechanical properties of tissues. Increased stiffness corresponds to an increase in collagen content, a protein that makes up the majority of the ECM. This study concluded the increase in passive tension that causes spastic muscle contracture is due to high stiffness in the extracellular matrix and increased sarcomere length in muscle[55].

Typically, muscles exist in an optimized state to maximize the force produced during contraction. Force produced during contraction is modulated by the sarcomere (a structural unit of muscle). If the sarcomere becomes too long, there will be insufficient overlap of myofilaments and less force will be produced, negatively affecting muscle tone and contraction. Muscle tone assists in maintaining posture and balance and reflex generation by maintenance of partial contraction of the muscle and is disrupted by muscular disorders[55].

Hybrid III dummies are created with intentionally increased muscle stiffness to represent a human bracing for impact. However, impaired voluntary muscle function disrupts the ability to brace for impact. Therefore, decreasing the model's stiffness can model an unbraced reaction to collision, and therefore a child with impaired voluntary muscle function.

Because the population of children with disabilities who are receiving adaptive ride-on toys has a wide range of mobility impairments and may suffer from a wide range of musculoskeletal

disorders, those with both decreased and increased muscle stiffness must be considered. Within this context, we chose to use the 6YO Hybrid II model with varying muscle, or joint stiffness.

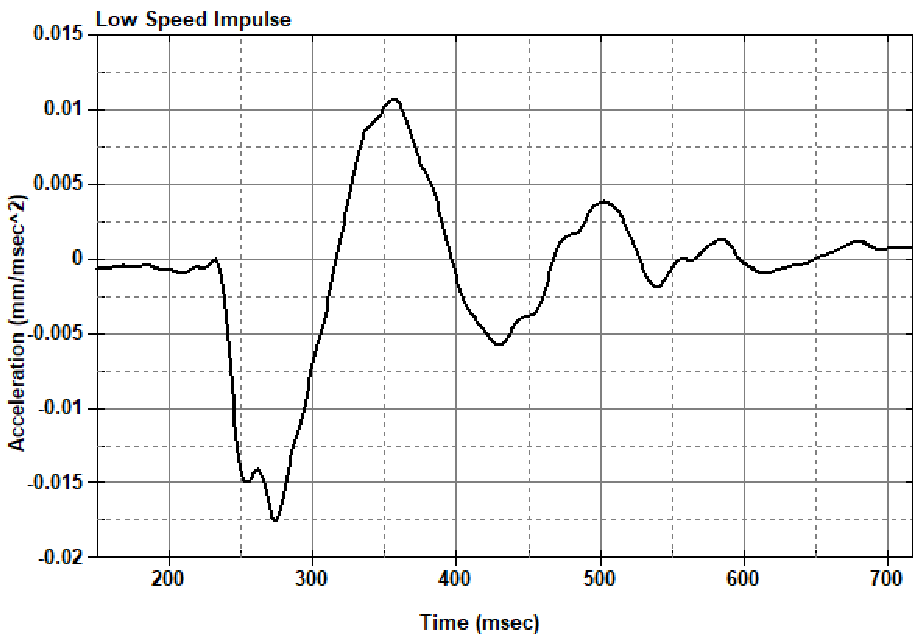
5. Study 1

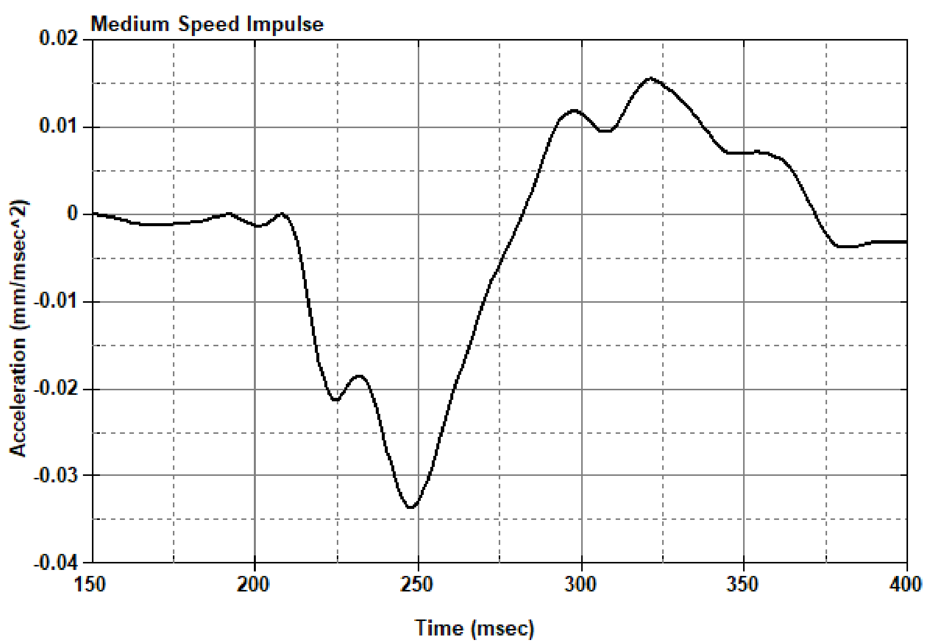
The study consisted of simulating a frontal collision using various parametrically altered versions of a powered ride-on toy car containing a Hybrid III finite element (FE) model. Two Hybrid III models representing children exist: a 50th percentile three-year-old (3YO) and a 50th percentile six-year-old (6YO). As many children do not receive ride-on toys until they are four to six years old, the six-year-old model was chosen as it best represented the population of focus for this study. Two versions of seat geometry were analyzed: the first represented the unmodified ride-on toy as it is commercially available, the second was modified to have increased seat back height. Three levels of restraint configuration were analyzed, including: 1) no seatbelt, 2) a lap belt, and 3) a 5-point harness. The computer program LS-DYNA was chosen as the simulation tool because it is capable of highly non-linear, transient dynamic finite element analysis using explicit time integration.

5.1. Impulse

Crash testing was conducted using a commercially available ride-on toy car (12 V-powered Mercedes AMG G63 ride-on battery powered car) to acquire experimental frontal impact data to supply as an acceleration impulse to the FE model. A 12 V car was chosen because it was appropriate for the size of the live occupant being used. Additionally, 12 V cars are often used because they accommodate any additional equipment that children with disabilities might have (e.g., oxygen tank etc.). The acceleration impulse was acquired using a live occupant of similar weight to that of the 6YO Hybrid III dummy (~50lbs). Three impact speeds were tested, as

there were three speeds available on the remote control of the vehicle. The reported speed range is ~2.5–5 km/h per the vehicle manual. The low, medium, and high speeds were experimentally found to be 2.2 km/h, 3.6 km/h, and 4.9 km/h respectively. Prior testing was conducted to determine the distances at which the car reached steady state speed, and an additional five feet was added to ensure a constant velocity. For the low-, medium and high-speed impacts, the car impacted a wall from 10 feet, 15 feet, and 25 feet, respectively. A printed circuit board (PCB) Piezotronics, Inc. (SN 115452) accelerometer was placed in the center of the floorboard of the car and operated at 10,000 Hz per the Society of Automotive Engineers (SAE) standard J211- 1 [56]. In addition, high-speed photography was used to capture each test. The vehicle was driven remotely into a wall at each of the three speeds, three times, for a total of nine tests. The data was filtered using MATLAB and cropped to only include the impulse and the vehicle's return to zero acceleration. The crash pulse was then exported to Microsoft Excel and imported directly into LS-DYNA. Only the forward movement impulse (y-direction) was used in the simulation since the x- and z-directions were negligible. For each speed, the most severe case (in terms of peak acceleration) was used in the FE simulations (Figure 5).


Low Speed Impulse
0.015
0.01
0.005
0
-0.005
-0.01
-0.015
-0.02
Acceleration (mm/msec^2)
200
300
400
500
600
700
Time (msec)


Medium Speed Impulse
0.02
0.01
0
-0.01
-0.02
-0.03
-0.04
Acceleration (mm/msec^2)
150
200
250
300
350
400
Time (msec)

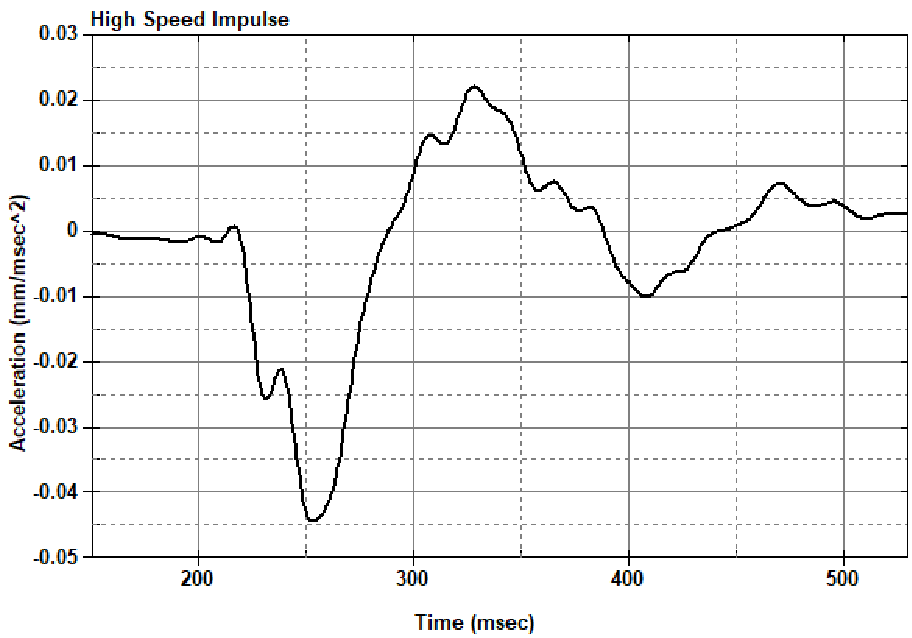


Figure 6: Frontal impact acceleration data acquired during crash testing.

5.2. Occupant Modeling

The FE models, which replicated the interior of a ride-on toy car, were created using shell elements and rigid material properties. The sleds were composed of two parts: the seat and the dashboard. Both models only varied in the height of the seat back; the modified seat model had a back height 320 mm greater than the unmodified model. The seat back modification was determined based on the height differential of the seat and the child's head and was extended to prevent hyperextension. Four seatbelt conditions were analyzed in the simulation, including unbelted, a lap belt only, and two variations of the 5-point harness. The variations in the 5-point harness were motivated by observations that installation of these belts is not always uniform due to the variability of the vehicles, as they are customized for each child. As geometry of the cars and children vary, so does belt placement. It is recommended that belt length should be as short as possible between mounting points, which is commonly regarded as best practice to prevent injuries [57]. However, harness attachment points may vary as the

cars are modified to be usable by children of varying size or to accommodate growing children, and therefore have shoulder attachment points located above the height of the child's shoulders. Therefore, two levels of shoulder attachment points were modeled: directly above the child's shoulders, as is suggested, and at the top of the extended seat, as some cars are modified to have. The sleds were constrained in all directions except the y-direction. The six-year-old Hybrid III model was then placed in an appropriate seated position in both vehicle environments. Automatic surface-to-surface contact was used to define contact between the seat and the axial skeleton, and between the feet and floorboard. Gravity was applied to the system for 150 milliseconds to allow the human model to settle into the seat, after which a load curve was applied using one of the three acceleration impulses from testing. The sled was constrained to a massless node, which serves as a common reference point, and a prescribed motion was applied using the massless node and the load curve. The lap belt and 5-point belt were created as a mixed belt (a belt containing both 1D seatbelt elements and 2D triangular, shell elements). The material of the 1D belts was defined using *MAT_SEATBELT and used loading and unloading curves. The material of the 2D shell component of the seatbelt was defined using *MAT_FABRIC with material values found experimentally for modeling seatbelts in LS-DYNA shown in Figure 6 [58]. Contact between the dummy and the seatbelt was defined between the axial skeleton and the belt using automatic surface-to-surface contact. The final seven models are shown below.

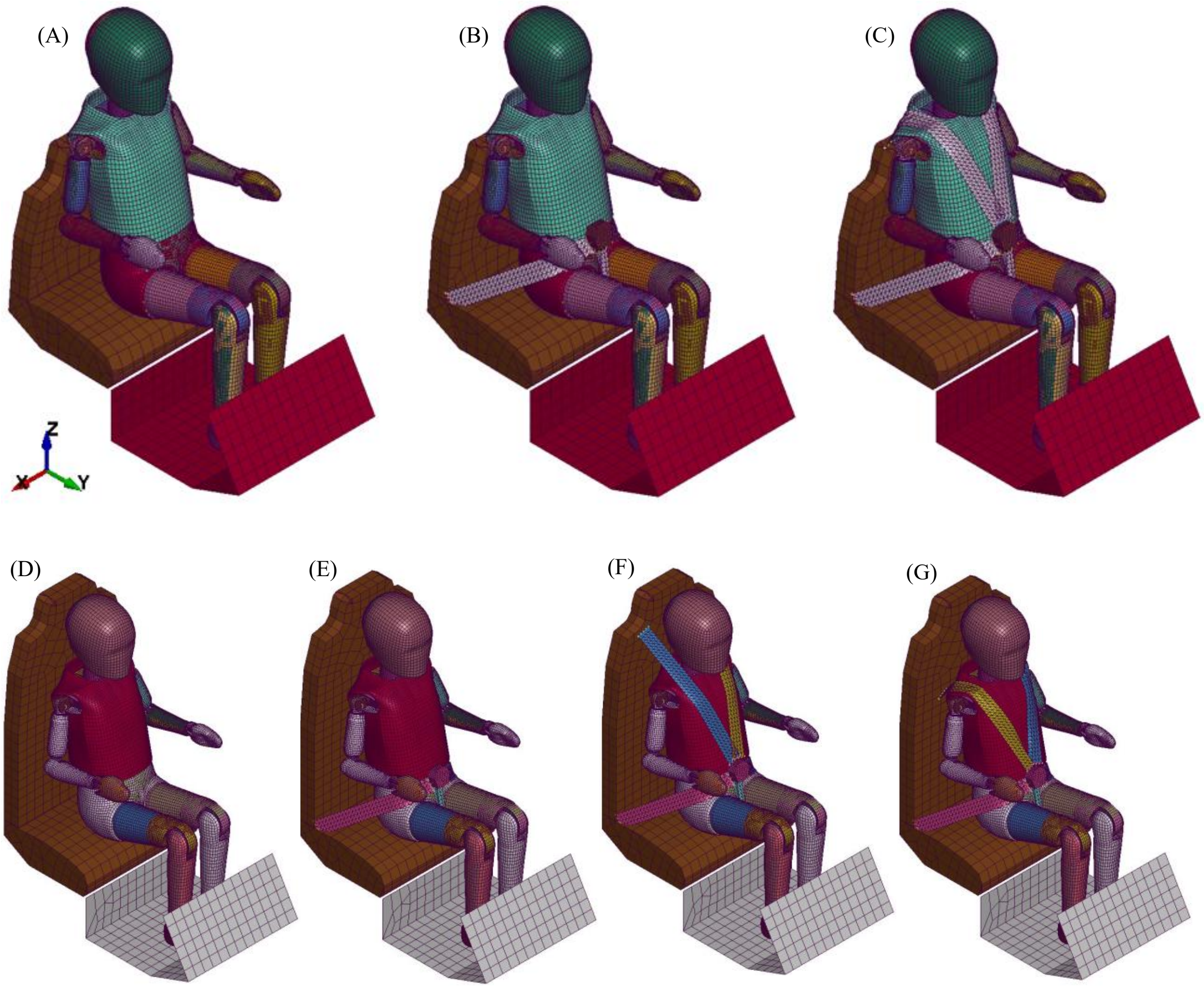


Figure 7: Occupant in unmodified model with: no seatbelt (A), lap belt (B), and 5-point harness (C). Occupant in modified model with: no seatbelt (D), lap belt (E), 5-point harness with high attachment (F), 5-point harness with low attachment (G).

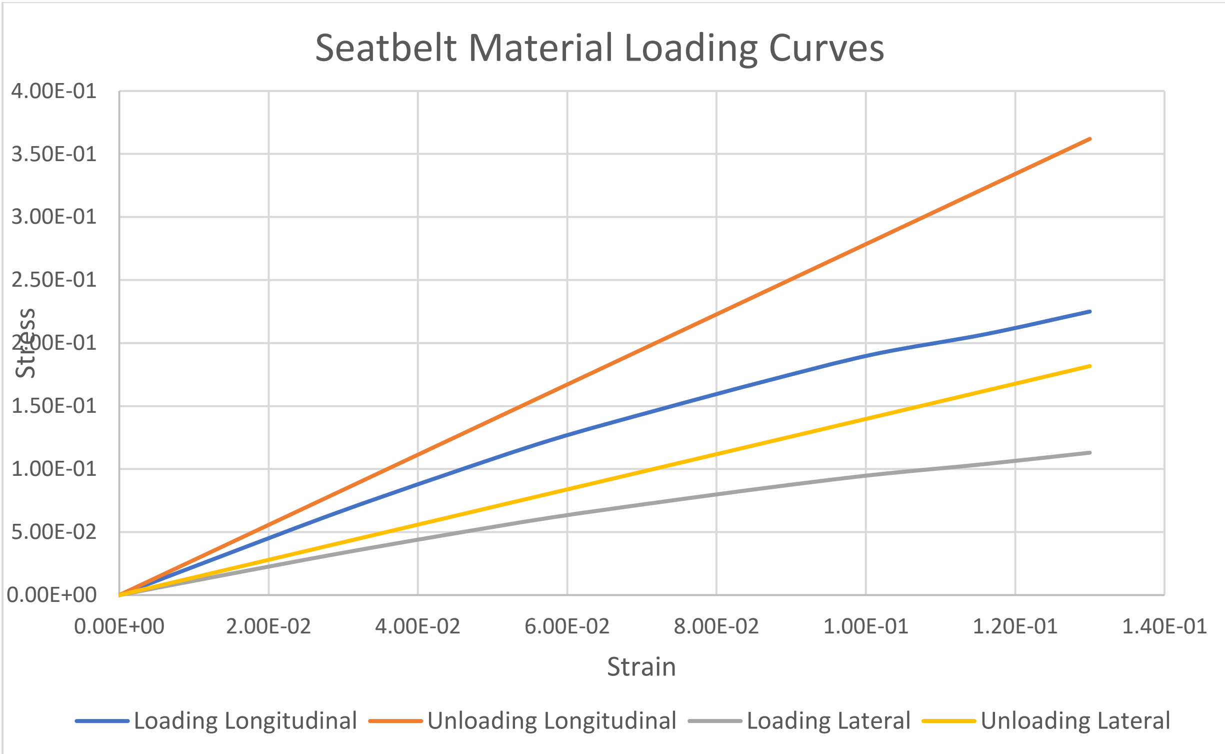


Figure 8: Average seatbelt values experimentally derived [40]

5.3. Post-processing

The following injury metrics were calculated for each simulation: head injury criterion (HIC), peak angular acceleration (PAA), peak linear acceleration (PLA), head and chest displacement, neck tension force, neck injury criterion (Nij), neck bending moment, and neck transverse shear. For these metrics, a higher value results in a higher risk of injury. HIC measures the likelihood of head injury arising from an impact [59]. Figure 8 is used to interpret HIC scores and shows the risk curve of sustaining a level 1 injury on the abbreviated injury scale (AIS1). An AIS1 injury is the most minor injury one can sustain and has a zero percent

probability of death [60]. Neck injury criteria is used to predict neck injuries in low speed rear-end collisions and risk of injury increases as the value approaches one [59]. Peak linear accelerations that are experienced in activities of daily living are typically on the order of ten g's [61]. Displacement is considered because impact with interior of the vehicle poses one of the greatest risks of injury.

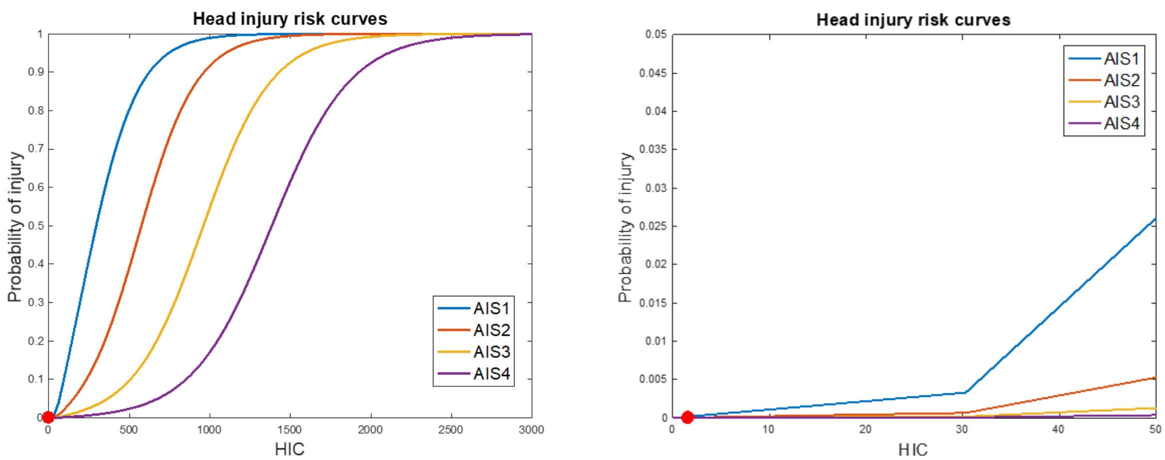


Figure 9: Head injury probability curves based on HIC values. The figure on the right is an inset of the figure on the left. The red dot indicates the maximum HIC value

5.4. Results

Results for all kinetic and kinematic outcomes are provided in Table 2. These data indicate a uniform decrease in chest displacement with the addition of a lap belt, and further decreases with the addition of a 5-point harness. Head displacement subtly increased with the use of a lap belt but decreased in all cases with a 5-point harness. From an absolute perspective, injury metrics were very low in all test conditions. In most cases, injury metrics tended to increase by modest amounts with the addition of any restraint condition. However, this trend was not uniformly present – several injury metrics decreased modestly and some increased markedly with the addition of restraints.

		Speed	HIC15	PAA (rad/msec^2)	PLA (g's)	Head Displacement (mm)	Chest Displacement (mm)	Neck Force Tension (kN)	Nij	Neck bending moment (kN-mm)	Neck Transverse Shear (GPa)
Unmodified seat	No Seatbelt	High	0.192	0.036	5.546	179.0	159.6	0.113	0.038	0.448	0.287
		Medium	0.181	0.026	3.947	154.5	134.7	0.079	0.025	0.429	0.284
		Low	0.176	0.017	4.387	94.9	105.9	0.037	0.014	0.471	0.287
	Lap Belt	High	0.332	0.071	10.543	184.0	138.9	0.148	0.055	0.275	0.343
		Medium	0.174	0.062	2.492	157.4	122.7	0.109	0.037	0.465	0.284
		Low	0.185	0.016	5.061	94.9	101.7	0.053	0.019	0.128	0.284
	5-point belt	High	1.599	0.083	4.779	156.0	124.5	0.377	0.136	1.005	0.285
		Medium	0.213	0.048	4.253	137.7	115.6	0.109	0.037	0.465	0.284
		Low	0.182	0.034	3.290	93.1	101.7	0.063	0.023	0.196	0.282
Modified seat	No Seatbelt	High	0.314	0.039	10.072	202.6	184.0	0.113	0.044	0.563	0.361
		Medium	0.209	0.043	8.979	207.9	148.2	0.078	0.031	0.354	0.305
		Low	0.204	0.035	2.817	164.4	131.2	0.061	0.023	0.280	0.128
	Lap Belt	High	0.861	0.114	11.434	246.0	150.2	0.308	0.108	0.443	0.338
		Medium	0.205	0.054	5.215	210.4	135.5	0.111	0.043	0.470	0.188
		Low	0.250	0.042	9.568	169.0	127.1	0.099	0.039	0.402	0.335
	5-point belt (High)	High	1.307	0.122	12.603	171.1	121.7	0.329	0.117	0.417	0.147
		Medium	0.204	0.063	4.903	157.3	114.3	0.108	0.038	0.434	0.125
		Low	0.206	0.051	7.878	136.6	108.3	0.074	0.028	0.468	0.282
	5-point belt (Low)	High	1.794	0.109	11.303	154.3	117.6	0.315	0.110	1.115	0.258
		Medium	0.603	0.113	8.021	145.4	114.0	0.215	0.076	0.602	0.182
		Low	0.211	0.043	4.554	132.4	108.7	0.083	0.023	0.934	0.178

Table 2: Results from the safety analysis of modifications made to a battery-powered ride-on toy car using the Hybrid III six-year-old ATD model

For unbelted conditions, displacements were typically larger for the modified seat versus the unmodified seat. Overall, displacements decreased with the addition of seatbelts, and the most significant decrease was seen with the addition of a 5-point harness, regardless of if it had a high or low-attachment point. The modified seat with a 5-point harness with low attachment points decreased displacement most. This was expected as this is the recommended use of a 5-point harness, however in some cases it is not possible to install the harness in this way. The results show that while a low attachment point is best, the addition of a 5-point harness sufficiently reduces displacements regardless of whether it is attached directly above the shoulders or higher on the seat. Therefore, while it is recommended to connect the harness to the vehicle at the lowest point possible, if it is not possible given the situation, it is acceptable to attach the harness at a higher connection point. HIC15 scores tended to increase with the addition of a seatbelt. In both modified and unmodified seat models, at low and medium speeds, addition of the lap belt and 5-point harnesses resulted in minor increases of HIC15. The HIC15

scores only increase significantly at high speed. However, the most severe HIC15 score (high-speed 5-point belt with low attachment) was 1.794. The greatest PLA experienced was 12.6 g's with the 5-point belt with a high attachment at high speed. All high-speed impacts resulted in similar PLA values. Medium and low speed resulted in lower PLA values. The PLA for the lower attachment of the 5-point harness on the modified seat was less than that with a high attachment point for high-speed and low-speed impacts but was higher for medium speeds. PAA increased in nearly every instance with increasing occupant constraint. For an unmodified seat, neck bending moment decreased at low speed for both the lap belt and 5-point belt but experienced an increase at high speed with the 5-point belt. Neck shear decreased minimally in almost all cases with the addition of seatbelts, with the high-speed lap belt being the only condition to increase neck shear. A modified seat resulted in lower neck bending moments. The 5-point harness with a modified seat showed the best improvement regardless of the attachment point.

5.5. Discussion

As anticipated, the 5-point harness with low attachment points on the modified seat was the best performer in terms of decreasing chest and head displacement, most notably for medium- and high-speed impacts. While the lap belt best functioned to reduce chest displacement, it was not effective at preventing forward displacement of the head.

The addition of a seatbelt prevents injurious contact with interior components as belt restraint provides a mechanism for earlier and more effective deceleration of the body. However, seatbelts can alter magnitudes of injury metrics from inertial loading as they restrict displacement of the torso, which has been observed to be associated with an increase head acceleration in the context of motor vehicle collisions. Despite the changes to injury metrics

observed in our simulations, all calculated values were small in comparison to known injury tolerance thresholds (e.g., studies show peak linear accelerations on the order of ten g's can be experienced in activities of daily living [61]. All recorded HIC values were extremely small when compared to AIS injury thresholds (Figure 8) [62]. The maximum HIC score recorded, 1.794, corresponds to a probability of an AIS1 injury of 3.6×10^{-35}. An AIS1 injury is considered to be a minor injury with the possibility of superficial lacerations and a zero percent probability of death. A 3.6×10^{-35} probability of an AIS1 injury is an extremely low probability of even a minor injury occurring.

A notable limitation to this current study is that occupant contact with interior components was not simulated (other than contact with the seat and restraint system). However, commercially available toys include a steering wheel, and common functional adaptations to ride-on toys include items such as joysticks or switches that can protrude even farther from the dash of the vehicle. While this is a necessary addition to assist in mobility control, this substantially increases the probability of contact with the interior of the vehicle and decreases the allowable chest displacement. Occupant contact with such interior components could increase the forces and accelerations experienced during a frontal collision, thereby mitigating differences that were observed in injury metrics between restrained and unrestrained simulations. To investigate this effect, a model was created by adding a simple joystick to the modified seat model without a seatbelt (shown in Figure 9). The joystick was placed 50 mm from the ATD's torso and aligned with the midline of the vehicle to replicate typical placement of the joystick. During a high-speed frontal impact simulation, the ATD impacted the joystick and all injury metrics increased. PLA increased by 133% (to 12.765 g's), higher than all other PLA values. Head displacement increased more significantly than chest displacement, as the

joystick impact increased the bending moment. These findings indicate that in situations where an occupant may be susceptible to contact with interior components, seatbelts represent a mechanism to not only reduce occupant displacement, but also to mitigate the magnitude of forces and accelerations. A future area of study would be further investigation into contact with interior components with the addition of multiple types of restraint harnesses. Another limitation to this study was the 12 V-powered Mercedes AMG G63 ride-on battery powered car and 6YO Hybrid III model that was used. The application of adaptive ride-on toy extends to occupants of varying size and weight; additionally, vehicles of varying size, weight, and battery power are used. Therefore, an area of future study would be to include the use of the 3YO Hybrid III model and collect crash data from varying ride-on toys.

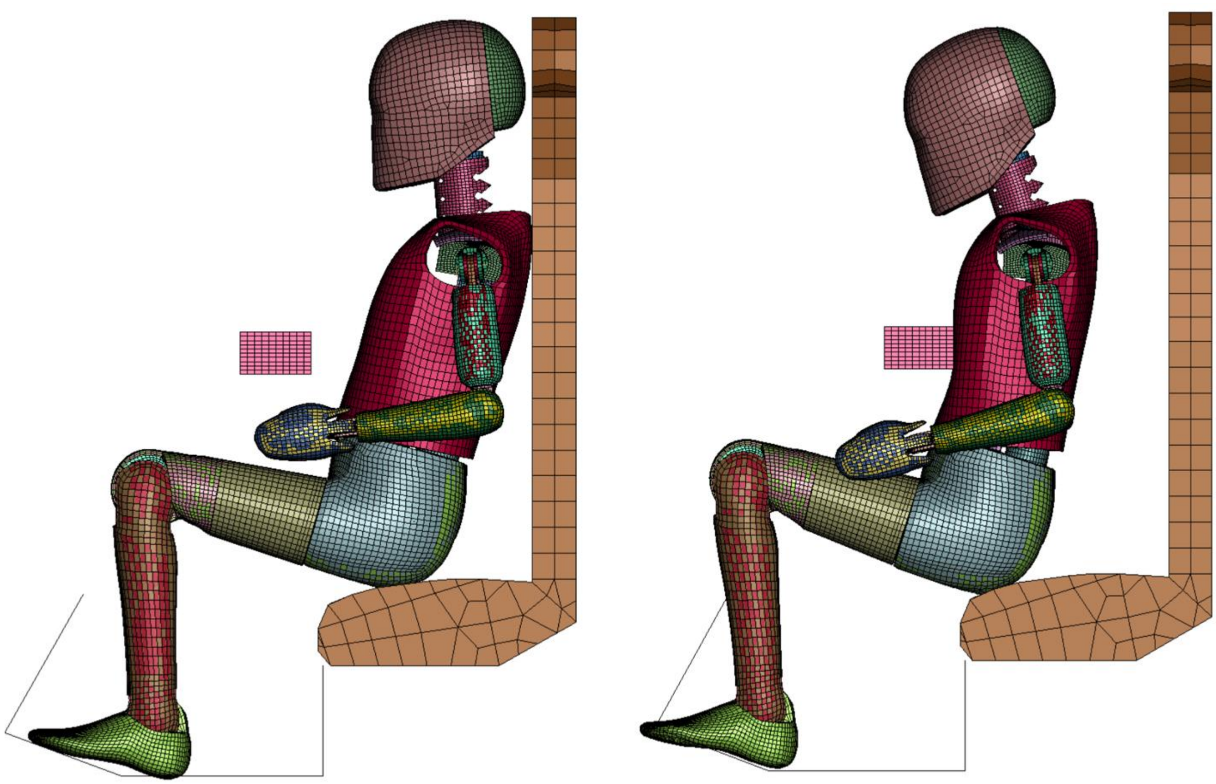


Figure 10: Unbelted model containing a joystick before impact (left) and at impact (right).

Results indicate that occupant displacement can be reduced using a lap belt, and further reductions in displacement are achieved with a 5-point harness. However, a 5-point harness with minimized belt attachment distance (as recommended) reduced displacements the greatest. As the greatest concerns for these ride-on toys are related to displacement (e.g., falls or impact with the interior of the vehicle), findings from this study support the use of these types of restraint systems. The placement of a joystick into a high-speed unbelted model reiterates this need as the child contacted the joystick and injury metrics were negatively affected.

5.6. Conclusion

With the addition of either restraint condition, injury metrics tended to increase by modest amounts. However, this trend was not uniformly present – several injury metrics decreased modestly and some increased markedly with the addition of restraints. An increase in distal forces is expected with greater proximal restraint. Injury metrics increased substantially only in high-speed impact conditions, where PLA from the lap belt test and HIC15 from the 5-point harness test increased significantly relative to the unbelted condition. However, it should be noted that in situations where an unbelted occupant makes contact with interior components, injury metrics can exceed those measured in any belted condition. Regardless, none of the collected injury metrics ever approach known tolerance thresholds[62], and most are well within the range that is experienced by a child in daily life [61]. Taken together, these results support the use of a 5-point harness system to minimize displacement related injuries with little-to-no added risk in inertial injury metrics. Finally, it should be emphasized that this study provides a completely novel framework to simulate ride-on toy collision events, which can be used in future studies to evaluate the safety of the toys as they are applied in a rehabilitation

setting, such as with children with disabilities. Avenues of future research could involve topics such as developing a restraint system for ride-on toys that exhibits more uniform behavior across impact speeds or optimizing interior component arrangements for individual children.

6. Study 2

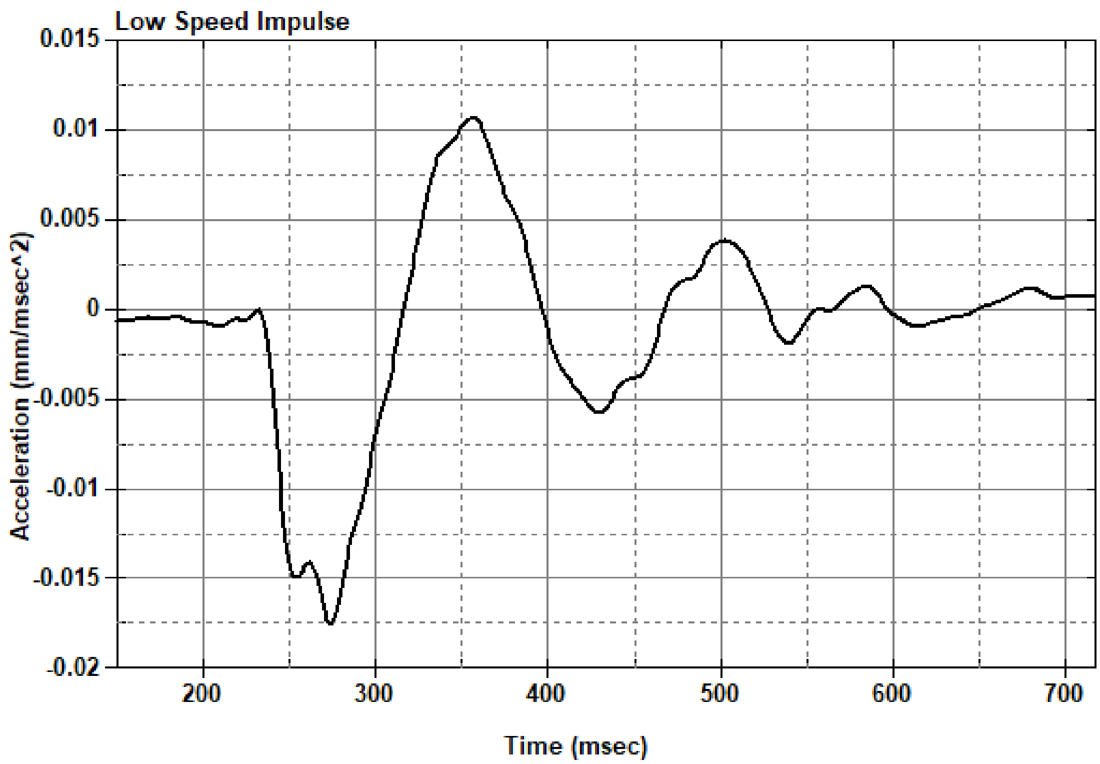
This study consisted of simulating a frontal collision using a modified version of a powered ride-on toy car containing parametrically altered versions of a Hybrid III finite element (FE) model. The 50th percentile 6YO model was used because many children do not receive ride-on toys until they are four to six years old and was therefore chosen to best represent the population of focus for this study.

Three restraint system configurations were analyzed: 1) no seatbelt, 2) a lap belt, and 3) a 5-point harness. The seat geometry used represents a ride-on toy with increased seat back height. The joint stiffness of the 6YO model was parametrically altered to be increased and decreased by both 90% and 50%. The simulation tool chosen was the computer program LS-DYNA because it is capable of highly non-linear, transient dynamic finite element analysis using explicit time integration.

6.1. Impulse

To supply an acceleration impulse to the FE model, frontal collision acceleration data was acquired via crash testing during the first study (see section 2.1 Impulse). Three impact speeds were tested, and used in the first study, as there were three speeds available on the remote control of the vehicle. However, when adapting the ride-on toys for children with disabilities the speed of the vehicles is reduced as a safety precaution. Because the scope of this study only extends to the effects of safety modifications on children with disabilities, the highest speed

was excluded from the study. Therefore, the two remaining speeds will be referred to as low- and high-speed. For each speed, the worst-case scenario (greatest peak acceleration) was using for the FE simulations. The average speeds of the vehicle were found to be 2.2 km/h and 3.6 km/h and are shown in Figure 10.



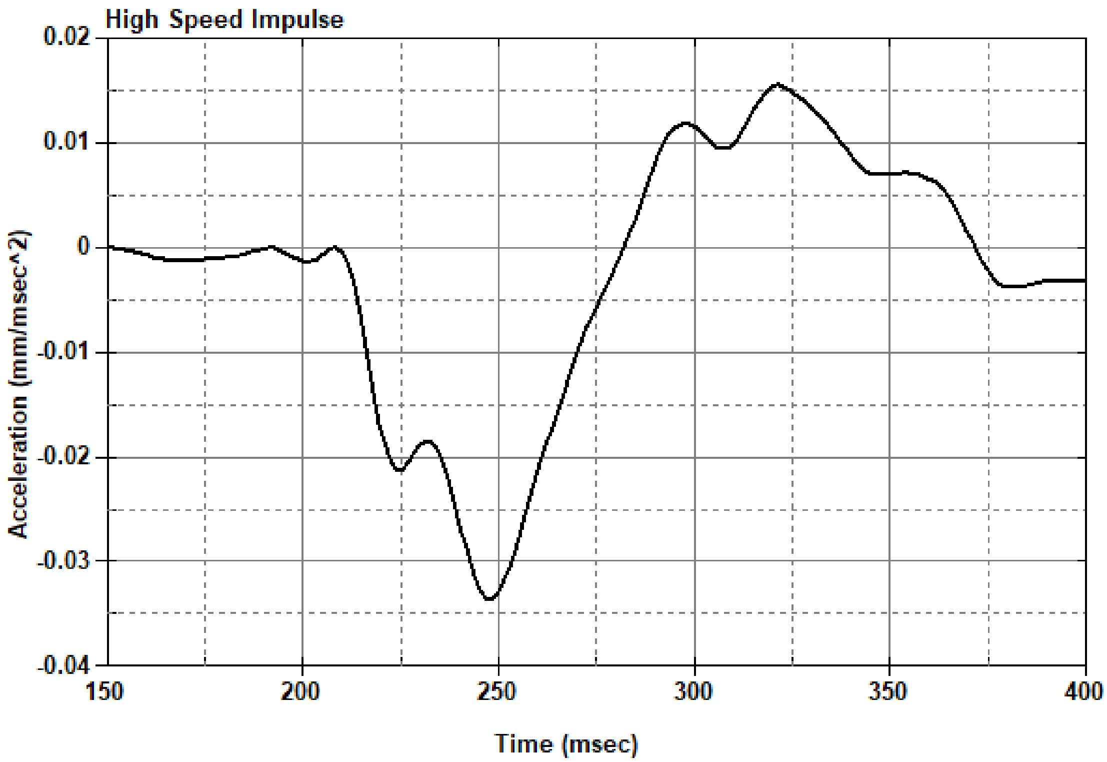


Figure 11: Experimental acceleration impulses for low and high speeds used in safety analysis of modifications made to a ride-on toy using the modified Hybrid III six-year-old ATD model.

6.2. Occupant modeling

The FE models replicated the ride-on toy car interior and were created using shell elements with rigid material properties. The vehicle model, or sled, composed of the seat and dashboard, had a back height 320 mm greater than an unmodified car. This increase was determined by the height differential between the seat and child's head and replicated the ride-on toy program's back and neck support modification intended to prevent neck hyperextension.

Three harness conditions were considered in this study: 1) unbelted, 2) lap belt, and 3) 5-point harness. In the previous study, two variations of the 5-point harness were considered; the difference being in attachment to the vehicle (closely attached above the shoulder, or higher above the shoulders). This was due to the variability in adaptations of the vehicles because

they are customized to each child. Both versions were found to be well within the threshold of known injury tolerances, but a shorter attachment point is regarded as best practice to prevent injury and was found to best decrease displacement of the occupant[17], [34]. Therefore, only the recommended 5-point harness low-attachment was considered in this study.

The sled was constrained to only allow forward motion in the y-direction. The 6YO FE model was place in a seated position in the vehicle model and automatic surface-to-surface contact defined interaction between the seat and axial skeleton, and between the feet and floorboard. Gravity acted on the system for 150 milliseconds before the acceleration impulse was applied to allow the human model to settle into the seat. The sled was constrained to a massless node which served as a global reference point and two load curves were defined, one for low speed and one for high. A prescribed motion was then applied using the massless node and load curve.

The restraint harnesses consisted of a mixed belt (containing both 1D seatbelt elements and 2D triangular, shell elements). The 1D belt material was defined using the seatbelt material keyword card in LS-DYNA and used loading and unloading curves. The 2D shell component material of the seatbelt was defined using the fabric material keyword card in LS-DYNA. The material values were found experimentally in a study conducted by LS-DYNA, who's purpose was for modeling seatbelts in the program[32]. The interaction between the dummy and seatbelt was defined using automatic surface-to-surface contact. The final models are shown below.

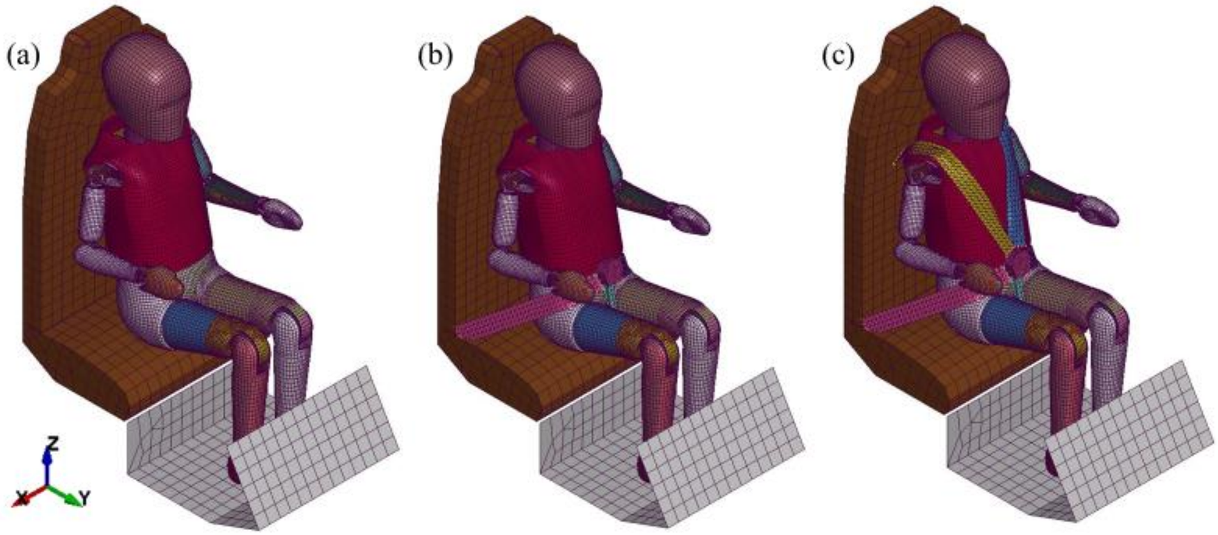


Figure 12: 6-year-old Hybrid III model in modified ride-on toy car model with: no seatbelt (a), lap belt (b), and 5-point harness (c).

There are 19 defined joints in the FE Hybrid III child model. These joints account for the head and neck, three in each leg, and six in each arm. To model a child with non-average joint stiffness, the stiffness of each was parametrically altered. The stiffness of each joint increased and decreased by 90 and 50 percent. Including the original model with unmodified joint stiffness, this totaled five versions of the 6YO dummy model (10%, 50%, 100%, 150%, 190% of original stiffness).

6.3. Post-processing

The following injury metrics were collected or calculated for each simulation: head injury criterion (HIC), peak angular acceleration (PAA), peak linear acceleration (PLA), head and chest displacement, neck tension force, neck injury criterion (Nij), neck bending moment, and neck transverse shear. For these, a higher risk of injury is represented by higher value results. The likelihood of head injury occurring from an impact is measured by HIC[59]. HIC scores can be interpreted using Figure 12 and depicts the risk curve of sustaining a level 1 injury on

the abbreviated injury scale (AIS1). An AIS1 injury is defined as the most minor injury one can sustain with a zero percent probability of death[63]. Neck injuries in low speed rear-end collisions can be predicted using neck injury criteria and risk of injury increases as the value approaches one[64]. The peak linear accelerations experience in daily activities of living are typically on the order of ten g's[61] Displacement depicts a risk of impacting the interior of the vehicle or falling from the vehicle, events that are both considered among the greatest risks of injury. As displacement increases, so does the risk of injury.

6.4. Results

All kinetic and kinematic outcomes are shown in Table 3.

Seatbelt Config.	Speed	Joint Stiff.	HIC15	PAA	PLA (g's)	Head Displacement	Chest Displacement	Neck Force Tension (kN)	Nij	Neck bending moment	Neck Transverse Shear
No Seatbelt	Low	No change	0.05	0.035	2.82	164.4	131.2	0.061	0.023	0.280	0.128
		-90%	0.03	0.025	2.71	179.0	139.1	0.043	0.015	0.166	0.117
		-50%	0.06	0.032	3.26	167.4	135.0	0.058	0.021	0.178	0.128
		50%	0.05	0.033	4.63	164.9	130.7	0.055	0.020	0.273	0.173
		90%	0.05	0.036	4.36	163.7	129.9	0.075	0.026	0.187	0.172
	High	No change	0.18	0.043	8.98	207.9	148.2	0.078	0.031	0.354	0.305
		-90%	0.22	0.057	9.47	222.3	150.5	0.078	0.035	0.770	0.324
		-50%	0.21	0.045	9.05	210.2	147.8	0.105	0.040	0.496	0.311
		50%	0.29	0.045	10.14	208.0	147.0	0.142	0.055	0.532	0.355
		90%	0.43	0.049	11.42	207.7	146.8	0.085	0.031	0.245	0.409
Lap Belt	Low	No change	0.24	0.042	9.57	169.0	127.1	0.099	0.039	0.402	0.335
		-90%	0.19	0.032	8.39	173.6	132.2	0.052	0.019	0.149	0.292
		-50%	0.22	0.047	9.63	170.1	128.8	0.090	0.036	0.401	0.334
		50%	0.33	0.051	11.62	167.3	125.8	0.089	0.032	0.375	0.393
		90%	0.29	0.042	10.77	166.9	125.8	0.087	0.033	0.380	0.367
	High	No change	0.20	0.054	5.21	210.4	135.5	0.111	0.043	0.470	0.188
		-90%	0.23	0.045	10.14	217.0	135.6	0.098	0.038	0.444	0.339
		-50%	0.18	0.060	4.10	211.3	135.4	0.113	0.043	0.361	0.132
		50%	0.21	0.055	6.77	207.6	134.4	0.135	0.048	0.210	0.236
		90%	0.19	0.050	5.94	207.6	134.9	0.101	0.038	0.328	0.219
5-Point Harness	Low	No change	0.12	0.051	7.88	136.6	108.3	0.074	0.028	0.322	0.282
		-90%	0.13	0.048	3.57	146.4	110.8	0.058	0.019	0.212	0.136
		-50%	0.16	0.037	6.74	145.1	110.0	0.063	0.024	0.152	0.246
		50%	0.16	0.038	5.00	143.6	109.0	0.059	0.023	0.366	0.197
		90%	0.14	0.038	4.41	143.7	109.5	0.059	0.024	0.449	0.169
	High	No change	0.17	0.063	4.90	157.3	114.3	0.108	0.038	0.434	0.125
		-90%	0.31	0.056	5.85	164.2	122.0	0.166	0.061	0.541	0.123
		-50%	0.20	0.053	6.46	168.2	121.3	0.114	0.040	0.284	0.170
		50%	0.28	0.061	5.51	167.3	120.4	0.160	0.052	0.689	0.181
		90%	0.23	0.070	5.24	168.1	120.3	0.157	0.047	0.886	0.176

Table 3: All kinetic and kinematic results from computer modelled frontal crash analysis

Injury Class 1: Excursion Related Injuries

Overall, displacements decreased with the addition of seatbelts. Chest displacement decreased with the addition of a lap belt and further lowered with the use of a 5-point harness. Head displacement increases marginally with the addition of a lap belt but decreased in every instance with the use of the 5-point harness.

When comparing all displacement values among respective models only considering joint stiffness change, there is no change greater than 9% from the original model. When joint stiffness decreases, a minor increase in displacement is seen in every instance and the largest displacement occurs at the lowest stiffness of 10% (with no seatbelt, at high speed). This model's displacement improves appreciably with addition of a 5-point harness. Therefore, despite change in joint stiffness, there is still a substantial decrease in displacement when comparing the 5-point belted model to unbelted conditions.

Injury Class 2: Injuries Related to Kinetic Variables

With the addition of a seatbelt, HIC15 scores generally increased. For models with unmodified joint stiffness, HIC15 increased most with the addition of a lap belt. However, with variant joint stiffness, the most severe HIC score (0.43) occurred at the highest stiffness value, 190%, with no belt, at high speed. In every instance, low-speed impulses resulted in lower HIC scores.

Nearly all PLA values were considerably below 10 g's. The greatest PLA experienced was 11.62 g's at low speed, 150% joint stiffness, and the lap belt. The only other instance on this magnitude was a PLA of 11.42 at high speed, 190% joint stiffness with no belt. For the models with no seatbelt and 5-point harness, low-speed impulses resulted in overall lower PLAs. In fact, PLA decreased with any change in joint stiffness at low speed and only increased marginally at high speed. The greatest PAA occurred at high speed, 190% joint stiffness, with a 5-point harness and was only 0.07 rad/mms^2. PAA increased or decreased minimally for each instance of no belt or with a lap belt. PAA decreased with every change in stiffness with a 5-point belt except at high speed with the greatest stiffness, 190%.

Generally, neck injury metrics (Nij and neck force tension, bending moment, and transverse shear) decreased marginally at low speed with change in joint stiffness and increased minimally at high speeds. The greatest change in each neck injury metric occurred under the greatest change in joint stiffness. Specifically, the largest neck force (0.166kN) and Nij (0.061) was seen at a 10% joint stiffness with a 5-point harness under high speed. The greatest neck bending moment (0.886kN-mm) and transverse shear (0.409GPa) both occurred at 190% joint stiffness under high speed, bending moment with the 5-point harness and transverse shear with no belt.

It is notable that with the use of a 5-point harness, neck injury metrics actually decreased with any change in stiffness at low speeds. The sole exception being that bending moment increased modestly at higher stiffness.

6.5. Discussion

Overall, injury metrics remained low for all test conditions when compared to common injury thresholds. Two classes of injury mechanisms were identified for review, excursion related injuries and injuries relating to kinetic variables. Excursion related injuries were identified in the literature as the primary cause of injuries and injuries relating to kinetic variables are used with common injury thresholds to gauge risk of injury.

Injury Class 1: Excursion Related Injuries

As seen in Study 1, displacement decreased with the addition of a lap belt, and more so with the addition of a 5-point harness system[65]. The lap belt sufficiently reduced chest displacement, however, was ineffective at preventing forward movement of the head. The 5-point harness successfully reduced both chest and head displacement. When considering displacement comparisons among the models with altered joint stiffness to that of the original

model (50th percentile 6YO joint stiffness), the largest displacement occurs with no seatbelt at the lowest joint stiffness, confirming the need for a 5-point harness for passengers with variant joint stiffness. Thus, the 5-point harness can prevent injurious excursion from the vehicle or contact with interior components and provides a mechanism for affective deceleration of the body, regardless of irregular joint stiffness.

Injury Class 2: Injuries Related to Kinetic Variables

Inertial loading caused by restricted movement of the torso can alter injury metric magnitudes and seatbelt addition has been observed to cause increased head accelerations in the context of motor vehicle collisions. The same minor increase in accelerations with the addition of seatbelts were observed here. While acceleration associated injury metrics increased with change in joint stiffness, all values were small in comparison to known injury thresholds. For example, studies show the peak linear accelerations experienced in activities of daily living are on the order of ten g's (e.g. "plopping" passively into a chair generates 10.1g's, falling down results in 16.3 g's, and being struck by a pillow generates 28.1g's)[61], [66] Furthermore, studies show the accelerations experienced by a car going over a speed bump ranges from 3-8 g's[67]. Although HIC15 scores generally increased, all recorded values were extremely small compared to AIS injury thresholds (Figure 12) [59] The maximum HIC score recorded, 0.43, corresponds to a probability of an AIS1 injury of 1.125 x 10^{-142}, or zero.

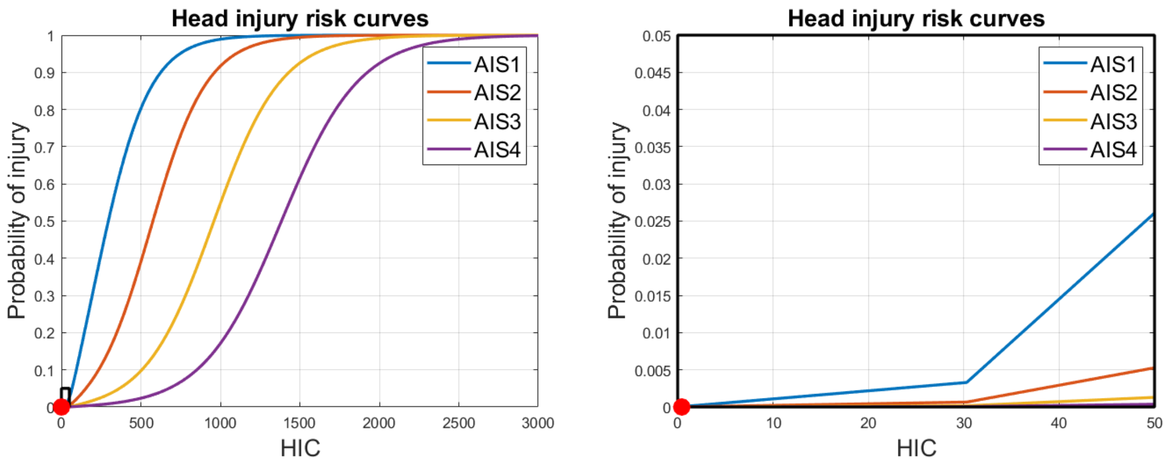


Figure 13: Head injury probability curves based on HIC values. The figure on the right is an inset of the figure on the left. The red dot indicates the maximum HIC value recorded in this study (0.43) with a corresponding probability of injury of 1.125 x 10^{-142}.

Within this context, the discussion of changes to injury metrics is discussed under the knowledge that no injury thresholds were exceeded. Generally, injury metrics decreased at low speeds with change in joint stiffness and increased at high speeds. The increase of PLA with use of a 5-point harness is likely because the high speed geometrically puts the head in an easier flex posture. The same can be said for the peak PLA value seen with the lap belt, because the lap belt causes the head to hinge forward. This bump in PLA is compounded in the HIC calculation.

A limitation to this study was the vehicle and model used. The 12 V-powered Mercecdes AMG G63 ride-on battery powered car used in this study only represents one possibility of vehicles used in ride-on toy programs. The adaptive ride-on toy programs use vehicles varying in size, weight and battery power and extend to occupants of varying weight and size and therefore offers a wide range of opportunity for future study. Furthermore, as modified joint stiffness is only a first-order approximation of modeling a child with disabilities, future work

would include taking further steps to model this population. Future test dummy model modifications would include varying geometry, modifying individual joint stiffness and more.

6.6. Conclusion

Precaution should be taken for children with musculoskeletal disorders, or severe joint stiffness, given peak kinetic values occur at joint stiffness extremes and high speeds. Furthermore, if there is a concern for those who suffer from considerable musculoskeletal deficiencies, low speed is recommended.

Displacement decreases with the addition of a 5-point harness regardless of change in joint stiffness. As seen in the previous study, addition of a seatbelt caused kinetic variables to rise; however, no known injury thresholds were approached. Therefore, using modified joint stiffness as a first-order approximation of a child with disabilities, results from this study indicate that it is acceptable to use a seat belt for this population. Furthermore, as excursion-related injuries are considered more critical to the user than injuries relating to kinetic variables and no known injury thresholds were exceeded, the addition of a belt is considered a necessary trade-off with little-to-no added risk.

7. Final Conclusion and Discussions

Considering the success of adaptive ride-on toy programs, it is necessary to ensure modifications act as intended without introducing additional risk of injury. The first study examined the safety of common adaptations made to battery-powered ride-on toys using the Hybrid III six-year-old dummy model. The results supported the use of seat modifications and a 5-point harness system as it successfully minimized displacement related injuries with little-to-no in inertial injury metrics. However, this study only considered the average six-year-old

model and these ride-on toys are used as rehabilitative tools for children with disabilities. Within this context, the second study examined the safety of these adaptations using the Hybrid III six-year-old dummy model with modified joint stiffness as a first-order approximation of a child with disabilities. This study discovered displacement decreased with the addition of a 5-point harness, regardless of changes in joint stiffness. Furthermore, because additional precaution should be taken for children with musculoskeletal disorders and peak kinetic values occur at large changes in joint stiffness under high speeds, low speed is recommended.

These studies are significant due to lack of research in the field of safety of pediatric rehabilitative devices, specifically adaptive ride-on toys. The proven success of these rehabilitative programs further shows these studies are a valuable tool intended to better equip pediatric care providers with knowledge on the safety of car modifications. Furthermore, the findings of these studies support the growth of adaptive ride-on toy programs to increase rehabilitation opportunities for children with disabilities. As the reach of these programs extends to a broader population, pediatric care providers can better implement safety modifications and know their efficacy when operated under safe conditions (e.g. low speeds and with restraint devices). This safety analysis of modified ride-on toy cars is intended to be a useful tool in ensuring safety of children with disabilities, and further research building on this work would increase the benefits and safety of adaptive ride-on toy programs.

8. Future Work

It is recommended that future students develop data sets that are more comprehensive and more broadly applicable to the range of adaptive vehicles that are used in adaptive toy

programs. This would mean using cars of varying size and weight to collect crash data to apply to the computer model and applying these new geometries to the computer model. This is recommended to verify that the result of impact doesn't vary with car size or weight.

Furthermore, it is suggested to consider a wide range of collision orientations. This additional method of checking the quality of safety modifications would provide greater detail for preventing potential injuries.

Another area of future study would be with other test dummies. The six-year-old hybrid III model was used in this study; however, a three-year-old (3YO) and ten-year-old (10YO) hybrid III model exists in the pediatric Hybrid III family. The use of the 3YO dummy would not only be a step in modeling a younger child but would also allow a child with disabilities with delayed growth and decreased weight to be studied. Furthermore, there are many other families of pediatric test dummies available. The more recently developed Q-series family of pediatric ATDs may yield different responses than the Hybrid III pediatric models and has six models ranging from a six-week-old newborn to a ten-year-old child[68]. Analysis using multiple families of ATDs would provide further verification of the safety of ride-on toys and the Q-series would be a fantastic continuation to this work.

A continuation of determining risk of injury is ultimately determining safe exposure levels for children with disabilities. However, there are limitations with drawing conclusions on risk of injury because there is a gap in the literature concerning injury thresholds of children with disabilities. Because volunteer data is unethical/difficult to obtain and cadaver data doesn’t exist, further testing could be done examining children with disabilities as they live their everyday life. Data is collected while the children experience small inertial events throughout the day (e.g. going over a speedbump) to establish a threshold of what is tolerable for these

children and give insight into interpreting the results of other various inertial events. This method is one of the most common biomechanical methods for determining “tolerable” levels of exposure when other options are not available.

Finally, a necessary area of continuation would be efforts to model a child with disabilities. There is no proposed way to simulate a child with disabilities and changing joint stiffness values only accounts for one piece of the puzzle for a child with impairments. (e.g. body geometry and position were not accounted for.) Simulating children with neuromusculoskeletal impairments are difficult. Simulating a child that represents the majority of children with disabilities is even harder. Therefore, the proposed method of changing the joint stiffness of the dummy to represent children with disabilities is only the first step in beginning to model this population. This is perhaps the greatest area of future work and includes geometry modifications, body positioning, non-uniform joint stiffness changes and more. The benefit of modeling a child with disabilities is broad reaching and could be applied to a myriad of study areas.

Ideally, as research continues in an effort to improve safety and functionality for battery-powered, adaptive, ride-on toys, computer analysis could be utilized to create an optimized ride-on toy. Seatbelt positioning and geometry could be analyzed to determine the ideal implementation of a seatbelt. Furthermore, varied seatbelt material properties could be analyzed to determine what produced the best displacement reduction and best kinetic injury metrics. Similar variations could be analyzed with seat/car geometry and material properties. Once optimum geometries and materials have been determined, this optimized model could be used as the framework for all adapted ride-on toy projects. Furthermore, for special cases (e.g.

exceedingly high-risk children), a ride-on toy model could even be optimized for a specific child's needs.

9. Ethical Considerations

Due to the limitations mentioned and various others, conducting safety analysis on a specific population such as children with disabilities is difficult, as there is no standardized methodology. Despite efforts to adapt the toy cars to increase safety, there is always a risk of injury with any ride-on toy and especially for children with special needs. Therefore, caution must always be used when these children are using the adaptive ride-on toy cars.

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