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BIOMECHANICS AND INJURY ASSESSMENT OF HOUSEHOLD FALLS IN
CHILDREN: CLINICAL, ANTHROPOMORPHIC SURROGATE, AND COMPUTER
SIMULATION STUDIES

By

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B.S., University of Louisville, 2005
M.Eng., University of Louisville, 2007

A Dissertation
Submitted to the Faculty of the
J.B. Speed School of Engineering of the University of Louisville
in Partial Fulfillment of the Requirements
for the Degree of

Doctor of Philosophy

Department of Mechanical Engineering
University of Louisville
Louisville, Kentucky

May 2011

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ACKNOWLEDGMENTS

This research was funded by the University of Louisville Grosscurth Biomechanics Endowment Fund and by the Department of Justice (DOJ), Office of Juvenile Justice and Delinquency Prevention (Award # 2009-DD-BX-0086). The opinions expressed herein are those of the authors and not necessarily those of DOJ.

I would first like to extend my deepest gratitude to my advisor and mentor, Dr. Gina Bertocci. Without her support and guidance, none of this work would have been possible. She pushed me to find solutions to seemingly impossible problems and instilled in me the drive and persistence necessary for success. Next, I would like to thank Dr. Mary Clyde Pierce. Witnessing her passion for this field of research has been an inspiration to me. Her knowledge and clinical perspective were invaluable to the success of this work. I thank the rest of my dissertation committee, Dr. Michael Voor, Dr. Peter Quesada, and Dr. Naira Campbell-Kyureghan, for their time, guidance, and input.

I owe much gratitude to all my peers in the iRAP lab. Zdravko, Raymond, Sheryll, Nathan, Craig, you have all supported me with my experiments, modeling, and general assistance when I was stumped with a problem. The iRAP lab has been the ideal place to grow and learn, and I owe that to all of you.

I need to thank my husband, Peter, for his never-ending words of encouragement. I thank my parents and family for their love and support, and all my friends who have been so supportive over the years.

ABSTRACT

BIOMECHANICS AND INJURY ASSESSMENT OF HOUSEHOLD FALLS IN CHILDREN: CLINICAL, ANTHROPOMORPHIC SURROGATE, AND COMPUTER SIMULATION STUDIES

Angela K. Thompson

May 14, 2011

Pediatric short-distance falls, especially from beds or other furniture, are common false histories given by caretakers to cover up abusive trauma. However, short-distance falls are also a common occurrence in young children. Knowledge of the types and severity of injuries that can result from these short falls can aid clinicians in distinguishing between inflicted and non-inflicted injuries. Early detection of abuse may lead to prevention of further escalating injuries and, in some cases, prevent the death of the child.

The purpose of this study was to describe relationships between biomechanical measures and injury potential in short-distance household falls. This study involved three components: case-based biomechanical fall assessments, fall simulations using an anthropomorphic test device (ATD), and development/validation of a computer simulation model used to investigate sensitivity of injury outcome measures to fall environment and child surrogate parameters.

Overall, the risk of severe or life-threatening injury in short-distance household falls is low. Fractures of the skull and extremities commonly result from these falls

(21.5% of falls resulting in Emergency Department visits). 2 of 79 fall cases involved small, contact-type subdural hematomas. These subjects both had unique fall dynamics that contributed to their injuries. Results of ATD experiments supported those from the clinical portion of the study with the exception of neck injury potential. Future studies are needed to both improve ATD neck biofidelity and determine more accurate pediatric neck injury thresholds.

Fall environment parameters (fall height and impact surface type) have been shown previously to influence injury potential, but this is the first study to investigate the influence of child or surrogate parameters (body mass index, overall mass, head stiffness, and neck properties) on injury potential. Additionally, through a parametric sensitivity analysis, it was found that fall environment and surrogate parameters that altered fall dynamics had the greatest influence on injury potential. These results highlight the need for obtaining detailed case histories when making injury assessments that include not only environment and child factors, but descriptions of the fall dynamics and orientation of the child upon impact with the ground.

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CHAPTER I
INTRODUCTION

Specific Aims

Child abuse is the leading cause of trauma-related fatalities in children less than four years of age.¹ Children aged one year or less are particularly at risk with approximately one out of every 41 children in this age group suffering from abuse.² In the United States alone in 2006, there were approximately 905,000 victims of child abuse. There were also approximately 1,530 fatalities due to child abuse with 78% of these cases involving children aged four years or less.² These numbers may be underestimated since it has been suggested that as many as 50-60% of deaths related to child abuse go unrecorded.³

Short-distance falls in children, especially from beds or other furniture, are a common false history given by caretakers to cover up abusive trauma. In up to 70% of cases of children with abusive injuries, the initial explanation for the injuries given by the caretakers is a fall.⁴⁻⁸ However, short household falls are also a common occurrence in young children. A study of emergency department visits by children less than one year of age found that 61% of accidental cases were injuries due to falls.⁹ Knowledge of the types and severity of injuries that can result from these short falls is necessary since clinicians are commonly asked to determine whether a child's injuries are consistent with

the stated cause of the injuries, when attempting to distinguish between inflicted and non-inflicted injuries. Early detection of abuse may lead to prevention of further escalating injuries and, in some cases, prevent the death of the child. Additionally, there is continuing controversy in the medico-legal community over whether short distance falls can lead to severe injuries or death.

The purpose of this study was to provide objective information about injury risk in short-distance falls to aid clinicians in distinguishing between inflicted and non-inflicted injuries in children. This was accomplished by investigating the injury outcomes and biomechanics associated with common household falls. Four specific aims were established to achieve this goal:

- 1. Determine injury types and severities that are associated with short falls from horizontal furniture surfaces (i.e. fall from bed, crib, couch, table, etc) in children ages 0-4.*
- 2. Describe fall dynamics and determine biomechanical measures associated with pediatric falls from horizontal furniture surfaces.*
- 3. Describe relationships between biomechanical measures and injury severity outcomes in pediatric falls from horizontal furniture surfaces.*
- 4. Determine whether fall environment factors (height of fall, impact surface), initial velocity and surrogate characteristics (mass, head properties, neck properties, soft tissue properties) influence fall dynamics and injury potential in falls from horizontal furniture surfaces.*

The overall hypothesis was that short-distance falls involving young children have a low potential for severe injuries, and that injury potential is influenced by both fall environment and surrogate (fall victim) characteristics.

This study involved three major methodological components to address the specific aims and obtain a better understanding of injury risk in short-distance household falls. The first component was a case-based biomechanical assessment of children who present to the emergency department of a metropolitan children's hospital with a history of a fall from a bed or other similar furniture (Chapter 2). Descriptions of fall dynamics and fall environment characteristics were obtained through interviews with the caregivers and in-depth scene investigations. Relationships between biomechanical measures and injury severity outcomes were determined.

The second component utilized an anthropomorphic test device (ATD), or human surrogate, representing a 12-month-old child to experimentally simulate falls from furniture surfaces in a laboratory setting (Chapter 3). The ATD was instrumented to obtain measures related to head, neck, and extremity injury potential.

The final component involved development of a validated computer simulation model based upon the ATD experiments (Chapter 4). Once validated, the computer model extended beyond the ATD experiments by allowing variation in fall environment and ATD parameters as part of a parametric sensitivity analysis (Chapter 5). Relationships between parameters and measures related to injury potential were described.

Background and Significance

Characteristics of Abusive versus Accidental Injuries

Head Injuries

Much work has been done to identify and distinguish injury characteristics associated with child abuse with those from accidental causes. Perhaps among the earliest of these studies, was that of Caffey,¹⁰ which defined the characteristics of “whiplash shaken infant syndrome” (also commonly called “shaken baby syndrome”) to be severe head injuries, specifically subdural hematomas (SDH), and retinal hemorrhages (RH) without any external signs of trauma. Although “shaken baby syndrome” is not the only abusive mechanism, the characteristic injuries remain the same.

Subdural hematomas are known to result from large rotational accelerations of the head. This causes the brain to move relative to the skull, rupturing the bridging veins.¹¹ In a study by Geddes et al.¹², SDH was found to be the most common injury among patients with abusive head injuries, present in 81% of cases. Bechtel et al.¹³ found a similar result with SDH in 80% of patients with abusive head trauma and only 27% of patients with accidental head trauma. SDHs have been reported in high-energy events such as motor vehicle accidents and falls from great heights. Duhaime et al.⁴ found three accidental cases of SDH, all occurring in motor vehicle accidents. Billmire and Myers¹⁴ found one case of SDH among 19 to be the result of a motor vehicle accident. Barlow et al.¹⁵ reported 1 SDH in a fall from greater than three stories. Musemeche et al.¹⁶ reported two SDHs in 70 falls from heights of ten feet or greater.

It has been estimated that between 65 and 95% of “shaken baby” cases involve retinal hemorrhage.¹⁷ RHs are likely due to a rise in intracranial pressure secondary to traumatic brain injury.¹⁸ RHs have been recorded in accidental cases, but these are much rarer and often differ by type and location from those seen in abusive cases. In a study by Bechtel et al.¹³, 60% of patients classified as having abusive head trauma were found to have RH versus only 10% in the accidental cases. Multiple and bilateral RHs were more likely to occur in abuse cases. Abusive RH also more often involved the pre-retinal layer and extended to the periphery of the retina. Another study found RHs in 10 of 100 children sustaining head injuries.⁴ Nine of the ten cases were classified as abusive, with the single accidental RH being the result of a high-speed motor vehicle accident. All 10 patients also had SDH. Geddes et al.¹² found 71% of 38 children with non-accidental head injury to have RHs. The authors also found a significant association between the presence of RHs and SDH.

Another brain injury commonly associated with abuse is diffuse axonal injury (DAI). DAI results from shear forces on the axons of neurons in the brain and can range from mild concussion to severe comas resulting in death. A recent study suggests that severe DAI is actually a rare result of abusive trauma. In a study of 37 infants with inflicted head injuries, only two were found to have severe DAI.¹⁹ Concussion on the other hand, has been reported commonly in both abusive and accidental cases. One study documents 20 cases of concussion in head-injured infants, with 13 due to accidental causes, and two of those were from falls out of bed.¹⁴ The remaining cases were due to motor vehicle accidents or falls from a caretaker’s arms onto a hard surface.

Skull fractures have been shown to occur in both abusive and accidental trauma. Billmire and Myers¹⁴ reported 78% of skull fractures occurring from accidental causes. However, 87% of the skull fractures were linear parietal fractures. Only four infants had complex, multiple fractures. All of these had associated intracranial hemorrhage and all were due to inflicted trauma. Another study reported 91% of skull fractures occurring from accidental trauma.¹³ Duhaime et al.¹⁷ reported that autopsies detect fractures in 25% of “shaken” infants. These fractures are most commonly in the posterior parietal bone or the occipital bone. Skull fractures have been documented frequently in falls. In a study of 66 free falls in children, there were 10 skull fractures, of which eight occurred from heights greater than two stories and two occurred from heights less than one story.²⁰ Lallier et al.²¹ also found 10 cases of skull fractures among 64 children who sustained falls greater than 10 feet. Among short-distance falls, 3 of 246 children who fell from a bed or sofa had skull fractures.²² Two of the three children were 6 months of age or less. Age was not specified for the third child. Another study of bed falls reported one skull fracture in 207 falls.²³ Five skull fractures were reported in a study of 69 stairway falls.²⁴

Neck Injuries

Cervical spine and spinal cord injuries are rarely reported in cases of child abuse. However, they are of interest because the mechanisms of the “shaken baby syndrome” would seem likely to cause whiplash injuries to the neck. One study reported that in order to reach acceleration levels necessary to cause the severe head injuries described in shaken baby syndrome, the thresholds for neck injury would be exceeded.²⁵ This publication was criticized however, and it was determined after attempts to repeat the

calculations, that neck forces were actually far below the threshold for injury.²⁶ Few studies have reported cases of neck injuries after inflicted trauma. In a study by Hadley et al.²⁷, 5 of 6 abuse patients who had retinal and intracranial hemorrhages were also found to have injuries at the cervicomedullary junction after autopsy. These included subdural and epidural hematomas on the spinal cord and cervical spinal cord contusions. Ghatan and Ellenbogen²⁸ reported a case an infant who sustained a vertebral atlantoaxial dislocation and rupture of the transverse ligament of the atlas. Another study reported cases of lower cervical spine injury in two infants as the result of abusive trauma.²⁹ One had a fracture of the C5 vertebral body and a resulting dislocation of C4 and spinal cord compression. The other infant had a fracture-dislocation of C5 onto C6. Although neck injuries are common in motor vehicle accidents, they are rarely reported in falls. Chiaviello et al.²⁴ reported that 1 of 69 children who fell down stairs sustained a C2 fracture. Other studies report spine fractures in falls from heights of 10 feet or more but do not specify whether these are cervical spine injuries.^{15-16, 21}

Fractures

Fractures are commonly seen in child abuse, appearing in 25 - 55% of child abuse cases.^{7, 30-34} In young non-ambulating children, fractures are strong indicators of abuse. Studies report that 26-56% of all fractures in children less than 1 year of age are inflicted.^{5, 35-38} Younger children are at an even greater risk. Leventhal et al.⁵ reported 24% of fractures in children less than 3 years old were inflicted, but in children less than 1 year old, this number increased to 39%. Similarly, Skellern et al.³⁵ reported 26% of fractures in children less than 1 year old as inflicted, but 50% of fractures in children less

than 4 months old were inflicted. Even still, fractures are commonly seen as a result of accidental trauma, such as a fall or motor vehicle accident, and many studies have attempted to distinguish fracture characteristics such as type and location in children with inflicted and non-inflicted injuries.

Of particular interest are long bone injuries. Fractures of the extremities are among the most common injured sites in both abusive and accidental cases.^{5, 31-35, 37, 39-42} Although studies have varying results as to which long bone is most commonly fractured, most report no significant differences between children with inflicted and non-inflicted injuries regarding which bones are fractured.^{5, 31-32, 35, 37} Fractures of the femur and humerus account for approximately 20% of all fractures.^{5, 32, 35, 39, 43} Additionally, a study by Leventhal et al.⁵ found that 81% of humerus fractures and 35% of femur fractures in children less than 3 years of age were abusive in nature. Among children less than 1 year of age, Leventhal reported that 82% of all extremity fractures were inflicted.

Several studies have attempted to characterize and distinguish inflicted and non-inflicted long bone fractures based on the type and location of the fracture along the bone. There are several different types of long bone fractures that occur commonly in both abusive and accidental cases. These include spiral, buckle, transverse, oblique, and classic metaphyseal lesions (CML) or corner fractures. Each of these fracture types is associated with a different loading mechanism.⁴⁴ Spiral fractures result from torsional loads, buckle fractures result from compression loads, transverse fractures result from tension or bending loads, oblique fractures result from a combination of bending or tension and torsional loads, and finally CMLs may result from shear or tensile loading. Spiral, transverse, and oblique fractures are generally midshaft while buckle and CML

fractures are typically in either the proximal or distal metaphyseal region of the bone.⁴⁴ Given that different fracture types are associated with different loading mechanisms, it is possible that fractures resulting from abusive mechanisms would appear different from those with accidental mechanisms.

A few studies have found the type of humerus fracture to be significantly different in children with inflicted and non-inflicted injuries.^{5, 7, 39, 45} In 14 cases of children less than 3 years of age with humerus fractures, Thomas et al.³⁹ reported that all of the accidental fractures were supracondylar. Supracondylar fractures occur at the distal end of the bone in the metaphyseal region and typically result when children fall impacting their elbows. In contrast to the accidental group, no supracondylar fractures were seen in the abuse group. Instead the inflicted fractures were transverse and oblique fractures in the humeral midshaft or metaphyseal regions. Strait et al.⁷ reported 3 of 10 humerus fractures in abused children were supracondylar. The remaining 7 fractures were spiral or oblique. In the accidental group (children less than 15 months old), there were 8 supracondylar fractures and one each of the buckle, spiral, and transverse fracture types. Worlock et al.⁴⁵ reported that spiral fractures of the humerus were significantly more common in children with inflicted injuries than those with non-inflicted injuries, seen in 9 of 25 abused children but in none of 116 children with accidental fractures. Herndon⁴⁶ also reported spiral fractures to be the most common fracture type in abused children.

Unlike humerus fractures, most studies have found no significant difference in the type and location of femur fractures between inflicted and non-inflicted children.^{5, 39, 47-48} Rex and Kay⁴⁷ compared femur fractures in 14 children with inflicted injuries and 33 children with non-inflicted injuries and found the midshaft to be the most common

fracture site in both groups (57% in the abuse group and 48% in the accidental group) and spiral fracture to be the most common fracture type in both groups (57% in the abuse group and 42% in the accidental group). Anderson⁴⁹ also reported the midshaft to be the most common fracture location in occurring in 62% of 122 total femur fractures (includes both inflicted and non-inflicted fractures). However, he reported that transverse fractures, followed by oblique and spiral fractures are the most common fracture types in both inflicted and non-inflicted children. Scherl's results agreed closely with Anderson's. Scherl et al.⁸ reported that the most common fracture types in abused children were transverse (36%), spiral (36%), and oblique (7%). Comparatively, the most common fracture types in falls were spiral (37%), transverse (33%), and oblique (14%). CMLs, however, are highly associated with abuse. Beals and Tufts⁴⁸ reported 4 of 24 of femur fractures in abused children to be CML type fractures, but only 1 of 39 in children with accidental fractures was this type. Loder and Bookout⁵⁰ reported that 28% of long bone fractures in abused children are CMLs, and Kleinman et al.⁴¹ reported that CMLs were the most common fracture type in 31 fatally abused infants.

Case-Based Fall Studies

Several studies have focused on injuries and fatalities associated with falls in children. It has been well established that fatalities rarely occur in short distance falls. In an early study of 34 free-falls in children, only two fatalities were reported.⁵¹ One was a 9-year-old who fell 40 feet; the other was an 8-month-old who fell nearly 37 feet head-first. Additionally, for feet-first falls from heights less than 25 feet, no injuries were

reported. In two additional studies, all children who fell three stories or less survived.¹⁵⁻¹⁶ A few studies have reported deaths resulting from short-distance falls, but the validity of these results have been debated. Hall et al.⁵² found 18 fatalities in falls from less than or equal to 3 feet, all due to severe head injuries. It has been argued that many of these deaths were actually due to abuse.⁵³ Chadwick et al.⁵⁴ found 7 fatalities from falls less than or equal to 4 feet, but only 1 fatality in 183 falls from 5 - 45 feet. However, the authors concluded that the 7 fatal falls from less than 4 feet likely had false histories. Plunkett⁵⁵ reported 18 fatal cases of head injuries due to falls from 2 – 10 feet from playground equipment. Chadwick et al. reported a mortality rate of less than 0.48 deaths per 1 million young children per year in falls less than 1.5 m (4.9 ft).⁵⁶

A few studies have investigated the types of injuries associated with bed falls. The first studies of this kind were done by Helfer et al.²² and Nimityongskul and Anderson⁵⁷. Both of these studies reviewed records of in-hospital falls and found no serious or life-threatening injuries. Nimityongskul and Anderson⁵⁷ found only one skull fracture and one tibia fracture in 76 cases. The remainder of the cases had no injuries or only minor swelling, contusions, bruises, and lacerations. Also, the tibia fracture occurred in a child with osteogenesis imperfecta. Helfer et al.²² reported one skull fracture out of 85 cases, and also found an additional two skull fractures, a humeral fracture, and three clavicle fractures in a survey asking parents in private pediatricians offices to report fall incidents (161 cases).

The studies by Helfer and Nimityongskul found serious injuries to be rare as a result of in-hospital bed falls. However, there are several other studies that found injuries from bed falls to be much more common.⁵⁸⁻⁶⁰ Henrikus et al.⁵⁹ found 115 patients with

orthopedic injuries resulting from bed falls or falls from other furniture surfaces over a 20-month period. The injuries included fractures and dislocations primarily of the upper extremities. The estimated maximum fall height across all cases was four feet. Child abuse was suspected in six of the 115 cases. Other studies found an increased injury risk when including falls from bunk beds. Macgregor⁶⁰ investigated 85 cases of falls from both upper and lower bunks as well as falls from conventional beds and cots and found 52% of the children to have significant injuries. Ten of the cases with significant injuries were due to falls from top bunks. In total, there were 25 fractures, 27 head injuries (no skull fractures or intracranial bleeding), 12 lacerations, and 21 soft tissue injuries. Belechri et al.⁵⁸ compared injury risk between falls from bunk beds and conventional beds. Out of 1881 reported bed fall injuries, 10.5% were from bunk beds, 10.4% were from cribs, 3.1% were from cots, and 76.0% were from conventional beds. However, the injury severity associated with the bunk bed falls was greater than that for conventional bed falls, and bunk bed falls were more likely to require hospitalization.

A study by Tarantino et al.⁶¹ investigated injuries resulting from short vertical falls (less than 4 feet) in infants less than 10 months of age. Of 167 subjects, 85% had minor or no injury and 15% had significant injuries. Significant injuries included seven long bone fractures (three femur, one humerus, two tibia, and one clavicle), and 18 closed head injuries. Two patients with intracranial hemorrhages were later determined to be victims of abuse. This study also compared injury outcomes by fall mechanism and discovered that being dropped by a caretaker was more likely to be associated with significant injury than rolling or falling from a bed or couch.

Although there are several studies that describe injuries associated with falls from beds and other furniture, few have attempted to describe falls biomechanically. Lyons and Oates²³ used a case-based approach similar to the previously described studies, but also estimated the momentum associated with each of the falls. This study reviewed 207 cases of in-hospital falls from a bed or crib, and based on the estimated fall height and weight of the child, the momentum at impact was calculated for each fall. The cases were separated into those with injuries (31 cases) and those without injuries, and the momentum was compared between the two groups. The injuries included 29 minor contusions or lacerations, one clavicle fracture, and one skull fracture. No significant difference was found in the impact momentum between the injured and non-injured groups.

Limitations of Case-based Studies

These studies provide a base of knowledge for the types of injuries that would be expected in falls or in cases of child abuse. However, they are limited by the fact that they rely on an assumption of whether the injuries are abusive or accidental. Incorrect assumptions can result in false conclusions, and cases of child abuse are commonly mistaken for accidental trauma. One study found 31% of cases of abusive head trauma were missed by a physician.⁴³ In some cases it took as many as 9 visits to the physician to recognize the abuse. Among the missed cases in this study, 28% suffered further injuries and 41% suffered medical complications as a result of the missed diagnosis. Some studies have tried to correct for this error by using an algorithm that takes into account injury type, associated injuries, and the given history, but even this relies on the

assumption that certain injuries are indicative of abuse.⁴ Another study of injuries resulting from free falls included only falls that were witnessed by someone other than the caretaker.⁶² A biomechanics approach, combined with a case-based approach, can provide vital information about relationships between fall characteristics, child characteristics, and injury potential that may aid clinicians in more accurate child abuse diagnoses.

Biomechanical Studies of Injury Risk in Falls and Abuse

Surrogate Studies

Anthropomorphic surrogates have been utilized in studies to determine injury risk in falls as well as abusive events such as the shaken baby syndrome. Duhaime et al.⁶³ first used anthropomorphic surrogates of a 1-month-old infant in simulations of shakes and shakes with impact. In this study, dolls were modified to match the head and body weight of a 1-month-old. The models were tested with and without an added “skull” for variable deformability of the head. Three different neck models were also tested (one hinge neck and two hollow rubber necks of different thickness and stiffness) to determine the effect of varying neck stiffness on the resulting parameters. Accelerations of the head were measured by a single accelerometer at the top of the head. The surrogates were vigorously shaken and then the back of the head was impacted against either a metal bar or a padded surface. The authors found that the accelerations associated with impact were much greater than those for shaking alone, and that the acceleration levels for shaking alone did not exceed injury thresholds for concussion, subdural hematoma, or

diffuse axonal injury. However, those accelerations resulting from impact exceeded thresholds for all three injury types. In shaking, the more flexible neck was associated with significantly greater accelerations and significantly shorter durations, but the neck condition had no effect in impact situations. The presence of the added skull was found to have no significant effect. Impacts against a padded surface had significantly smaller accelerations and significantly longer durations than impact onto a metal bar.

A more recent study built upon that by Duhaime by using a more biofidelic infant surrogate. Prange et al.⁶⁴ simulated shaking and shaking with impact as in the previous study, as well as several short distance falls using a 1.5-month-old surrogate. A hinged neck was used to represent a worst-case scenario, and the “skull” and “scalp” materials were chosen to represent infant skull properties. An angular rate sensor attached to the top of the head measured angular velocities. Angular accelerations were then calculated by taking the derivative of the velocity. Falls were simulated for three different fall heights (1, 3, and 5 feet) and three different surfaces (4 inch thick foam, 0.25 inch thick carpet pad, and a concrete floor). The same surfaces were also used in simulations of inflicted impacts, except a stone bench was used instead of the concrete floor. The surrogate was initially in a horizontal position for fall experiments with the head slightly lower than the body to ensure that the head would contact first. Overall, falls from greater heights and falls onto harder surfaces resulted in greater angular accelerations. For the shaking and impact scenarios, it was found that inflicted impacts against the carpet pad and stone surfaces resulted in significantly greater accelerations and lower time durations than those from impacts against foam or from shaking. The authors concluded that shakes produced responses similar to those from minor falls, but inflicted

impacts produced responses that were significantly higher, and therefore were more likely to be associated with brain injuries.

Studies by Coats and Margulies⁶⁵ and Ibrahim and Margulies⁶⁶ investigated head injury potential in short-distance falls using 1.5-month-old infant and 18-month-old toddler surrogates, respectively. The skull and neck properties of the surrogates were designed to replicate that of a human child (using material property data obtained from pediatric cadaver specimens). In both studies, the surrogate was dropped from three heights (1, 2, and 3 ft) onto various surfaces (mattress, carpet pad, concrete). The surrogate was positioned so that the initial impact would occur to the occiput. Three-dimensional head angular accelerations were measured. Both studies reported increases in peak angular acceleration with increasing fall height and increasing surface stiffness. Additionally, peak axial rotation accelerations were as high as peak sagittal plane accelerations. Peak coronal plane accelerations were significantly lower than those for sagittal and axial rotations. Peak head angular accelerations for the toddler were nearly double those of the infant. Based on these comparisons, Ibrahim and Margulies concluded that the toddler is likely less vulnerable to skull fracture (due to a greater skull thickness) but more vulnerable to neurological injuries (due to greater peak accelerations) than the infant.

There have been several studies by Bertocci to investigate injury risk associated with short-distance falls using anthropomorphic test dummies. In one study, Bertocci et al.⁶⁷ simulated bed falls using a Hybrid II 3-year-old test dummy. Feet-first free falls were simulated in another study using the same Hybrid II test dummy.⁶⁸ In both studies, linear head acceleration, pelvis acceleration, and femur loads (including compression,

bending, and torsional loads) were measured. Head Injury Criteria (HIC) were calculated as a measure of head injury risk. Four different impact surfaces were tested (linoleum, wood, padded carpet, and playground foam), and for the free falls, three different heights were tested (27, 47, and 64 inches measured from the ground to center of mass of the dummy). Only one fall height (27 inches) was tested in the bed fall simulations. In free fall experiments, it was found that fall height had no significant effect on either head acceleration or HIC, although it did have some effect on femur loading. Impact surface type was found to have a significant effect on head acceleration and HIC in both studies with playground foam producing the lowest values. Despite these effects, there was a low risk of contact-type head injury for all surfaces and heights tested.

Several studies have examined the effects of varying fall conditions on injury risk. In addition to fall height and impact surface which have been tested in the previously mentioned studies, Deemer et al.⁶⁹ also investigated the effects of falls onto wet versus dry surfaces. Using a 3-year-old Hybrid III test dummy, short-distance feet-first free falls onto wet and dry linoleum surfaces were simulated. It was found that head acceleration and HIC were significantly greater on the dry surface; however femur compressive and bending loads were significantly greater on the wet surface. Cory and Jones⁷⁰ developed a simulation system to test the head injury potential of different surface mixtures. Several top surface layers, including carpets and linoleums of various thicknesses and types, were tested over three underlying surfaces (wood, concrete, and chipboard). The authors found that while the top surface type and thickness has some effect, the underlying surface primarily dictates the risk of head injury. It was also found that locations on the floor directly over joists produced the greatest head injury risk.

Computer Simulation

Computer simulation is a useful tool to investigate injury-producing events, and to study the effect of changing event parameters on injury risk. Computer simulation has been widely used by the automotive industry to study car crash events, and has also been used in a few studies to investigate falls. Several different software types exist including MADYMO (Mathematical Dynamic Models, TNO, Netherlands), Dynaman (GESAC, Boonsboro, Maryland), and Visual Nastran 4D (MSC software, Santa Anna, California). MADYMO is unique in that it can combine multi-body modeling and finite element techniques. Also, MADYMO contains a database of validated models of anthropomorphic test devices (ATDs) including the CRABI child dummies.

Among the first studies that used computer simulation to investigate injury risk in falls was that by Mohan et al.⁷¹. In this study, detailed investigations were performed for 30 cases of head-first free falls in children aged 1-10 years old and for one head-first fall in a 21 year old adult. Seven of the cases were then selected for further analysis using computer simulation. These included six children aged 1.1 to 6.5 years falling from heights of 3.1-9.9 m and one 21 year old adult falling 3.4 m. The cases were reproduced using the MVMA Two-Dimensional Crash Victim Simulation Model. Since this was the first use of this software to simulate falls, free-fall experiments were performed with an instrumented anthropomorphic dummy and then used to validate the model. Head and pelvis accelerations and overall body kinematics in the model were found to correlate well with the dummy experiments. The 2-D model consisted of a nine mass, ten segment body linkage. For each case that was simulated, biomechanical data was used to define the material properties of the head and each body region, and based on the fall victim's

height and weight, anthropometric data was used to estimate segment masses, lengths, moments of inertia and joint properties. Head accelerations, peak head deflections, peak normal forces, and energy absorbed were calculated in each model. Additionally, impact angles and surface properties were varied to examine their effect on head response. The authors found a good correlation between increasing head acceleration and increasing head injury severity, except in the case of the youngest child (1.1 years old) who fell the greatest distance but suffered less injuries. Changing impact angle through 20 degrees had no significant effect on the head response in the six child cases. There was a greater effect of changing impact angle in the adult model likely due to the greater torso mass. For falls onto soil and sand surfaces, peak head accelerations were reduced to 30-50% and 15-20% of the rigid surface values, respectively.

O'Riordain et al.⁷² also used computer simulation to reconstruct falls. Four cases of falls that resulted in a focal head injury were modeled using MADYMO. The four cases included a 76 year old who fell backwards off a doorstep (13 cm tall), an 11 year old who fainted and fell directly backwards, a 37 year old who fell off a 136 cm gate impacting the lateral side of the head, and a 24 year old who was standing on a chair (44 cm high) and fell forwards and to the right impacting the lateral side of the head. For each case, the accident site was investigated and witnesses were interviewed to determine the conditions of the fall and the environment. The pedestrian ATD model that most closely represented the fall victim in terms of height and weight was selected for the model. In each case, the initial velocity and head contact properties were varied so that a total of six simulations per case were run. The initial velocities tested included the actual value, and values 0.1 m/s and 0.1 rad/s higher and 0.1 m/s and 0.1 rad/s lower. Two sets

of head contact properties were used: the original characteristics for the ATDs in MADYMO and alternative force-deflection curves from another study of cadaveric head impacts. Peak linear and angular velocities, peak linear and angular accelerations, peak impact forces, and Head Injury Criteria (HIC) were calculated. The authors found that changing the head contact properties had a significant effect on the outcomes. The simulations using the original head contact properties produced higher accelerations and velocities than those using alternative head contact properties, and the injury severity in the simulations with the alternative properties was much closer to the injuries seen in the falls. Changes due to varying initial velocity were less significant and no specific trend was clear.

Bertocci et al.⁷³ used computer to investigate the effects of stair characteristics on injury risk in stair falls. A computer simulation of a 3-year-old child falling down the stairs was developed using Working Model 3D. The child was developed to match the properties of the 3 year old Hybrid III ATD. The effect of varying stair properties (number of steps, slope of stairs, surface friction, and surface elasticity) on injury risk of the upper leg was determined. Upper leg impact velocity, energy, and momentum were determined. It was found that the potential of upper leg injury increases with an increasing number of steps, decreasing surface friction, decreasing surface elasticity, and increasing slope.

In another study, a computer simulation of a pediatric bed fall was developed and validated.⁷⁴ Bed fall experiments were first conducted using a 12-month-old CRABI ATD. The ATD was initially placed horizontally on a 66 cm high surface, and pushed off onto the floor using a pneumatic actuator. The same fall was then recreated in Visual

Nastran 4D. The model was validated by adjusting joint stiffness values and contact properties of the head, torso, arms and legs until the head and torso acceleration curves matched what was measured experimentally. Validation of computer simulation models is necessary to ensure reliability of the results. Of the described studies, only those by Bialczak et al. and Mohan et al. were validated using controlled experiments.^{71, 74}

Injury Criteria

Head Injury

The most widely accepted measure of head injury risk in impacts is the Head Injury Criterion (HIC). HIC was developed for use in the automotive industry to assess risk in motor vehicle crash testing. The HIC have also been used to assess head injury risk in falls, particularly in the playground safety area to determine critical fall heights for playground equipment. It has been stated that the HIC is “considered to be the best model available to predict the likelihood of injuries from falls”.⁷⁵ The HIC is based on the time-history of the linear head acceleration and is defined as

$$HIC = (t_2 - t_1) \left[\frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t) dt \right]^{2.5} \quad (1)$$

where $a(t)$ is the resultant linear head acceleration measured in g's, and t_1 and t_2 , the start and finish times of the acceleration spike. HIC values are calculated over 15 millisecond durations (HIC_{15}) to compare with proposed thresholds. Tolerance limits have been

established by the National Highway Traffic Safety Administration (NHTSA) for ages and sizes corresponding to specific anthropomorphic test dummies (ATDs), including a large adult male, mid-size adult male, small adult female, 6-year-old child, 3-year-old child, and a 1-year-old child (Table 1-1). These limits represent a 31% probability of skull fracture.⁷⁶

Large Male	Mid-size Male	Small Female	6-year-old	3-year-old	1-year-old
700	700	700	700	570	390

Table 1-1. Suggested HIC₁₅ limits for various dummy sizes.

Another method of assessing head injury risk has been to simply consider the maximum linear head acceleration recorded during an impact, sometimes called the “peak g” method. However, there is a wide range of tolerance limits suggested by the literature. Sturtz⁷⁷ reported a critical load value of 83 g for impact durations greater than or equal to 3 ms based on reconstructions of pedestrian accidents. Above this load irreversible injuries are possible. By using computer simulations to reconstruct free falls resulting in serious head injuries, Mohan et al.⁷¹ proposed conservative tolerance limits of 200 – 250 g peak accelerations for children. Others have reported tolerance limits for children ranging from 50 – 200 g where 50 g is the maximum before-injury threshold and 200 g is the threshold for fatal injury.⁷⁵

Neither of the previously discussed methods account for head injury due to rotational loads, which often account for severe brain injuries. Subdural hematoma (SDH) and diffuse axonal injury (DAI) both result from exposure to rotational

accelerations. Sturtz⁷⁷ proposed an angular acceleration limit of 2000 rad/s² for impacts lasting 10 ms or longer. Most other studies have related rotational accelerations to particular injury types. Reported rotational accelerations necessary to cause concussion are 4,500 rad/s² for an adult and 10,000 rad/s² for an infant.¹⁸ Similarly, accelerations necessary to cause severe (DAI) have been reported as approximately 18,000 rad/s² for an adult and 40,000 rad/s² for an infant.¹⁸ Margulies and Thibault⁷⁸ established tolerance curves for DAI based on peak rotational acceleration and peak change in rotational velocities (Figure 1-1). These curves were derived from a combination of animal experiments, physical models, and analytical model simulations. Duhaime et al.⁶³ used a tolerance limit of approximately 35,000 rad/s² for SDH in an infant with a 500 gram brain mass. It has been reported that accelerations necessary to cause acute SDH and deep intracerebral hemorrhage are much greater than those necessary to produce mild DAI.¹⁸

The injury potential is often dependent on the duration of the acceleration pulse. In general, the shorter the acceleration duration, the greater the acceleration necessary to cause injury. This is due to the viscoelastic nature of biological tissues. Also, for a given head acceleration, different types of brain injuries will occur for different durations. Three injury zones have been described for a constant acceleration.⁷⁹ For very short durations (high strain rates), the brain experiences very little strain, so extremely high accelerations are necessary to cause injury. As the duration increases, strains occur on the surface of the brain and cause damage primarily to vascular tissue resulting in SDH, for example. Lastly, as the duration increases further, the strains penetrate deeper into the brain causing damage to the brain tissue. This produces injuries such as concussion and DAI.

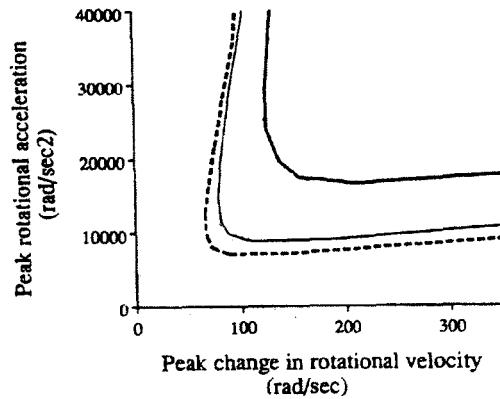


Figure 1-1. DAI thresholds for infant (500 g brain mass, heavy solid line) and adult (1067 g brain mass, solid line; 1400 g brain mass, dashed line).

Neck Injury

NHTSA has also established Neck Injury Criteria, or N_{ij} values, to assess the risk of neck injuries.⁷⁶ These are based on combined axial and rotational loading in the sagittal plane and can be calculated as follows:

$$N_{ij} = \frac{F_z}{F_{int}} + \frac{M_y}{M_{int}} \quad (2)$$

where the subscripts ij represent the four combined loading mechanisms: tension-extension (TE), tension-flexion (TF), compression-extension (CE), and compression-flexion (CF). F_z and M_y are the axial force and flexion/extension moment, respectively, and F_{int} and M_{int} are the critical load values. The critical load values are specific for age of the test dummy and are used to normalize the N_{ij} values. These are presented in Table 1-2. An $N_{ij} = 1$ represents a 22% probability of an Abbreviated Injury Scale (AIS) 3

injury. Neck injuries may include “vertebral fractures, contusions, lacerations, and transections of the cord, as well as brain stem injuries and basilar skull fractures that occur as a result of loading to the neck.”⁷⁶

Dummy	Tension (N)	Compression (N)	Flexion (Nm)	Extension (Nm)
12-month-old	1465	1465	43	17
3-year-old	2120	2120	68	27
6-year-old	2800	2800	93	39
Small female	3370	3370	155	62
Mid-sized male	4500	4500	310	125
Large male	5440	5440	415	166

Table 1-2. Proposed critical intercept values for N_{ij} calculation.

Long Bone Fractures

There are three main failure mechanisms in the long bones: compression, bending, and torsion. The strength of the bone depends upon the direction of the applied load. Adult bone strength has been well-studied and femur and humerus fracture thresholds are shown in Tables 1-3 and 1-4.⁸⁰ Despite this, little information is available on pediatric bone strength. A few studies have investigated pediatric femur strength⁸¹⁻⁸⁴, but no known studies have investigated pediatric humeral strength. Pediatric bone is more plastic than adult bone, and certain fracture types are seen only in immature bone. For example, buckle fractures and greenstick fractures, a type of fracture that does not go through the entire width of the bone but only extends through the cortex, are more commonly seen in pediatric bones.⁸⁰

	Male	Female
Compression (kN)	4.98	3.61
Bending Moment (Nm)	151	85
Torque (Nm)	70	55

Table 1-3. Adult Humerus Fracture Thresholds

	Male	Female
Compression (kN)	7.72	7.11
Bending Moment (Nm)	310	180
Torque (Nm)	175	136

Table 1-4. Adult Femur Fracture Thresholds

Currey and Butler⁸¹ performed static bending tests on cortical bone samples of 18 subjects with ages ranging from 2 to 48 years. The samples from children had lower bending strengths and lower elastic moduli than the adult samples. For children less than five years of age, average bending strengths ranged from 150 to 177 MN/m², and average elastic moduli ranged from 79 to 99 GN/m². For subjects greater than 5 years of age, average bending strengths ranged from 184 to 225 MN/m², and average elastic moduli ranged from 115 to 162 GN/m². The authors also found that peak deflection was greater in children less than 5 years, with values ranging from about 1.7 to 1.9 mm, compared to 1.5 to 1.6 mm in teenagers and 1.1 to 1.3 mm in adults. Martin and Atkinson⁸⁵ also performed bending tests on femoral shaft specimens, and found, in one 2.5 year old subject, a bending strength of 21.2×10^8 dyn/cm² (212 MN/m²) and maximum bending load of 5.3 dyn-cm (53 Nm).

Hirsch and Evans⁸² investigated properties of femur cortical bone in seven infants ranging in age from a newborn to a six month old child. Unlike Curry, Hirsch and Evans

tested the bone specimens under tensile loads. They reported breaking loads and ultimate tensile strengths ranging from 10.5 to 19.5 kg (103-191 N) and 5.68 to 13.16 kg/mm² (56-129 MN/m²), respectively.

Chung et al.⁸³ investigated the shear strength of the capital epiphyseal plate on the femoral head in children aged 5 days to 15 years old. The shear strength tended to increase with age ranging from 3.98 kg/cm² (0.39 MN/m²) in the 5 day old specimen to 14.51 kg/cm² (1.42 MN/m²) in an 8 year old specimen.

Using data from quasi-static bending and compression tests of pediatric femur specimens, Sturtz⁷⁷ estimated the dynamic loads necessary to produce a fracture. This calculation was based on the assumption that dynamic load limits are 20% higher than quasi-static load limits. The dynamic bending fracture criteria for a 7 year old and 3.6 year old child were 116-131 Nm and 62-73 Nm, respectively. Also the dynamic axial (compression) fracture criteria were 1800 and 1000 N for a 6 year old and 3 year old, respectively.

Each of the above studies measured the strength of bone specimens removed from child femurs. Another study measured the load necessary for fracture in whole pediatric cadavers in both quasi-static and dynamic bending tests.⁸⁴ In the quasi-static tests, the thighs 18 subjects ranging from 1 hour to 6 years old were loaded in 3 point bending to fracture. Fracture forces tended to increase with age ranging from 470 N (in a 6 day old child) to 2920 N (in the 6 year old child). Exceptions occurred for a 1 hour old infant and a fifteen month old child which required forces of 2720 N and 5700 N, respectively. Fractures types seen were transverse, oblique, metaphyseal, wedge, and fissure (hairline) fractures. Dynamic tests were performed on 10 subjects aged 2-27 months. In these

tests, the subjects were dropped from a height of 70-90 cm onto an impactor at the lateral mid-thigh. Impact forces ranged from 250 to 2370 N, and impact speeds ranged from 13.3 to 16.8 km/hr. A fracture occurred in only 2 cases. One was a transverse fracture in a 2 month old, and the other was a hairline fracture in a 9 month old.

Limitations of Injury Criteria

The injury tolerance of children is much different from that of adults due to differences in size, structural, and material properties. However, much of the injury tolerance information available for the pediatric population, including the head and neck injury thresholds presented, has been scaled from adult data. This is due to a lack of cadaver and volunteer testing in children. Scaling often takes into account both geometric and material differences, but the information available is limited in its accuracy. Several studies have begun to investigate skull and brain tissue properties in children with the intent of better understanding pediatric tolerance to head injury.⁸⁶⁻⁸⁹ In a study of infant skull and suture properties investigating loading at rates similar to those that would occur in short falls, it was found that pediatric suture deforms 30 times more than pediatric cranial bone and 243 times more than adult cranial bone.⁸⁶ Also, brain tissue properties have been found to be age-dependent.⁸⁸⁻⁸⁹ Thibault and Margulies⁸⁸ applied scaling based on brain tissue properties to angular acceleration thresholds for concussion, subdural hematoma, and diffuse axonal injury originally derived from brain mass scaling alone, and found that the injury thresholds were reduced. Just as differences in skull and brain properties exist between adults and children, it is likely that differences

also exist in the neck, long bones, and other body regions. These differences need to be studied further to develop more accurate pediatric injury criteria.

Child Restraint/Air Bag Interaction (CRABI) 12-month-old Test Dummy

The CRABI 12-month-old anthropomorphic test device (ATD) represents a 50th percentile 12-month-old child in terms of overall height and weight, as well as weights and inertial properties for body segments. Table 1-5 lists weight specifications for the CRABI. Table 1-6 and Figure 1-2 describe the external dimensions of the CRABI.⁹⁰

Biofidelic impact response requirements for the head and neck have been established for the CRABI 12-month-old.⁹¹ These were created by scaling the response requirements of the Hybrid III mid-size adult male ATD based on differences in size, mass, and material properties of bone. The original requirements for the Hybrid III adult ATD were derived from human volunteer and cadaver tests. The head impact response is based on drop tests in which the forehead impacts a flat rigid surface and peak resultant head accelerations are measured. The neck impact response is measured by mounting the ATD head and neck to the end of a pendulum. The pendulum is released and impacted with a block of aluminum honeycomb material. Requirements for neck flexion and extension exist as a function of head to torso angle and the moment about the occipital condyles.

Segment Assembly	Specified Weight	
	Metric (kg)	English (lbs)
Head Assembly	2.64 ± 0.05	5.81 ± 0.11
Neck Assembly	0.38 ± 0.03	0.84 ± 0.07
Torso Assembly	3.68 ± 0.10	8.10 ± 0.22
Arm Assembly	0.60 ± 0.03	1.32 ± 0.07
Leg Assembly	1.05 ± 0.03	2.31 ± 0.07
Total Weight	10.00 ± 0.30	22.00 ± 0.66

Table 1-5. Weight specifications for the CRABI 12-month-old ATD.

Dimension	Description	Metric (mm)	English (in)
A	Total sitting height	469.9 ± 7.6	18.25 ± 0.30
B	Shoulder pivot height	284.2 ± 7.6	11.19 ± 0.30
C	Hip pivot height	33.0 ± 5.1	1.30 ± 0.20
D	Hip pivot from back line	45.2 ± 5.1	1.78 ± 0.20
E	Shoulder pivot from back line	55.4 ± 5.1	2.18 ± 0.20
F	Thigh Clearance	68.1 ± 5.1	2.68 ± 0.20
G	Elbow pivot to fingertip	184.2 ± 7.6	7.25 ± 0.30
I	Shoulder pivot to elbow pivot	106.7 ± 7.6	4.20 ± 0.30
J	Elbow rest height	157.7 ± 7.6	6.21 ± 0.30
K	Buttock to knee length	210.3 ± 7.6	8.28 ± 0.30
L	Popliteal height (reference to seat)	146.3 ± 7.6	5.76 ± 0.30
M	Knee pivot height	172.7 ± 7.6	6.80 ± 0.30
N	Buttock popliteal length	152.4 ± 7.6	6.00 ± 0.30
O	Chest depth with jacket	115.1 ± 7.6	4.53 ± 0.30
P	Foot length	97.5 ± 5.1	3.84 ± 0.20
Q	Stature	740.4 ± 12.7	29.15 ± 0.50
R	Buttock to knee pivot length	183.6 ± 5.1	7.23 ± 0.20
S	Head breadth	129.5 ± 7.6	5.10 ± 0.30
T	Head depth	157.5 ± 7.6	6.20 ± 0.30
U	Hip breadth	166.1 ± 7.6	6.54 ± 0.30
V	Shoulder breadth	208.3 ± 7.6	8.20 ± 0.30
W	Foot breadth	44.2 ± 5.1	1.74 ± 0.20
Y	Chest circumference with jacket	465.1 ± 12.7	18.31 ± 0.50
Z	Waist circumference	459.7 ± 12.7	18.10 ± 0.50
AA	Reference location for chest circumference and chest depth with jacket	261.6 ± 5.1	10.30 ± 0.20
BB	Reference location for waist circumference	111.8 ± 5.1	4.40 ± 0.20
CC	Shoulder height	307.3 ± 7.6	12.10 ± 0.30
DD	Chin height	297.2 ± 7.6	11.70 ± 0.30

Table 1-6. List of external dimensions for CRABI 12-month-old dummy.

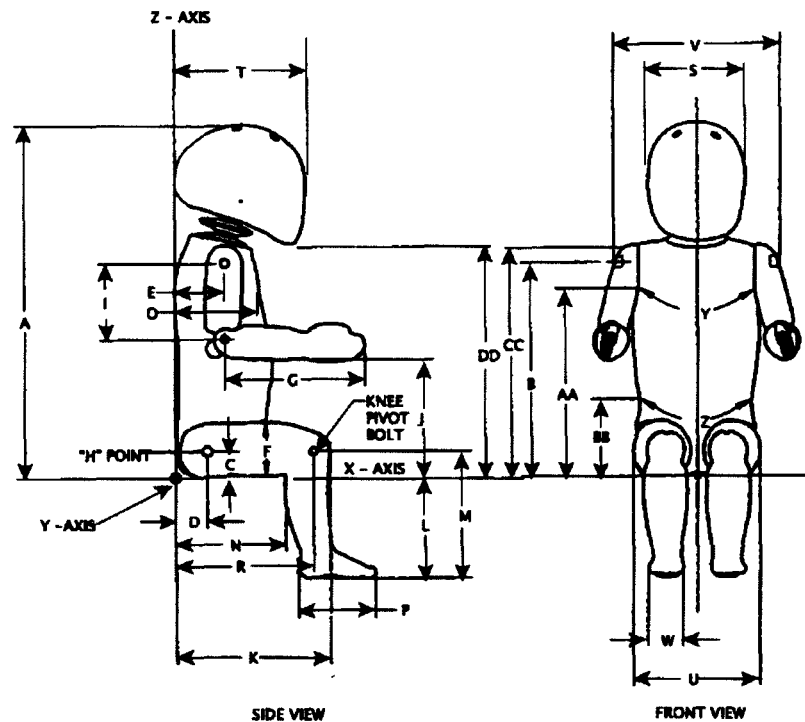


Figure 1-2. Diagram of external dimensions for CRABI 12-month-old dummy.

Summary

Although several clinical studies have investigated injuries resulting from bed or other furniture falls, only one of these considered the effects of momentum on injury risk, and none have explored the effects of fall dynamics on injury risk. Additionally, a few biomechanical studies have investigated loading and injury risk associated with falls. However, these are limited by the biofidelity of the surrogates used and the simplicity of the falls studied. This study is unique because it utilizes computer simulation to investigate the biomechanics of pediatric falls. Computer simulation can be used to understand how slight variations in the fall characteristics can affect the dynamics and injury risk as well as improve upon the limitations of the surrogates used in experiments.

The proposed work will expand on present knowledge, and by investigating the falls from three different perspectives will provide a more complete understanding of the biomechanics of short household falls. The results of this study can be used to aid clinicians in distinguishing between inflicted and non-inflicted injuries. Since this decision often depends on the clinician's experience, objective information about injury risk in these falls will improve the likelihood of earlier identification of child abuse, and also prevent innocent families from false accusations of abuse.

CHAPTER II
PEDIATRIC SHORT-DISTANCE HOUSEHOLD FALLS: BIOMECHANICS AND
ASSOCIATED INJURY SEVERITY

Overview

Short-distance household falls are a common occurrence in young children, but are also a common false history given by caretakers to conceal abusive trauma. The purpose of this study was to determine the severity of injuries that result from accidental short-distance household falls in children, and to investigate the association of fall environment and biomechanical measures with injury outcomes. Children aged 0-4 years who presented to the Emergency Department with a history of a short furniture fall were included in the study. Detailed case-based biomechanical assessments were performed using data collected through medical records, interviews, and fall scene investigations. Injuries were rated using the Abbreviated Injury Scale (AIS). Each case was reviewed by a child abuse expert; cases with a vague or inconsistent history and cases being actively investigated for child abuse were excluded. Seventy-nine subjects were enrolled in the study; 15 had no injuries, 45 had minor (AIS 1) injuries, 17 had moderate (AIS 2) injuries, and 2 had serious (AIS 3) injuries. No subjects had injuries classified as AIS 4 or higher, and there were no fatalities. Children with moderate or serious injuries resulting from a short-distance household fall tended to have fallen from greater heights,

have greater impact velocities, and have a lower body mass index than those with minor or no injuries. Children aged 0-4 years involved in a short-distance household fall did not sustain severe or life-threatening injuries, and no children in this study had moderate or serious injuries to multiple body regions. Biomechanical measures were found to be associated with injury severity outcomes in short-distance household falls. Knowledge of relationships between biomechanical measures and injury outcomes can aid clinicians when assessing whether a child's injuries were the result of a short-distance fall or some other cause.

Introduction

Short falls in children, especially from beds or other furniture, are a common false history given by caretakers to conceal abusive trauma. In up to 70% of cases of children having abusive injuries, the initial explanation for the injuries given by the caretaker is a fall.⁴⁻⁸ However, short household falls are also a common occurrence in young children. A study of emergency department visits by children less than one year of age found that 61% of accidental cases were injuries due to falls.⁹ Clinicians are commonly asked to distinguish between abusive and accidental injuries by determining whether a child's injuries are consistent with the stated cause of the injuries. An improved understanding of biomechanical factors and injury severity in short household falls may aid clinicians in this decision. Early detection of abuse may lead to prevention of further escalating injuries and, in some cases, prevent the death of the child. Additionally, there is

continuing controversy in the medico-legal community regarding whether short distance falls can lead to severe injuries or death.^{52-55, 92}

Several studies have investigated the types of injuries associated with bed falls and other short distance falls.^{22-23, 57-58, 60-62} However, few studies have investigated relationships between biomechanical factors and injury outcomes in short pediatric falls.^{23, 64, 68, 93} The purpose of this study was to determine the types and severity of injuries that result from short-distance household falls in children, and to investigate the influence of fall environment and biomechanical measures on injury outcomes. This was accomplished through detailed case-based biomechanical assessments of short-distance household falls in children who presented to the Emergency Department (ED) of a regional children's hospital. Based on a review of prior studies, the authors hypothesized that serious injuries would make up less than 10% of cases.

Methods

Study Design

This was a prospective, descriptive study approved by the University of Louisville Institutional Review Board (IRB #08.0011) using an informed consent process. To determine injury types and severities occurring in children in common household falls, the medical records of children ages 0-4 years who presented to the ED with a given history of a fall from a bed or other similar furniture were obtained. Interviews with the caregivers and in-depth scene investigations were conducted to obtain information

regarding fall dynamics and to determine biomechanical measures associated with these falls.

Study Setting and Population

Children less than 4 years of age who presented to the ED of Kosair Childrens' Hospital (Louisville, KY) between May 2008 and July 2009 with a complaint of a household fall from a bed, sofa, or similar furniture were eligible for inclusion in the study. A research team was available 24 hours/day, 7 days/week and was notified by triage of eligible patients in the ED. Any children being actively investigated for suspicion of abuse were excluded from the study. Additionally, all cases were reviewed by a study physician with expertise in pediatric emergency medicine and child abuse diagnoses. Each case was rated on a six-point scale as definite abuse, likely abuse, questionable abuse, questionable accident, likely accident, or definite accident using predefined criteria.^{5, 39} These criteria include: completeness and consistency of the given history, whether the injury was consistent with the history, whether there was a delay in seeking treatment, whether the fall was witnessed by someone other than the caregiver, and whether the child's behavior was consistent with the injury. Only cases meeting criteria for definite accident and likely accident were included in the data analysis. The parent/guardian could select one of three options for participation in this study:

Option 1: Review of their child's medical records

Option 2: Caregiver interview and review of child's medical records

Option 3: Investigation of the fall scene at their home, caregiver interview, and review of their child's medical records

Study Protocol

The type of data collected from the medical records, caregiver interviews, and fall scene investigations are shown in Table 2-1. For cases in which interviews and scene investigations were conducted, measurements obtained at the scene investigations were used in place of those obtained during interviews. Additionally, the reliability of the furniture height estimates provided by caregivers was evaluated to assess whether cases without fall scene investigations could be included in the biomechanical analysis.

Medical Record Review	Caregiver Interview	Fall Scene Investigation
Subject age	Subject demographics	Subject height and anthropomorphic measures (if not obtained during interview)
Subject weight	Detailed fall description including pre-fall position, post-fall position, and dynamics	Furniture height
Detailed description of injuries	Approximate height of furniture child fell from	Type of impact surface and underlying subfloor construction
	Type of impact surface and underlying surface	Surface coefficient of restitution (COR)
	Subject height and other key anthropomorphic measurements	

Table 2-1. Type of data obtained from medical record reviews, caregiver interviews, and fall scene investigations.

At the fall scene investigations, surface coefficients of restitution (COR) were measured to quantify impact surface properties. COR is a measure of the conservation of kinetic energy in impacts. COR values were measured using a resiliency tester (IDM

Instruments, Victoria, Australia). The resiliency tester drops a steel ball (15 mm diameter) from a known height onto the impact surface (Figure 2-1). The steel ball bounce height was recorded, and the COR was calculated as the square root of the ratio of bounce height to drop height. Multiple measurements were taken over the impact area to account for variations in floor properties. Because COR is a measure of the interaction between two objects, the COR for a child/surface impact would likely differ from the steel ball/surface impact. However, the COR was measured in this study only for comparative purposes across various household surfaces. Surfaces with a higher COR deform more upon impact leading to longer impact durations (the fall victim comes to a stop more slowly). This reduces the peak accelerations transferred to the victim. Conversely, surfaces with a lower COR deform little and thus have shorter impact durations (the fall victim comes to a stop more rapidly) and greater peak accelerations.⁹³ Greater peak accelerations are generally associated with a greater injury risk. Therefore, injury risk tends to be greater on surfaces with low COR values than surfaces with high COR values.

Measurements of each child's height and weight were used to determine body mass index (BMI).

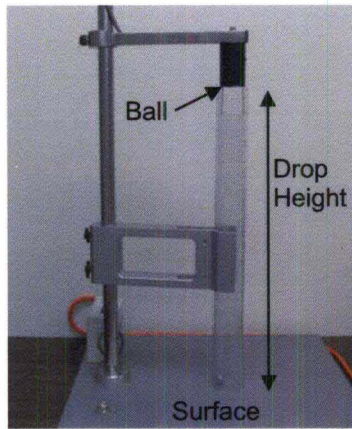


Figure 2-1. Resiliency tester used to measure surface coefficients of restitution.

Data Analysis

Injury Assessment

Subject injuries were rated according to the Abbreviated Injury Scale (AIS). The AIS describes injury severity on a six-point scale (Table 2-2). Injuries are rated using predefined criteria based on location, type (e.g. skeletal injury, vascular injury, muscular injury), and severity.⁹⁴ Each injury was assigned an AIS severity score, and the maximum AIS (MAIS) was determined for each subject.

AIS Code	Description
1	Minor injury
2	Moderate injury
3	Serious injury
4	Severe injury
5	Critical injury
6	Maximal (currently untreatable) injury

Table 2-2. Abbreviated Injury Scale Code Descriptions.

Biomechanical Assessment

To better characterize the fall event, several biomechanical measures were assessed. The impact velocity was determined using

$$V = \sqrt{2gh} \quad (1)$$

where g is the acceleration due to gravity (9.81 m/s^2), and h is the fall height. The fall height was defined as the distance from the child's center of mass at the start of the fall to the ground. The fall height was estimated based on the height of the furniture surface that the child fell from, the position of the child just prior to the fall, and anthropometric measures of the child. The potential energy was determined using

$$E = mgh \quad (2)$$

where m is the mass of the child, h is the height of the fall, and g is the acceleration due to gravity. Finally, the change in momentum during impact was determined using

$$M = mV(COR + 1) \quad (3)$$

where m , V , and COR are the mass of the child, impact velocity, and coefficient of restitution of the impact surface, respectively.

Statistical Analysis

A power analysis was conducted using preliminary data to determine the sample size required to test the hypothesis that less than 10% of subjects would have serious injuries (MAIS 3 or greater). Using 85% power and alpha equal to 0.05, this analysis revealed a desired sample size of 76. To determine whether biomechanical variables were related to injury severity, subjects were divided into two injury severity groups: those with no or minor injuries (MAIS 0 and 1) and those with moderate or serious injuries (MAIS 2 or greater). For each of the continuous independent variables obtained (impact velocity, energy, change in momentum, fall height, impact surface COR, and child factors including mass, age, and body mass index), t-tests were performed to determine if there were significant differences between subjects with no/minor and moderate/serious injuries. In cases where the assumptions of normality were not met, the non-parametric equivalent test was used. For each of the categorical variables obtained (impact surface and sub-floor types, pre-fall and post-fall positions of the child, general fall dynamics, and whether or not the child was in motion prior to the fall), chi-square tests were performed to determine if the variables were significantly associated with injury severity level (no/minor vs. moderate/serious injuries). Statistical analysis was performed using SPSS v12.0.1. Statistical significance was defined as $p < 0.05$.

Results

Figure 2-2 provides details of study participation. Seventy-nine cases met the criteria for analysis. The subjects ranged in age from 1-47 months with a mean age of 18

months. Fifty-four percent (54%) of the subjects were male. Sixty-five percent (65%) of subjects were White, 29% were African American, and 6% were Hispanic.

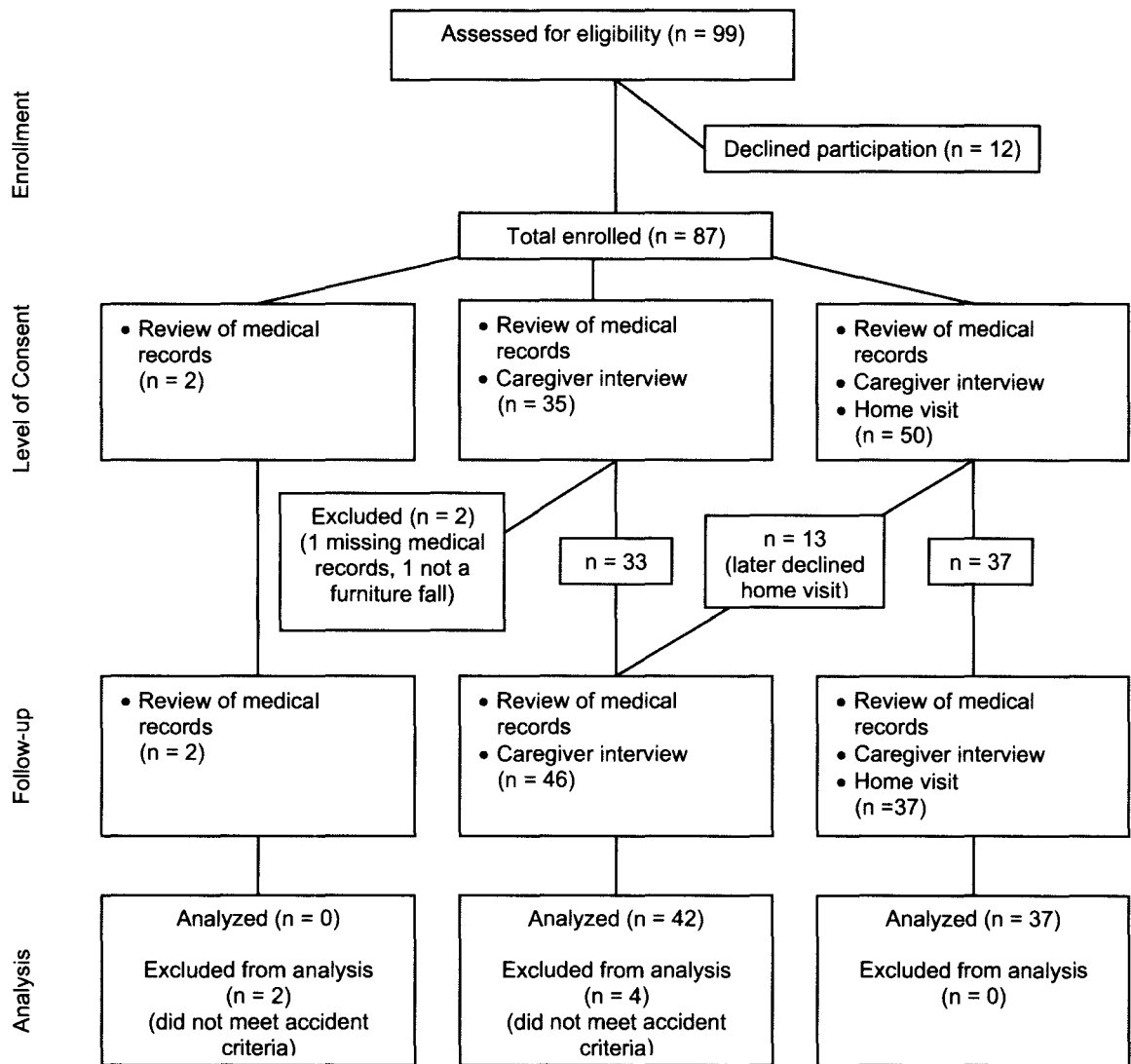


Figure 2-2. Flow diagram of subject progression through study.

Injury Assessment

Figure 2-3 shows the distribution of cases based upon MAIS injury level. No subjects had injuries classified as AIS 4 or higher. Injuries classified as AIS 1 included

mostly lacerations and contusions and 2 cases with radial head subluxation (Nursemaid's elbow). AIS 2 injuries consisted of fractures (6 skull, 2 clavicle, 3 radius and ulna, 4 supracondylar humerus, 1 femur, and 1 metatarsal). There were two AIS 3 injuries which were both small isolated subdural hematomas. The first was a 3 mm subdural hematoma located in the left posterior frontal region. The second was a very thin right frontoparietal subdural hematoma accompanied by a right parietal minimally depressed skull fracture. Both children with subdural hematomas were clinically well-appearing and had no neurological abnormalities. No subjects had AIS 2 or greater injuries to more than one body region.

Six cases were excluded from the study because the history was inconsistent or vague. These cases did not meet criteria for definite or likely accidents. These subjects were not excluded on the basis of injury severity; the injuries of the excluded subjects were no more severe than those of subjects included in the study. Of the 6 excluded cases, 5 children had injuries that were classified as MAIS 1 and 1 child had injuries that were classified as MAIS 2 (clavicle fracture).

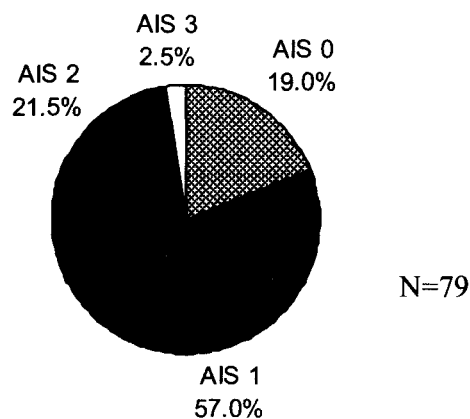


Figure 2-3. Distribution of accidental cases by Maximum Abbreviated Injury Scale (MAIS).

Reliability Analysis

Impact velocity, potential energy, and change in momentum are dependent on measurements of furniture height and COR obtained at the fall scene investigations. Because only estimates of furniture height (from caregivers) were available for most cases (n = 42), the reliability of these estimates was evaluated. For 35 cases, both an estimate of the furniture height (obtained from caregiver during interview) and a measurement of the true furniture height (obtained during fall scene investigation) were available (Figure 2-4). The intraclass correlation coefficient (ICC) between these two data sets was 0.76. In general, caregivers tended to overestimate the height. To account for this bias, the linear relationship between the height estimates and height measurements was determined. This resulted in the following equation which was used to predict furniture height based on the caregiver-provided height estimate:

$$\text{Predicted furniture height(cm)} = 0.718 * \text{Estimated furniture height(cm)} + 11.736 \quad (4)$$

The coefficient of determination (R^2), a measure of the goodness of fit between the linear equation and height data, was 0.80. For each case without a scene investigation, the predicted height was calculated using the caregiver-estimated height collected during the interview. The predicted furniture heights were then used in the biomechanical analysis in place of the estimated heights.

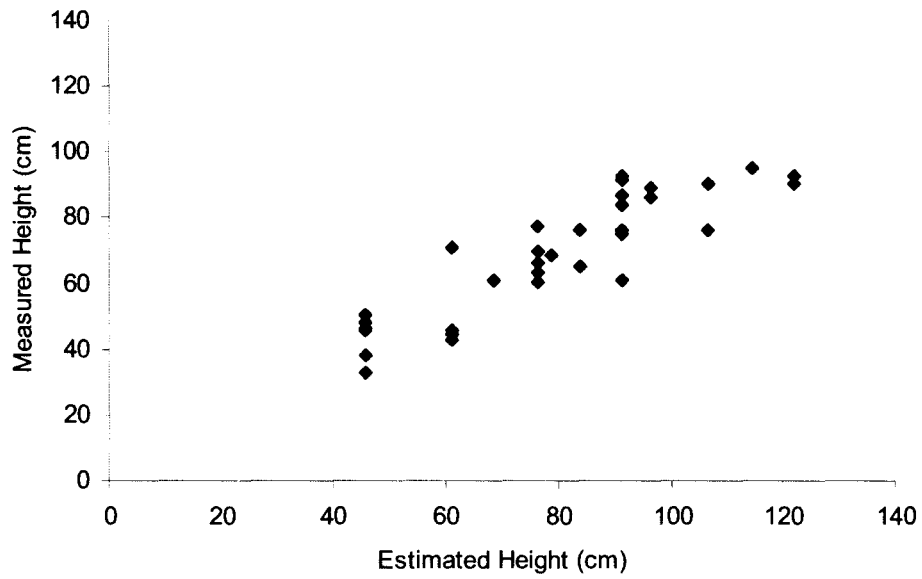


Figure 2-4. Measured furniture height (from fall scene investigation) vs. estimated furniture height (from caregiver interview) for 35 cases.

Biomechanical Assessment

Several fall, environment, and child characteristics were investigated to determine whether there were significant differences between subjects with no or minor injuries and subjects with moderate or serious injuries (Table 2-3).

Measure	Subjects with no or minor injuries		Subjects with moderate or serious injuries	
	Mean (95% confidence interval)	Number of subjects	Mean (95% confidence interval)	Number of subjects
Subject age (months)	17 (14-20)	60	21 (13-28)	19
Subject mass (kg)	10.9 (10.0-11.7)	60	11.2 (9.4-13.0)	19
Subject body mass index ^{*,a}	17.7 (17.2-18.3)	58	16.4 (15.3-17.6)	17
Furniture height (cm) [*]	62.0 (57.3-66.7)	60	75.6 (68.1-83.0)	19
Fall height (cm) [*]	79.8 (73.6-86.0)	60	91.3 (83.9-98.8)	19
Impact velocity (m/s) [*]	4.0 (3.8-4.2)	60	4.3 (4.2-4.5)	19
Potential energy (Nm)	91.3 (79.6-103.0)	60	107.3 (85.7-128.9)	19
Change in momentum (kgm/s) ^b	56.2 (48.1-64.3)	26	57.8 (40.4-75.1)	11
Surface coefficient of restitution ^b	0.39 (0.35-0.43)	26	0.39 (0.30-0.48)	11

*. indicates a statistically significant difference ($p < 0.05$)

a. body mass index was not calculated for four subjects due to missing subject height data

b. includes only cases in which a fall scene investigation was conducted

Table 2-3. Subject, fall environment, and biomechanical measure mean values by injury category.

Fall Environment

The frequency distributions of falls based upon type of furniture and impact surface are shown in Figures 2-5 and 2-6. Twelve subjects were initially placed in a car-seat, bouncy seat, or adult's lap prior to the fall, but this factor was not found to be significantly associated with injury severity. Surface COR measurements were obtained for 37 cases. The mean COR for cases with similar surface/sub-floor combinations is shown in Table 2-4. Neither surface, subfloor, nor COR were found to be significantly associated with injury severity.

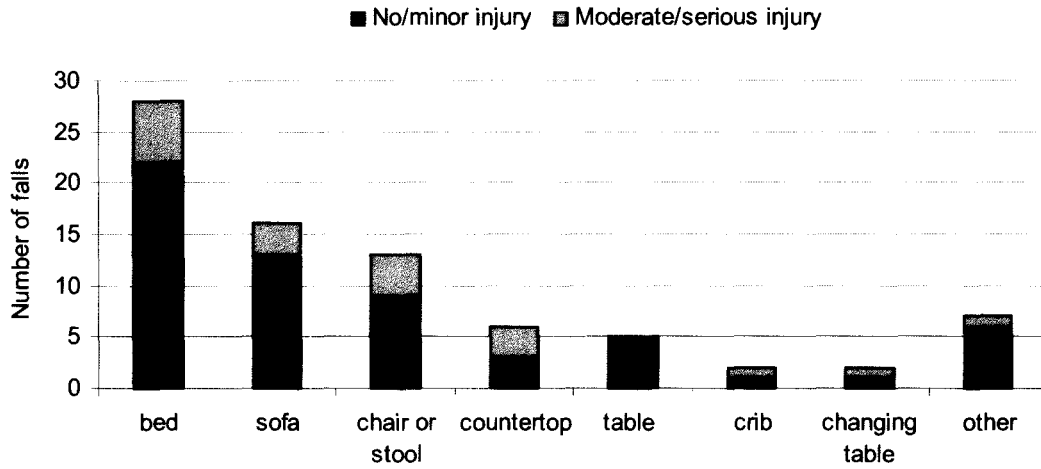


Figure 2-5. Frequency distribution of falls for each injury severity category based upon furniture type. N=79

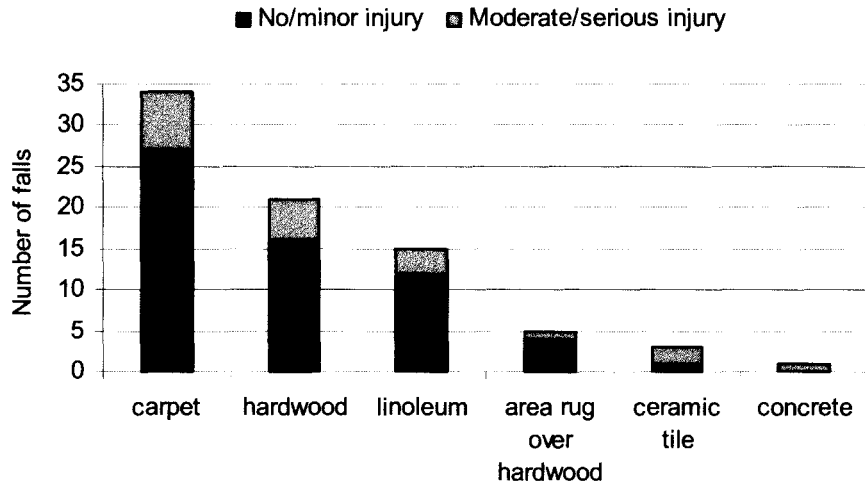


Figure 2-6. Frequency distribution of falls for each injury severity category based upon impact surface type. N=79

Subfloor	Surface	Mean COR (standard deviation)	Number of Falls
Wood	Carpet	0.49 (0.05)	12
	Area rug over hardwood	0.41 (0.02)	5
	Hardwood	0.33 (0.03)	9
	Linoleum	0.35 (0.08)	5
	Ceramic tile	0.44	1
Concrete	Carpet	0.51	1
	Hardwood	0.16	2
	Linoleum	0.17	1
	Ceramic tile	0.47	1

Table 2-4. Mean coefficient of restitution (COR) measured for each impact surface-subfloor combination. Measurements were obtained for only 37 cases where fall scene visits were conducted.

Fall Characteristics

Information regarding the child's position just prior to the fall was obtained in 69 cases, and information regarding the child's position immediately after the fall was obtained in 67 cases (Table 2-5). However, a description of the fall dynamics was obtained in only 40 cases (Table 2-5). This is because nearly half of the falls were not witnessed (44%). Pre-fall position, post-fall position, and description of fall dynamics were not significantly associated with injury severity. Additionally, 25 subjects were noted to have been in motion prior to the fall (e.g., jumping, crawling, or rolling). However, this factor was not significantly associated with injury severity.

Pre-fall Position	Number of Cases	Fall Dynamics	Number of Cases	Post-fall Position	Number of Cases
Lying prone	7	Head-first	16	Lying prone	26
Lying supine	18	Feet-first	2	Lying supine	18
Side-lying	1	Tumbling	14	Side-lying	13
Sitting	21	Other	8	Sitting	7
Standing	16			Other	3
Other	6				

Table 2-5. Frequency distribution of falls based upon pre-fall positions, descriptions of fall dynamics, and post-fall positions. Information not available for all cases.

Biomechanical Assessment of MAIS 3 Cases

The fall that resulted in a 3 mm left posterior subdural hematoma was a fall from a sofa involving a 42-month-old female. The child was seated on the back of the sofa, approximately 1 meter high, and fell backwards. She landed on her side and hit her head on the hardwood floor. This child had a mass of 11.8 kg and a BMI of 12.7. She was in the 45th percentile for her age by height, but only the 5th percentile by mass. The estimated impact velocity, energy, and change in momentum for this fall were 4.7 m/s, 259 Nm, and 74 kgm/s, respectively.

The subject whose fall resulted in a thin right frontoparietal sudural hematoma and skull fracture was 1-month-old male. In this case, the child was sleeping on his mother's chest while she was lying in bed. The mother fell asleep and rolled over causing the child to fall off the side of the bed. He struck his head on a humidifier that was adjacent to the bed and then landed supine on the carpeted floor. The fall scene investigation revealed a bed height of 86 cm and a COR of 0.56 for the carpet. These measurements produced impact velocity, energy, and change in momentum values of 4.6

m/s, 118 Nm, and 40 kgm/s, respectively. The child had a mass of 5.5 kg and a BMI of 12.6. He was in the 95th percentile for his age by both height and mass.

Discussion

Serious injuries resulting from pediatric short household falls are rare. Less than 3% of the cases seen in this study were classified as a serious injury (MAIS 3), and no severe or life-threatening injuries were seen. Seventy-six percent (76%) of the cases in this study had no injuries or only minor injuries. These results are consistent with other studies of injuries resulting from short distance pediatric falls. Previous studies report no severe or life-threatening injuries and fracture rates ranging from 1-29% (mean 13%).^{9, 22-23, 57-58, 60-61, 95-96} The rate of fractures seen in our study (all AIS 2 injuries) was 21.5%. Very few studies have reported intracranial hemorrhages resulting from short distance falls.^{52, 55} The two subdural hematomas seen in this study were small contact type injuries. The clinical presentation and course for each of these children was benign. Both subdural hematomas resulted from falls from heights over 1 m (distance from floor to center of mass of the child). The impact velocities in these cases were similar. The sofa fall involving the 42-month-old child was associated with much greater energy and change in momentum values than the bed fall involving the 1-month-old child. The child fell backwards off the sofa and likely landed directly on her head. Therefore, she was likely unable to have an active protective response to the fall. The 1-month-old child likely struck his head on the edge of the humidifier. The smaller contact area on the edge

of the humidifier would have led to a greater, more concentrated impact force than if he had simply struck the floor.

Our study rated injury severity using the AIS scale. To the authors' knowledge, only two other studies have used the AIS scale to categorize pediatric fall injuries. Morrison et al.⁹⁷ investigated furniture-related injuries in children 0-5 years of age, and found 8% of injuries were AIS 1, 75% were AIS 2, 10% were AIS 3, and 1% were AIS 4. However, Morrison et al. only included injuries that required hospital admission. Thus, the results are skewed towards more severe injuries. Chiaviello et al.²⁴ used a modified AIS scale referred to as the Modified Injury Severity Scale (MISS) to rate stair fall injuries in children 0-5 years of age. The MISS is determined by summing the squares of the AIS scores for the three most severely injured body regions. Therefore, an MISS 2 score represents a condition with only minor injuries (AIS 1) to two body regions. Only 4% of subjects in the Chiaviello study had an MISS > 2. In our study, all subjects with an MAIS 2 or MAIS 3 injury would translate to an MISS > 2. Therefore, we found a much lower incidence of moderate and serious injuries than Morrison et al., but a greater incidence of moderate/serious injuries than Chiaviello et al.

In this study, the MAIS was used as an overall injury score for each subject. Another commonly used overall injury scoring system is the Injury Severity Score (ISS) which is calculated by summing the squares of the AIS scores for the three most severely injured body regions. MAIS was chosen rather than ISS because few subjects had injuries to multiple body regions and no subjects had injuries greater than AIS 1 to multiple body regions.

Of the fall environment factors studied, only furniture height and fall height were found to be significantly different in subjects with no/minor injuries compared to subjects with moderate/serious injuries. The mean fall height for subjects in the no/minor injuries category was 80 cm compared to 91 cm in the moderate/serious injury category. This illustrates that small differences in height (11 cm in this case) can have a significant influence on injury severity outcomes. Other biomechanical studies have shown that increasing fall height leads to an increasing risk of injury.^{64, 68}

Impact surface type was not found to be significantly associated with injury severity for the sample of pediatric falls evaluated in our study. Because there are many variations in surface type, surface COR was measured to quantify the resiliency of the surface-subfloor combinations. However, COR was not significantly different for the two injury severity categories. (We were only able to measure COR on a subset of cases; therefore the lack of significant differences may be due to an inadequate sample size.) Several biomechanical studies have shown impact surface to be associated with injury risk.^{64, 67-68, 70, 93} These studies investigated surface effects in a laboratory setting, where variations in fall dynamics and other environmental factors were controlled. In our study, there were many other factors that could contribute to injury (e.g., fall height, initial position of the child, fall dynamics, and child mass).

In addition to fall environment factors, our study investigated fall dynamics and biomechanical measures. A previous study of short feet-first falls found that fall dynamics played a significant role in measures of head injury risk.⁹³ However, initial position and fall dynamics were not found to be significantly associated with injury severity our current study. Impact velocity, energy, and change in momentum were

determined for each case to further characterize the fall. Only impact velocity was found to be significantly different across the two injury severity categories. Fall energy and change in momentum were determined since they each account for a combination of child and fall characteristics, and it was predicted that these measures would be a better overall measure to compare with injury outcomes. In 25 cases, the subjects were said to have been in motion (jumping, rolling, etc) prior to the fall. However, due to the unreliability of such initial velocity estimates, it was assumed that every child was at rest prior to the fall. If initial velocities had been accounted for, this would lead to an increase in the impact velocity, energy, and momentum values for these cases. A study by Lyons and Oates²³ also assessed impact momentum for pediatric falls and found no significant difference in momentum between the injured and non-injured subjects. With a greater number of subjects and more accurate measures of fall height and surface COR, it is possible that significant differences in energy and change in momentum values between subjects with no/minor injuries and subjects with moderate/serious injuries would emerge.

Child BMI was found to be significantly lower for subjects with moderate/serious injuries compared to subjects with no/minor injuries. Many studies have found a decreasing fracture risk with increasing BMI in adults.⁹⁸⁻¹⁰¹ This is likely due to a protective effect of a greater soft tissue thickness in individuals with a higher BMI.¹⁰¹ Additionally, studies have shown that bone mineral density increases with increasing mass.⁹⁹ Higher bone mineral density suggests a greater bone strength which is often associated with a decreased fracture risk. A few studies have compared BMI and injury outcomes in children. Brown et al.¹⁰² compared injury outcomes in obese and non-obese

children (aged 6-19 years) who were admitted to the intensive care unit and found obese patients suffered less severe (lower AIS) head injuries than non-obese patients, but no significant differences were found in chest, abdominal, and extremity injuries. Rana et al.¹⁰³ found no significant differences in AIS scores between obese and non-obese children (ages 6-20 years) who suffered traumatic injuries but found a lower incidence of closed head injuries and abdominal injuries in the obese patients. Our study found that children with moderate and serious injuries had a lower mean BMI than children with no or only minor injuries. Studies of children have primarily focused on comparisons of obese to non-obese and did not investigate whether an underweight child may have a greater risk for injury. The relationship between BMI and injury severity outcomes in pediatric falls needs to be investigated further.

Limitations

Our study found 21.5% of pediatric falls resulted in moderate injury and 2.5% resulted in serious injury. This number is likely an overestimate of injury severity associated with household falls because only children who presented to the ED were included. Falls are a common occurrence in young children, and often result in no injuries or only minor injuries for which the parents do not seek care.^{22, 104}

The sample size was relatively small. With a greater sample size, differences in energy, change in momentum, surface COR, and other variables could emerge for different levels of injury severity. Additionally, due to the small sample size, each of the variables was analyzed independently for relationships with injury severity. A greater

number of subjects would allow for a multifactor analysis in which interaction between variables could be investigated. In cases where scene investigations were not conducted (n = 42, 53%), furniture heights were predicted based on estimates provided by the caregiver and measurements obtained in cases involving scene investigations. Due to this transformation of the height data, the confidence interval presented is likely underestimated. The predicted furniture heights were further used to determine fall heights, impact velocities and potential energies which would introduce a source of error in these measures and their associated confidence intervals.

Another limitation of this study is the possibility that cases of child abuse were misidentified and included in this study or true accidents were falsely excluded. Since this study sought to examine injury in short distance falls, any cases of abuse that were falsely included in the study could contaminate the findings. In an attempt to reduce this possibility, all cases were reviewed by a child abuse expert and judged to be accidental or abusive using predefined criteria.⁵ Any cases that did not meet criteria for a definite or likely accident were excluded from the data analysis. Six cases had vague or inconsistent histories and therefore, did not meet criteria for definite or likely accident. These cases were excluded from the study. It is worthwhile to note that the excluded cases were all classified as MAIS 1 (minor injuries only) except one which was MAIS 2 (clavicle fracture), and were not excluded on the basis of severe injury.

In this study, COR was used to quantify surface properties. COR describes the interaction between two colliding objects (in this case, a steel ball and the floor surface) and does not represent properties of the surface alone. COR depends in part on the mass of the two colliding objects. Thus, different COR values would exist for the child/floor

surface impact than those measured using the resiliency tester and steel ball. Stiffness and damping properties would better describe surface characteristics but were not easily obtainable in site visits. Future studies should investigate alternative methods for quantification of floor surface properties.

Conclusions

This study provides a comprehensive evaluation of the biomechanics of short-distance household falls and investigated the association of biomechanical and fall environment measures with injury severity. Children aged 0-4 years involved in a short-distance household fall did not sustain severe or life-threatening injuries. No children in this study had moderate or serious injuries to multiple body regions. Furniture height, impact velocity and child BMI were found to have the greatest influence on injury severity outcomes. Children with moderate or serious injuries tended to have fallen from greater heights, had greater impact velocities, and had a lower BMI than those with minor or no injuries. By identifying factors associated with injury outcomes, the results of this study provide first steps toward development of an injury prediction model for short-distance pediatric falls.

CHAPTER III
ASSESSMENT OF INJURY POTENTIAL IN PEDIATRIC BED FALL
EXPERIMENTS USING AN ANTHROPOMORPHIC TEST DEVICE

Overview

Falls from beds and other furniture are common scenarios provided to conceal child abuse but are also common occurrences in young children. To aid clinicians in distinguishing abusive from accidental injuries, this study investigated biomechanical outcomes related to injury potential in falls from beds and other horizontal surfaces using an anthropomorphic test device representing a 12-month-old child. The potential for head, neck, and extremity injuries and differences due to varying impact surface was determined. Linoleum over concrete was associated with the greatest risk of head and neck injury compared to other tested surfaces (linoleum over wood, carpet, wood, playground foam). The risk of severe head and extremity injuries in these falls was low. However, results suggest that fractures, particularly involving the skull and humerus, are possible in these falls. Neck injury potential in falls needs to be studied further as limitations in ATD biofidelity and neck injury thresholds based solely on sagittal plane motion may reduce accuracy in current pediatric neck injury assessments.

Introduction

Falls from beds and other furniture are common scenarios provided to conceal child abuse.⁴⁻⁸ However, short-distance household falls are common occurrences in young children and sometimes result in injury. Because of this, clinicians may have difficulty distinguishing between accidental and inflicted injuries, particularly when the scenario provided is a household fall. Objective information about injury potential in these falls may aid clinicians in distinguishing between abusive and accidental injuries. The biomechanics associated with short falls has been investigated in previous studies, but was primarily focused on head injury outcomes.^{64-68, 93} In this study, biomechanical outcomes relating to head, neck, and extremity injury were determined.

To investigate biomechanical outcomes relating to injury potential in short household falls, simulations of falls from a horizontal surface (representing a bed or other elevated furniture surface) with a 12-month-old anthropomorphic test device (ATD) were performed. In Chapter II, rolling off of a bed or other horizontal surface was found to be the most common short-distance fall scenario in infants and toddlers. Therefore, in this study, the ATD was positioned to recreate this “rolling off the bed” scenario. The effect of different impact surfaces on injury potential was determined.

Methods

Test Setup

A Child Restraint Air Bag Interaction (CRABI) 12-month-old ATD (First Technology Safety Systems, Plymouth, Michigan) was placed on the edge of a 24 in (61 cm) high horizontal surface representing a bed, couch, or other similar furniture (Figure 3-1). The ATD was positioned on the bed in an initial side-lying position and pushed off the surface onto the floor using a pneumatic actuator. The actuator was positioned to impact the ATD in the center of the torso (approximately the center of mass location). The actuator provided a consistent initial force to ensure repeatability. Five different impact surfaces were tested. Nine drops were performed for each test scenario based upon a power analysis of prior experiments.

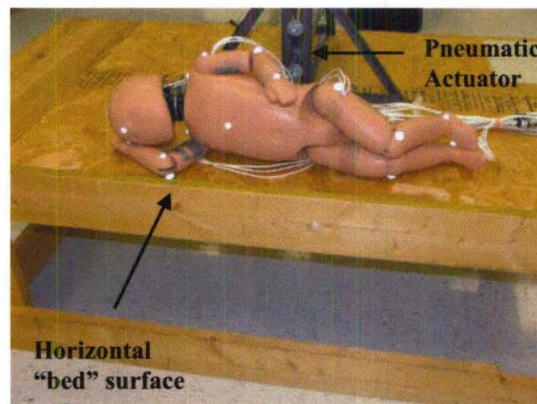


Figure 3-1. CRABI anthropomorphic test device (ATD) in side-lying initial position for bed fall experiments. The pneumatic actuator (used to deliver a force to the posterior torso of the ATD to push it from the surface) is shown behind the ATD.

ATD Instrumentation

The CRABI ATD represents a 50th percentile 12-month-old child in terms of overall height and mass, as well as geometric and inertial properties of individual body segments. The ATD was instrumented with tri-axial linear accelerometers (Endevco, Model 7264-2000) at the center of mass of the head, at the overall body center of mass in the torso, and in the pelvis. Additionally, two angular rate sensors (ATA Sensors, Model ARS-06) were placed at the center of mass of the head to measure angular velocities in the anterior-posterior (AP) and medial-lateral (ML) directions. Two six-axis load cells were located at the superior and inferior aspects of the neck (approximately the C1 and C7 vertebrae locations) to measure neck loads. Three uniaxial strain gages and one shear strain gage (Vishay Micro-Measurements) were adhered to each arm and leg at the center of a metal rod representing the humerus or femur. The strains from the three uniaxial gages (120 degrees apart around the rod circumference) were used to determine humerus and femur axial loads and moments, and the strain measured by the shear strain gage was used to determine torsional loads.

Prior to each fall, ATD joint angles were adjusted using a goniometer to ensure repeated positioning for all testing. Joints were calibrated to manufacturer specifications which are to tighten the joints until the friction is just sufficient to support the weight of the limb.

Impact Surfaces

Five different impact surfaces were tested: playground foam, padded carpet, wood, and two types of linoleum flooring. Carpet, wood, and one of the linoleum

surfaces were placed over a 6 x 6 ft (183 x 183 cm) wooden platform. The platform, built to standard building codes, consisted of 3/4 inch plywood covering 2 x 4 in (5.1 x 10.2 cm) joists spaced 16 in (40.6 cm) from the center of one joist to the center of the next. The carpet was open loop and 1/2 in (1.3 cm) thick with 3/8 in (1.0 cm) thick foam padding underneath and will be tacked to the platform. A layer of 3/4 in (1.9 cm) thick plywood served as the wood surface. The linoleum over the wooden subfloor was no-wax self-adhesive vinyl flooring adhered to the platform (0.039 in or 1 mm thick). The playground foam surface consisted of 2 x 2 ft (61.0 x 61.0 cm) tiles, 2 in (5.1 cm) thick and was placed over a concrete subfloor. The other linoleum surface was linoleum tile 1/8 in (0.32 cm) thick placed over a concrete floor (different from the linoleum used over the wood floor). Coefficients of friction and restitution for the tested surfaces were previously measured.⁹³

Motion Capture

All falls were videotaped (30 Hz) to capture overall fall dynamics. Each fall was also captured using a three-dimensional digital motion capture system (Motion Analysis Co., Santa Rosa, CA). This system consisted of five infrared cameras using a 100 Hz frame rate. For these falls, 48 reflective markers were placed on the ATD (4-5 markers per body segment).

Data Acquisition and Analysis

A LabView program was created for data acquisition. Accelerometer, rate sensor, load cell, and strain data was sampled at 10,000 Hz and filtered according to SAE J211 standards.¹⁰⁵ The filter was a 4th order low-pass Butterworth filter. Head acceleration, angular velocity, and neck force data were filtered with a 1,000 Hz cutoff frequency. Neck moments and femur and humerus strains were filtered with a 600 Hz cutoff.

Head Injury Outcomes

Linear head acceleration was evaluated by examining both the maximum resultant acceleration for each fall and by calculating Head Injury Criteria (HIC). The formula for HIC is defined as

$$HIC = (t_2 - t_1) \left[\frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t) dt \right]^{2.5} \quad (1)$$

where $a(t)$ is the resultant linear head acceleration measured in g's, and t_1 and t_2 , the start and finish times of the acceleration spike. HIC values were calculated over 15 millisecond durations (HIC_{15}) to compare with proposed injury criteria.⁷⁶

Angular head accelerations were determined by differentiating (finite difference method) the measured angular head velocities from the angular rate sensors. Peak angular accelerations, peak change in angular velocities (for the primary impact), and impact durations were determined for each fall for comparison with head injury thresholds.

Neck Injury Outcomes

Peak neck loads at the occipital condyles (transformed from the upper neck load cell) were determined for comparison with proposed injury criteria. Also, neck forces and moments were used to calculate Neck Injury Criteria, or N_{ij} values, for combined axial loading and moments as established by the National Highway Traffic Safety Administration (NHTSA).⁷⁶ N_{ij} were calculated as

$$N_{ij} = \frac{F_z}{F_{int}} + \frac{M_y}{M_{int}} \quad (2)$$

where the subscripts ij represent the four combined loading mechanisms in the sagittal plane: tension-extension (TE), tension-flexion (TF), compression-extension (CE), and compression-flexion (CF). F_z and M_y are the tension/compression force and flexion/extension moment, respectively, measured at the occipital condyles and F_{int} and M_{int} are the critical load values (Table 3-1).

Mechanism	Critical Load
Tension (N)	1465
Compression (N)	1465
Flexion (Nm)	43
Extension (Nm)	17

Table 3-1. Critical intercept values for N_{ij} calculation associated with the 12-month-old CRABI ATD.

Upper and Lower Extremity Injury Outcomes

The measured strains were used to determine the axial compression, bending moment, and torsional load in each humerus and femur¹⁰⁶. The axial compression loads (F) were calculated using

$$F = AE \left(\frac{\varepsilon_1 + \varepsilon_2 + \varepsilon_3}{3} \right) \quad (6)$$

and the bending moments (M) were calculated using

$$M = \frac{IE(\varepsilon_1 - \varepsilon_3)}{\sqrt{3}r \cos \theta} \quad (7)$$

and

$$\theta = \tan^{-1} \left[\frac{1}{\sqrt{3}} \left(1 - \frac{2(\varepsilon_2 - \varepsilon_1)}{(\varepsilon_3 - \varepsilon_1)} \right) \right] \quad (8)$$

where A is the cross-sectional area of the humerus/femur rod, E is the modulus of elasticity, I is the area moment of inertia, r is the radius of the humerus/femur rod, θ is the angle from one of the gages to the axis about which the bending moment is acting, and ε_1 , ε_2 , and ε_3 are the maximum, middle, and minimum measured strains, respectively. The torsional loads on the femurs and humeri were calculated directly from the shear strains measured by the shear strain gages using

$$T = \frac{JG\gamma}{r} \quad (9)$$

where J is the polar moment of inertia, G is the shear modulus of the material, r is the radius, and γ is the measured shear strain.

Statistical Analysis

Each of the outcome variables was analyzed separately using one-way analysis of variance (ANOVA) tests to determine if surface type led to significant differences in the outcome measures. Post-hoc Tukey tests were also conducted to further examine where significant differences occurred. Statistical significance was set at $p \leq 0.05$. SPSS v.12.0.1 was used to perform all statistical analysis.

Results

Fall Dynamics

After actuator contact, the ATD rolled about the edge of the bed surface (Figure 3-2). Initially, the longitudinal (mid-sagittal plane) axis of the body was parallel with the ground. During the fall, the ATD continued to rotate about its longitudinal axis and landed on its side with the head leading (feet still elevated above the floor at the time of impact). The head and left shoulder of the ATD impacted the floor surface at approximately the same time. After the initial impact with the floor, the ATD rebounded upward off the ground before finally coming to rest.

A digital motion capture system was used to quantify fall dynamics. However, a five-camera system proved to be insufficient in tracking fall dynamics. Markers were obscured from view by the bed, resulting in incomplete data. Therefore, data were used only for a qualitative description of fall dynamics.

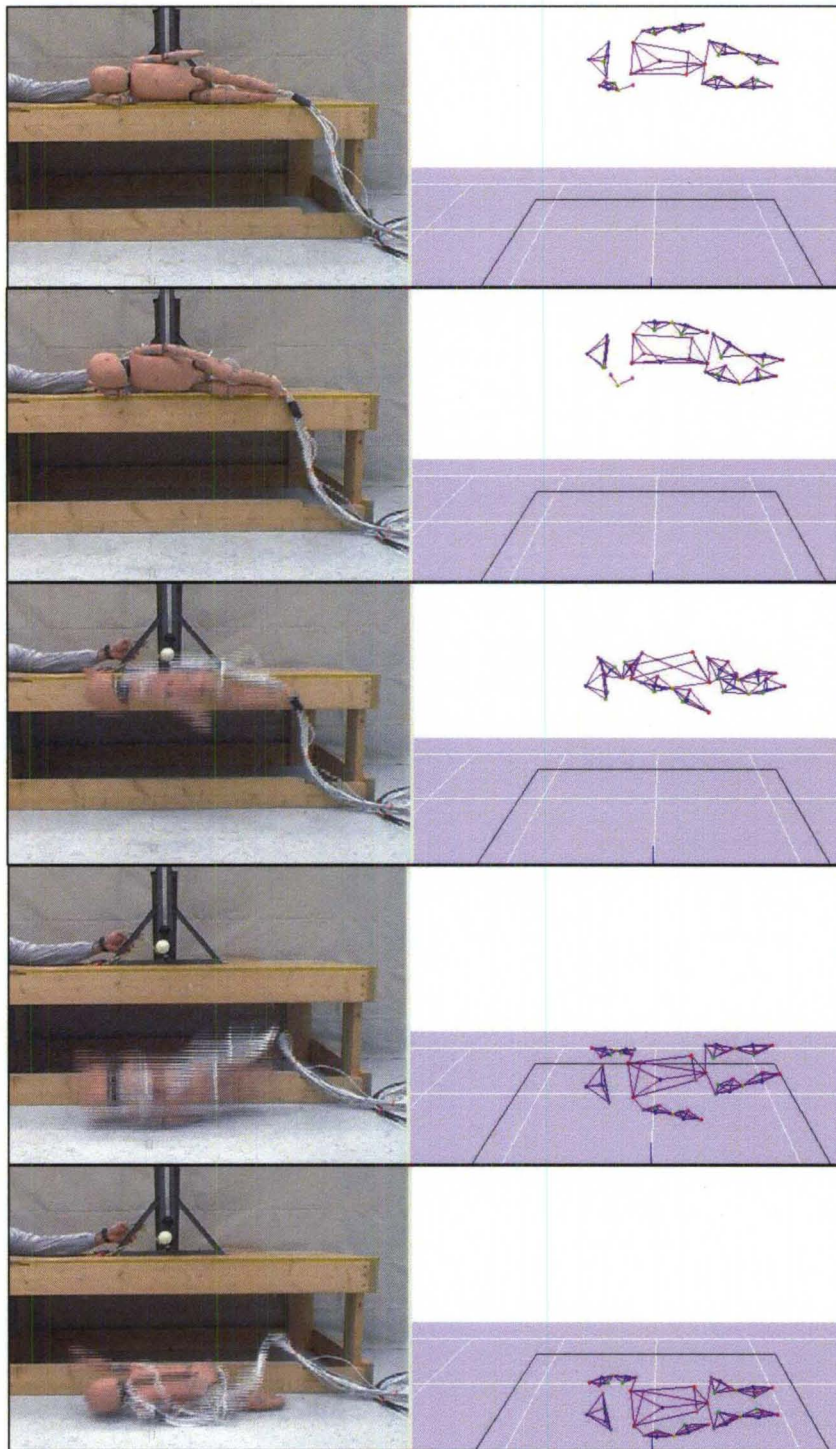


Figure 3-2. Video and motion tracking image sequence of a representative fall onto the linoleum over wood impact surface.

Head Injury Outcome Measures

The mean peak resultant linear head acceleration across all surfaces was 135.6g (Figure 3-3). Falls onto linoleum over concrete produced the greatest values, with a maximum of 423.3g. Linoleum over concrete was associated with significantly greater peak linear head acceleration values than all other surfaces ($p < 0.001$). Additionally, the wood impact surface was associated with significantly greater peak linear head accelerations than playground foam ($p = 0.011$) and carpet ($p = 0.043$).

The mean HIC₁₅ value across all trials was 160 (Figure 3-3). A maximum HIC₁₅ of 334 occurred in a fall onto linoleum over concrete. Linoleum over concrete associated with significantly greater HIC₁₅ values than all other surfaces ($p < 0.001$). There were no other significant differences between other impact surfaces.

The mean peak angular head accelerations across all surfaces were 3,675 rad/s² and 6,172 rad/s² for AP and ML directions, respectively (Figure 3-4). Peak angular head accelerations were generally greater in the ML direction than in the AP direction. The greatest peak ML angular head acceleration was 11,730 rad/s² and occurred in a fall onto linoleum over concrete. As with linear head accelerations, linoleum over concrete was associated with significantly greater peak AP and ML angular head accelerations than all other surfaces ($p < 0.001$). Additionally, wood and linoleum over wood were associated with significantly greater peak ML angular head accelerations than playground foam and carpet surfaces ($p < 0.001$). Wood was associated with significantly greater peak AP angular head accelerations than linoleum over wood ($p = 0.007$), playground foam ($p < 0.001$), and carpet ($p < 0.001$). Linoleum over wood was associated with significantly

greater peak AP angular head accelerations than playground foam and carpet ($p < 0.001$). ML peak angular head accelerations have been plotted along with peak change in angular velocity for comparison with proposed injury thresholds (Figure 3-5).

Head impact durations ranged from 2.7-19.1 ms with a mean of 11.5 ms (Figure 3-6). Linoleum over concrete was associated with significantly shorter impact durations than all other surfaces ($p < 0.001$). Linoleum over wood and wood were associated with significantly shorter impact durations than playground foam and carpet ($p < 0.001$).

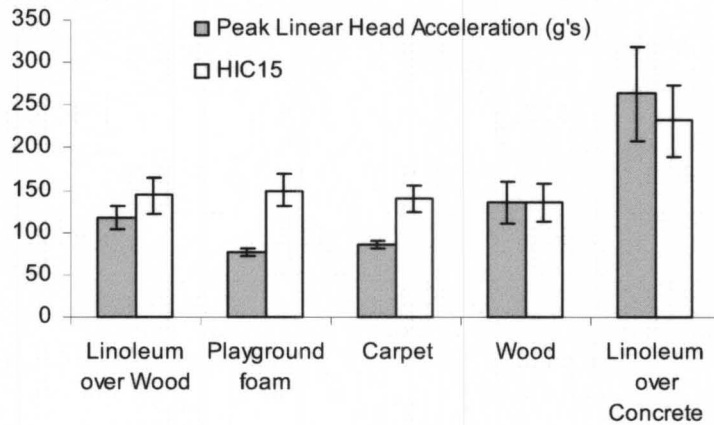


Figure 3-3. Peak resultant linear head acceleration and corresponding HIC₁₅ values for falls onto various surfaces. Error bars represent 95% confidence intervals.

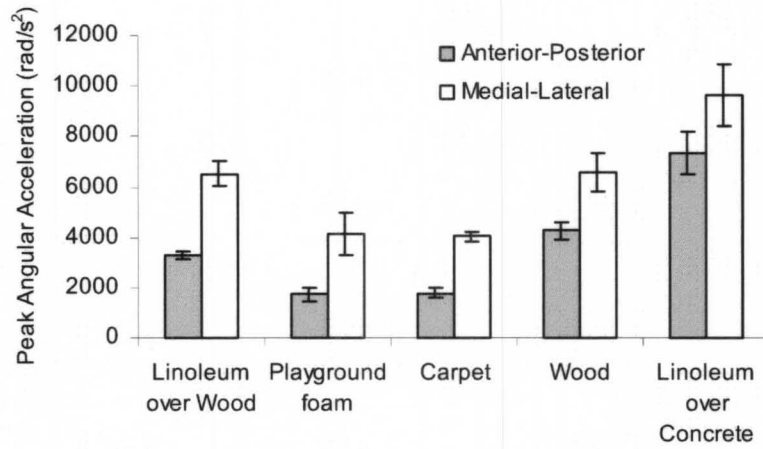


Figure 3-4. Peak angular head acceleration in the anterior-posterior and medial-lateral directions for falls onto various surfaces. Error bars represent 95% confidence intervals.

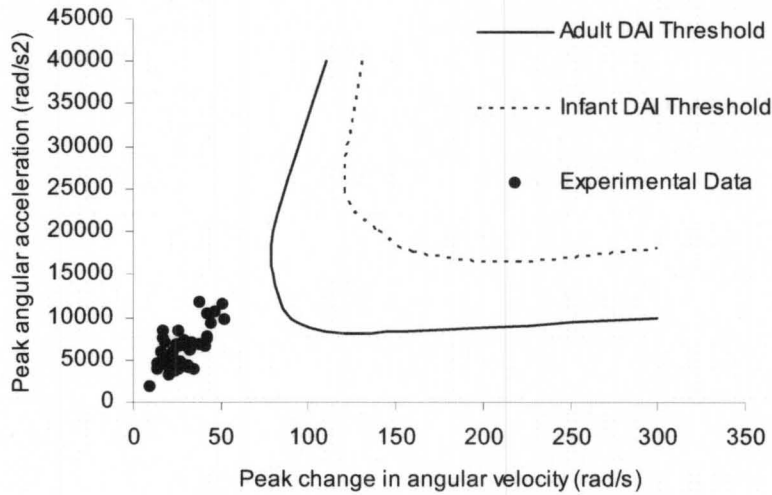


Figure 3-5. Peak medial-lateral angular head accelerations and peak change in angular velocity: experimental data compared to diffuse axonal injury (DAI) thresholds⁷⁸.

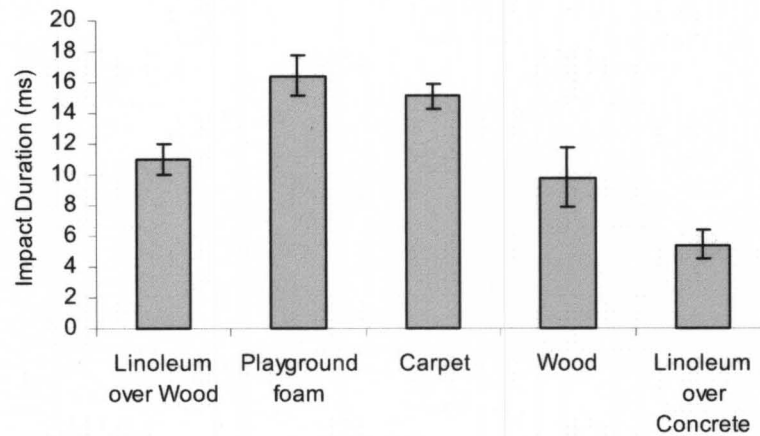


Figure 3-6. Head impact durations for falls onto various surfaces. Error bars represent 95% confidence intervals.

Neck Injury Outcome Measures

The mean peak neck axial compression force across all trials was 779 N (Figure 3-7). The greatest bending moments occurred in the lateral direction with a mean of 13.4 Nm (Figure 3-8). Falls onto linoleum over concrete produced the greatest neck loads in all measured directions. Peak neck loads were as follows: axial compression – 1,504 N, flexion – 17.9 Nm, extension – 4.2 Nm, lateral bending – 19.2 Nm, torsion – 4.1 Nm. Linoleum over concrete was associated with significantly greater peak neck compression forces than linoleum over wood ($p = 0.006$), playground foam ($p < 0.001$), and carpet ($p = 0.017$). Linoleum over concrete was associated with significantly greater peak neck flexion moments than carpet ($p = 0.019$). Linoleum over concrete was associated with significantly greater peak neck extension moments than wood ($p = 0.003$) and carpet ($p = 0.002$). No significant differences in lateral bending moments were found between the

tested surfaces. Linoleum over concrete was associated with significantly greater peak neck torsion moments than linoleum ($p = 0.020$) and carpet ($p = 0.002$). Playground foam was associated with significantly greater peak neck torsion moments carpet ($p = 0.031$).

N_{ij} calculations were performed to evaluate combined loading mechanisms in the sagittal plane. The greatest N_{ij} values occurred for the compression-flexion loading mechanism (N_{CF}) (Figure 3-9). The mean N_{CF} value across all surfaces was 0.7. Five of the nine falls onto linoleum over concrete produced N_{CF} values greater than one (maximum 1.3), and one fall onto the linoleum over wood surface produced an N_{CF} equal to 1.0.

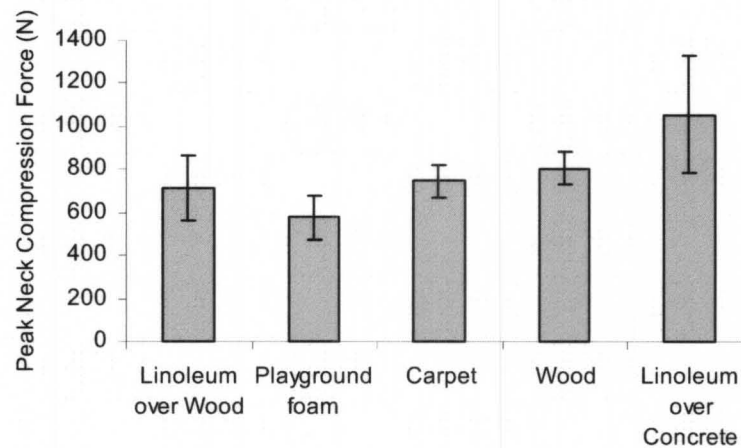


Figure 3-7. Peak neck compression force for falls onto various surfaces. Error bars represent 95% confidence intervals.

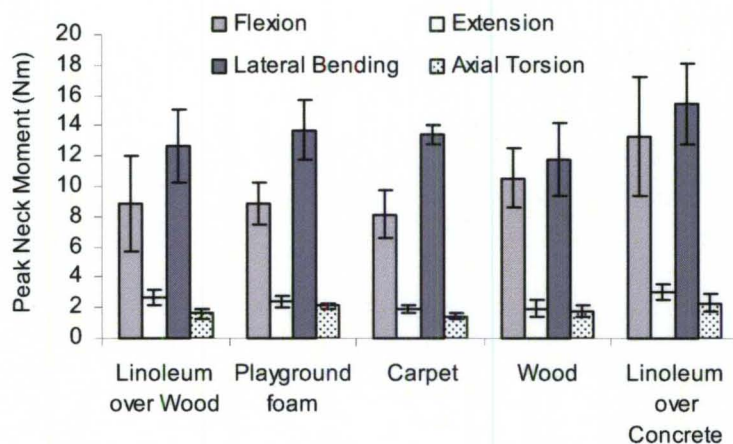


Figure 3-8. Peak neck moments applied in various directions (flexion, extension, lateral bending, torsion) for falls onto various surfaces. Error bars represent 95% confidence intervals.

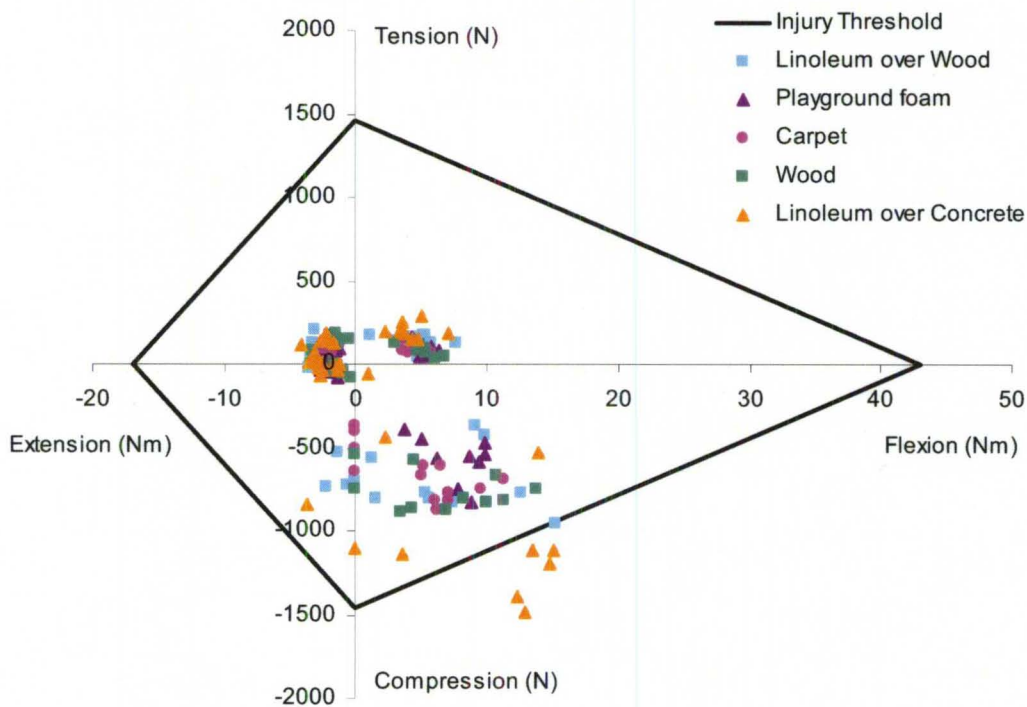


Figure 3-9. N_{ij} neck injury criteria 12-month-old CRABI threshold and results for falls onto various surfaces.

Extremity Injury Outcome Measures

Mean peak compression forces were much greater in the arms than in the legs (Figure 3-10). The greatest compression forces occurred in the left arm and in falls onto linoleum over wood (maximum of 6,712 N). Unlike head and neck outcome measures, surface trends were less evident in extremity loads. The only significant differences in compression forces across surfaces occurred for the left leg. Left leg compression forces in falls onto linoleum over concrete were significantly greater than those in falls onto linoleum over wood ($p = 0.040$), wood ($p = 0.047$), and carpet ($p = 0.013$).

Mean peak bending and torsion moments were also highest in the left arm (Figures 3-11 and 3-12). The maximum bending moment across all trials was 26.1 Nm and occurred in a fall onto linoleum over concrete. The maximum torsion moment was 23.6 Nm and occurred in a fall onto linoleum over wood. Linoleum over concrete was associated with significantly lower right leg peak bending moments than wood ($p = 0.020$), playground foam ($p = 0.012$), and carpet ($p = 0.040$). Linoleum over concrete was associated with significantly greater left leg peak bending moments than linoleum over wood ($p = 0.016$) and carpet ($p = 0.001$). Linoleum over concrete was associated with significantly greater left arm peak bending moments than playground foam ($p = 0.045$) and carpet ($p = 0.022$).

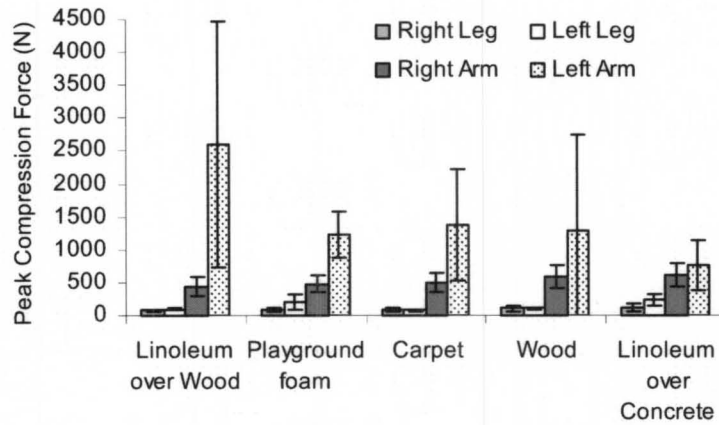


Figure 3-10. Peak axial compression force for each extremity and for falls onto various surfaces. Error bars represent 95% confidence intervals.

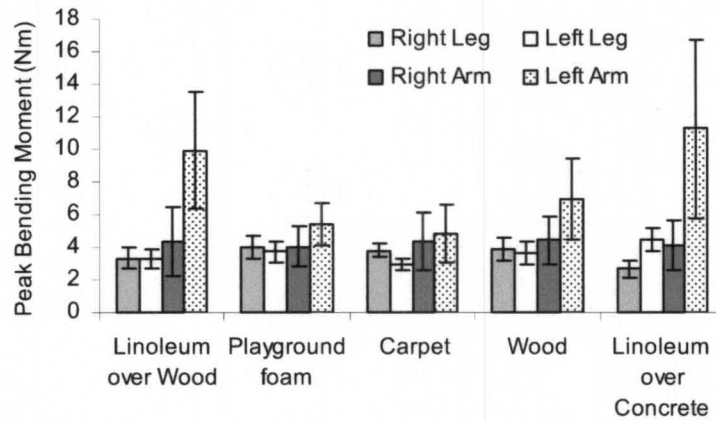


Figure 3-11. Peak bending moment for each extremity and for falls onto various surfaces. Error bars represent 95% confidence intervals.

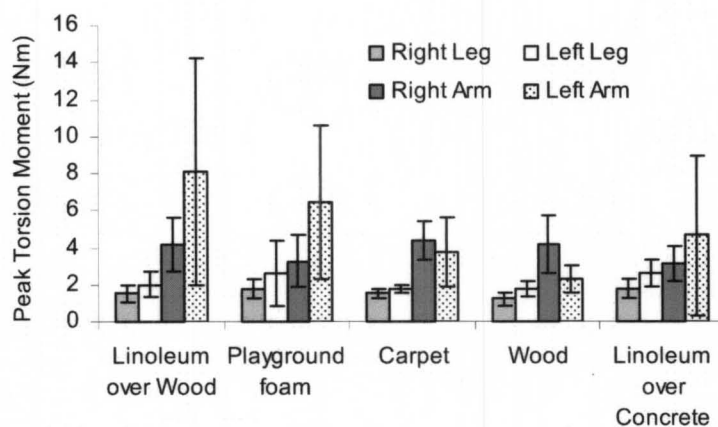


Figure 3-12. Peak torsional moment for each extremity and for falls onto various surfaces. Error bars represent 95% confidence intervals.

Discussion

Effect of Surface

Significant differences in outcome measures were found across the evaluated surfaces. Linoleum over concrete was associated with significantly greater linear and angular head accelerations, HIC_{15} values, and shorter impact durations than all other surfaces. Since greater accelerations and shorter impact durations are generally associated with an increased risk of head injury, the greatest head injury risk in short-distance horizontal falls would be for linoleum over concrete or similar surfaces. Additionally, wood and linoleum over wood are associated with a greater risk of head injury than carpet or playground foam surfaces. Similarly, linoleum over concrete was associated greater neck forces and moments (and thus a greater risk of neck injury) in

these falls. Few differences in extremity loads were found across the various impact surfaces. This is likely due to the high variation in these measures. Nine trials per scenario were conducted based upon a power analysis of head injury outcome measures in previous fall experiments using the CRABI ATD. The results of these experiments, however, suggest that a greater number of trials would be necessary for investigation of extremity loads. In future studies, additional fall trials should be conducted to further elicit differences across various impact surfaces.

Head Injury Potential

To determine the potential for head injury in these falls, the results can be compared to published injury thresholds. Head injury thresholds can be separated into two types: those based on linear acceleration (which generally predict the potential for focal or contact-type head injuries) and those based on angular or rotational acceleration (which generally predict the potential for inertial or diffuse brain injury). In this study HIC values were determined. The HIC was developed for use in the automotive industry to assess head injury risk in motor vehicle crash testing, and today is the most widely accepted measure of head injury risk in impacts. HIC have also been used in several studies to assess head injury risk in falls^{67, 70-71, 93, 107}. The proposed HIC₁₅ limit for the CRABI 12-month-old ATD is 390⁷⁶. For a HIC₁₅ of 390, the risk of skull fracture is approximately 31%⁷⁶. The maximum HIC₁₅ in this study was 334 which occurred in a fall onto linoleum over concrete. All other surfaces were associated with HIC₁₅ values less than or equal to 200. This suggests a risk of skull fracture less than 31% in falls from

the studied height onto linoleum over concrete, and a very low risk of skull fracture for falls onto other surfaces.

A large range of injury thresholds based on the peak linear acceleration have been proposed for children. Sturtz⁷⁷ proposed tolerance limits of 83g (6-7 year-old children) for impact durations greater than or equal to 3 ms based on reconstructions of pedestrian accidents. Above this load Abbreviated Injury Scale (AIS) level 2+ injuries are possible. By using computer simulations to reconstruct free falls resulting in serious head injuries, Mohan et al.⁷¹ proposed conservative tolerance limits of 200 – 250g peak accelerations for children. Others have reported tolerance limits for children ranging from 50 – 200g where 50g is the maximum before-injury threshold and 200g is the threshold for fatal injury⁷⁵. Peak linear accelerations fell at or below 200g for all surfaces except linoleum over concrete. Linoleum over concrete produced a maximum linear head acceleration of 423g. There is such disagreement in the thresholds, however, that the risk of head injury in these falls is difficult to assess using linear acceleration alone. Additionally, peak g thresholds do not account for the duration of impact. Longer impact durations generally increase the injury risk. Although linoleum over concrete was associated with the greatest peak linear accelerations, these falls also produced the shortest impact durations (mean duration 5.4 ms).

As with linear head acceleration, many angular acceleration tolerance limits for head injury have been proposed and are often specific to injury type. Additionally, the direction of head motion is important, as some thresholds differ depending on the direction of the load. The brain is more susceptible to diffuse axonal injury (DAI) under lateral rotation than anterior-posterior rotation⁷⁹. However, subdural hematomas are

more likely to result from rotation in the sagittal plane (anterior-posterior). For this reason, both anterior-posterior and medial-lateral angular accelerations were measured in this study. Reported concussion thresholds are approximately 6,500 rad/s² for a young child (800 gm brain mass) and 10,000 rad/s² for an infant (400 gm brain mass)¹⁸. Similarly, accelerations necessary to cause mild diffuse axonal injury (DAI) have been reported as approximately 18,000 rad/s² for a young child and 30,000 rad/s² for an infant. Margulies and Thibault⁷⁸ established tolerance curves for moderate DAI based on peak angular acceleration and peak change in angular velocities (Figure 3-5). These curves were derived from a combination of animal experiments, physical models, and analytical model simulations. Duhaime et al.⁶³ used a tolerance limit of approximately 35,000 rad/s² and 40,000 rad/s² for subdural hematoma (SDH) and DAI, respectively, in an infant with a 500 gram brain mass. Depreitere et al.¹⁰⁸ proposed a SDH tolerance level of approximately 10,000 rad/s² for impact durations less than 10 ms based on adult cadaver impact tests. In our study, ML angular accelerations were generally greater than AP angular accelerations (because the ATD landed on the side of its head). The maximum ML angular acceleration across all tested surfaces was 11,730 rad/s² (occurred in a fall onto linoleum over concrete). Falls onto surfaces other than linoleum over concrete produced ML angular head accelerations less than 7,400 rad/s². In comparing our results to proposed thresholds, DAI would not be expected in these falls as all data fell below proposed pediatric thresholds. However, concussion is possible, particularly for falls onto linoleum over concrete where several trials exceeded 10,000 rad/s². The maximum AP angular acceleration was 9,322 rad/s² (occurred in a fall onto linoleum over concrete). AP angular accelerations were below 5,000 for all tested surfaces except

linoleum over concrete. As these results fall below proposed SDH thresholds, the risk of SDH in these falls is low.

Neck Injury Potential

Neck injury has been studied much less than head injury, particularly in infants, and thus there are fewer published pediatric neck injury thresholds. One of the most commonly used neck injury assessment thresholds is the N_{ij} criteria. The N_{ij} criteria, like the HIC, were developed for use in the automotive industry to assess injury risk in frontal impact motor vehicle crash testing. In this study, compression-flexion was the primary loading mechanism (of the four included in N_{ij}). Several falls onto linoleum over concrete exceeded the threshold and one fall onto linoleum over wood met the N_{ij} threshold of 1.0. An $N_{ij} = 1$ represents a 22% probability of AIS 3 (serious) neck injury, suggesting that serious neck injuries are possible in these falls⁷⁶. These results are particularly concerning since N_{ij} is only calculated for sagittal motion and the primary loading direction in our experiments was in the coronal plane. No published injury thresholds were found for lateral bending and torsional neck loading.

Extremity Injury Potential

In general, peak loads in the upper extremities were greater than those in the lower extremities, and peak loads on the impact (left) side of the body were greater than those on the non-impact (right) side. The fall dynamics were such that the ATD initially

landed on the left side of the body, causing left upper extremity and left lower extremity forces and moments to be greater than those on the right side of the body. Additionally, the ATD's left shoulder impacted the ground approximately the same time or just after head impact (before the remainder of the body) leading to substantially greater loads in the left upper extremity compared to the other extremities. Upper extremity loads tended to be greater than lower extremity loads, possibly due to the larger mass of the lower extremities and thus more soft tissue. The metal rods representing the humeri and femurs were the same diameter (0.25 in), but the overall lower and upper extremity diameters (including the soft tissue material) were approximately 2.5 in and 1.5 in, respectively. Greater thickness of soft tissue in the lower extremity combined with an increased air cavity between the "bone" and soft tissue would allow for more cushioning and subsequently reduce the peak loads experienced by the lower extremity as compared to the upper extremity.

Adult bone strength has been well studied, and femur and humerus fracture thresholds are shown in Table 3-2⁸⁰. Femur and tibia injury criteria for adult ATDs have been established to assess injury risk in automotive crash testing. Femur compression thresholds for the adult Hybrid III 50th percentile (male) and 5th percentile (female) ATDs are 10 kN and 6.8 kN, respectively⁷⁶. Proposed tibia compression thresholds for the adult Hybrid III 50th and 5th percentile ATDs are 35.9 kN and 22.9 kN, respectively. Proposed tibia bending moment thresholds for the adult Hybrid III 50th and 5th percentile ATDs are 225 Nm and 115 Nm, respectively⁷⁶. Little information is available on pediatric bone strength. A few studies have investigated pediatric femur strength, but no known studies have investigated pediatric humeral strength⁸¹⁻⁸⁴. Using data from quasi-

static bending and compression tests of pediatric femur specimens, Sturtz⁷⁷ estimated the dynamic loads necessary to produce a fracture. This calculation was based on the assumption that dynamic fracture thresholds are 20% higher than quasi-static thresholds. The dynamic bending fracture criteria for a 7 year old and 3.6 year old child were 116-131 Nm and 62-73 Nm, respectively. Also the dynamic axial (compression) fracture criteria were 1800 and 1000 N for a 6 year old and 3 year old, respectively.

Load Mechanism	Femur Thresholds		Humerus Thresholds	
	Male	Female	Male	Female
Compression (kN)	7.72	7.11	4.98	3.61
Bending Moment (Nm)	310	180	151	85
Torque (Nm)	175	136	70	55

Table 3-2. Fracture thresholds for the adult femur and humerus bones⁸⁰.

The peak femur compression force, bending moment, and torque across all trials were 647 N, 6.8 Nm, and 8.5 Nm, respectively. These values fall well below femur fracture thresholds for the adult and the pediatric thresholds proposed by Sturtz. Therefore, a low risk of femur fracture is associated with the tested fall scenario. Peak humerus compression force, bending moment, and torque across all trials were 6712 N, 26.1 Nm, and 23.6 Nm, respectively. Humerus bending moments and torques are below adult injury thresholds. However, the maximum humerus compression load exceeds fracture thresholds for the adult (Table 3-2). As humerus fracture thresholds for the child would likely fall below fracture thresholds for adults, this suggests a risk of humerus fracture due to compressive loading in these falls.

Comparison to other Biomechanical Studies

Several studies have investigated injury potential in pediatric falls^{64-68, 93, 109}. Bertocci et al.⁶⁷ simulated bed falls from a 0.68 m high horizontal surface using a Hybrid II 3-year-old ATD. Although a similar initial position was used by Bertocci et al. as compared to our study, the legs or pelvis of the 3-year-old ATD made first contact with the ground rather than the head. Peak head accelerations and HIC15 values were comparable to those measured in our study. Angular head accelerations were not measured. Femur compression and bending loads measured by Bertocci were comparable to those measured in our study. However, torsional loads measured by Bertocci were up to ten times the values measured in this study. This is likely due to the feet-first impact orientation seen in those falls.

Ibrahim and Margulies⁶⁶ simulated falls using an 18-month-old surrogate. The surrogate was dropped from various heights (1-3 ft) onto carpet pad and concrete. The surrogate was initially suspended above the floor in a supine position with the head slightly below the rest of the body (so that the head would impact the ground first). This differs from our study which simulated the entire fall event (rolling off the bed). Peak angular accelerations for the primary head loading direction (medial-lateral rotation in our study versus anterior-posterior rotation in the Ibrahim study due to different impact orientations) were compared. Peak angular accelerations reported by Ibrahim and Margulies were more than double those measured in our study. This is likely due to differing skull and neck properties of the surrogates. In particular, the CRABI neck is

stiffer than Ibrahim's surrogate model (approximately 0.115 Nm/degree versus 0.0637 Nm/degree in flexion and lateral bending).

Comparison to Clinical Studies

The results of this study are consistent with epidemiological studies of pediatric falls. Two studies of bed falls found no serious head injuries in a combined 512 cases^{22, 57}. There were four skull fractures reported in these studies, but all were of a non-serious nature. Additionally, one humeral fracture and three clavicle fractures were reported by Helfer et al. A study by Tarantino et al.⁶¹ investigated injuries resulting from short falls (less than 4 feet) in infants less than 10 months of age. Of 167 subjects, 85% had minor or no injury and 15% had significant injuries. Significant injuries included seven long bone fractures (three femur, one humerus, two tibia, and one clavicle), and 18 closed head injuries. Two patients had intracranial hemorrhages but were later determined to be victims of abuse. Hennrikus et al.⁵⁹ found 115 patients with orthopedic injuries resulting from bed falls or falls from other furniture surfaces over a 20-month period. The injuries included fractures and dislocations primarily of the upper extremities. A previous study of fall cases (Chapter II), which reported injuries in 79 clinical cases of household falls, found 6 skull fractures, 9 upper extremity fractures, and 2 lower extremity fractures. This study also reported 2 small isolated SDH. One of the falls that produced a SDH involved a 1-month-old rolling of an 83 cm high bed. However, this child also hit his head on the edge of a humidifier placed next to the bed. The second case occurred when a 42-month-old child fell rearward from the back of a couch. In both cases, the children recovered

fully. The results of previous studies are consistent with our study which found a moderate (less than 30%) risk of skull fracture, and a very low risk for more severe head injuries (such as SDH). A moderate risk of humerus fracture was found in our study, which is consistent with other studies that report injuries to the upper extremities commonly resulting from short-distance falls. However, the risk of femur fracture in this study was very low.

Our study also found a small potential for neck injuries in bed falls. However, neck injuries have rarely been reported in short falls. Chiaviello et al.²⁴ reported that 1 of 69 children who fell down stairs sustained a C2 vertebral fracture. To the authors' knowledge, no neck injuries have been reported from bed falls or other short-distance furniture falls. The neck loads reported in this study should be interpreted with caution as the CRABI neck is stiffer than an actual 12-month-old child's neck. Additionally, the CRABI neck was designed to investigate injury risk in frontal impact motor vehicle crash tests. Therefore, neck response in lateral bending or axial compression (the two primary loading mechanisms in the simulated falls) were not of interest for biofidelity requirements in ATD design.

Limitations

This study has several limitations. First, the biofidelity of the CRABI 12-month-old ATD has been questioned. As previously mentioned, the CRABI neck is likely too stiff. A more flexible neck would allow for increased head rotation on impact. Therefore, the head accelerations reported in this study (particularly angular

accelerations) may be underestimated. Conversely, increased neck flexibility would likely decrease neck forces and moments. This suggests that the neck loads reported in this study may be overestimated compared to those experienced by a 12-month-old child. The biofidelity of the CRABI head impact response has similarly been questioned^{66,91}. One study compared the head impact response of a CRABI 6-month-old ATD to that of pediatric cadaveric specimens in drop tests and found the results to be comparable in vertex, occiput, and forehead impacts¹¹⁰. However, the impact response of the CRABI in lateral impacts was much stiffer than that of the cadaveric specimens. Therefore, the peak linear head accelerations and HIC values reported in this study may be overestimated compared to what would be experienced by a 12-month-old child. As with the head and neck, the CRABI soft tissue is stiffer than that of a human child. A previous clinical study of household falls (Chapter II) found significant differences in child body mass index (BMI) between children with minor or more severe injuries. This suggests that soft tissue may have a protective or cushioning effect. In addition to questions of head, neck, and soft tissue biofidelity, ATD joints (shoulders, elbows, hips, and knees) are limited to motion in the sagittal plane. As the impact orientation in the simulated falls occurred primarily in the coronal plane, the joint constraints may have affected the fall/impact dynamics and thus the resulting injury outcome measures. Of particular interest are constraints of the left shoulder. With shoulder motion constrained to a single pin joint in the sagittal plane, additional loads may have been transferred to the left arm which may have otherwise been absorbed through shoulder motion in other directions. Therefore, the upper extremity loads in this study may be overestimated.

In addition to limitations of the ATD, assessments of injury risk are based on injury criteria that have primarily been determined through scaling of adult or primate data. This is due to the paucity of information concerning material properties of pediatric tissues and pediatric injury tolerance. Scaling generally accounts for mass differences, but in some cases (the HIC for example) may account for differences in geometry and material properties. Angular head acceleration thresholds for pediatric brain injury were determined through mass scaling alone. However, Thibault and Margulies⁸⁸ found that including differences in brain tissue material properties reduced thresholds for concussion, DAI, and SDH. More accurate pediatric injury criteria are needed to improve assessments of injury potential in falls.

It should be noted that only one initial position was simulated in these falls (side-lying to simulate a rolling motion from the bed surface). Changing initial positions would likely change the orientation of the ATD upon impact, leading to differences in the injury outcome measures. Additionally, the rate at which the ATD was pushed from the bed surface was held constant. Changes to the initial velocity of the ATD or the push force would likely affect the fall dynamics and injury outcome measures. Any significant deviation from the simulated scenario (a 12-month-old child rolling off the bed) would require further investigation to more accurately assess injury potential.

Conclusions

This study investigated biomechanical outcomes relating to injury potential in falls from beds and other horizontal surfaces using an ATD representing a 12-month-old

child. The potential for head, neck, and extremity injuries was determined. Differences in injury outcome measures due to varying impact surfaces were also investigated. The risk of severe head and extremity injuries in these falls was low. However, fractures, particularly involving the skull and humerus, are possible in these falls. Neck injury potential in pediatric falls should be studied further as limitations in ATD biofidelity and neck injury thresholds based solely on sagittal plane motion may reduce accuracy in current pediatric neck injury assessments. Linoleum over concrete was associated with the greatest risk of head and neck injury compared to other evaluated surfaces (linoleum over wood, carpet, wood, playground foam). These results may aid clinicians in distinguishing between abusive and accidental injuries when the stated cause of the injuries is a short-distance household fall and further highlight the importance of obtaining a detailed history when assessing compatibility between injury and the stated cause.

CHAPTER IV
PEDIATRIC BED FALL COMPUTER SIMULATION MODEL PART I:
DEVELOPMENT AND VALIDATION

Overview

Falls from beds and other household furniture are common scenarios stated to conceal child abuse. Knowledge of the biomechanics associated with short-distance falls may aid clinicians in distinguishing between abusive and accidental injuries. Computer simulation is a useful tool to investigate injury-producing events, and to study the effect of altering event parameters on injury risk. In this study, a pediatric bed fall computer simulation model was developed and validated. The simulation was created within MADYMO® software using the CRABI 12-month-old anthropomorphic test device (ATD) to represent the fall victim and validated using data from physical fall experiments of the same scenario with an instrumented CRABI ATD. Validation was conducted using both observational and statistical comparisons. Future parametric sensitivity studies using this model will lead to an improved understanding of relationships between child (fall victim) parameters, fall environment parameters, and injury potential.

Introduction

Falls from beds and other household furniture are common scenarios stated to conceal child abuse.⁴⁻⁸ A better understanding of the true injury risk associated with these falls is needed to aid clinicians in distinguishing between abusive and accidental injuries. Fall environment factors, such as fall height and impact surface, as well as child factors, such as body mass index, have been shown in previous studies to be related to injury risk in short falls.^{64-68, 93, 109} However, many of these studies have been limited by the biofidelity of anthropomorphic surrogates used to represent the fall victim.^{64-68, 93} Moreover, little information is available regarding the injury tolerance and biomechanical response of children. Therefore, most pediatric surrogates are based on scaled adult cadaver or primate data and may not accurately represent a human child, particularly in low-energy events such as falls.

Computer simulation is a useful tool that can be used to investigate injury-producing events, and to study the effect of changing event parameters on injury risk. Computer simulation has been widely used by the automotive industry to study motor vehicle crash events, and has also been used in a few studies to investigate falls.^{72, 111-115} Development of a pediatric bed fall computer model can lead to a deeper understanding of relationships between biomechanical factors, fall environment parameters, child parameters and potential for injury. Additionally, a computer model can extend beyond surrogate experiments by allowing the user to vary surrogate properties. The purpose of this study was to develop a validated 3D computer model simulating an anthropomorphic test device (ATD) representing a 12-month-old child falling from a horizontal surface

such as a bed. In Chapter II, rolling off of a bed or other horizontal surface was found to be the most common short-distance fall scenario in infants and toddlers. Therefore, in this study, a computer simulation model was developed to recreate the “rolling off the bed” scenario. This model will later be used to investigate the effect of changing fall environment and ATD (fall victim) parameters on biomechanical measures and potential for injury (Chapter V).

Methods

A computer simulation model of a pediatric bed fall was developed using MADYMO® version 7.0 (MATHematical DYnamic Modeling; TNO, Netherlands). MADYMO® is a rigid-body dynamics software. One advantage of MADYMO® is that it contains a built-in database of models representing the anthropomorphic test devices (ATD). For this study, the Child Restraint Air-Bag Interaction (CRABI) 12-month-old anthropomorphic test device (ATD) was selected to represent the fall victim. This ATD represents a 50th percentile 12-month-old child in terms of overall height (74 cm) and mass (10 kg), as well as geometric and inertial properties of individual body segments. The model was validated using results from physical bed fall experiments with an instrumented CRABI 12-month-old ATD (Chapter III). Once validated, the predictive capability of the model was assessed by changing the impact surface type and comparing the outcome measures with experimental results.

ATD Fall Experiments

Bed fall experiments were performed using the CRABI ATD (First Technology Safety Systems, Plymouth, MI). The ATD was placed in a side-lying position on a horizontal surface representing a bed (Figure 4-1). The bed was 61 cm (24 in) above the ground. Before each fall, ATD joint angles were adjusted using a goniometer to ensure repeated positioning for all testing. Joints were calibrated to manufacturer specifications whereby the joint was tightened until the friction was just sufficient to support the weight of the limb. A pneumatic actuator was mounted to the horizontal surface representing the bed and used to push the ATD off the edge of the bed (Figure 4-1). Nine falls were conducted onto two different impact surfaces (playground foam and linoleum) for a total of 18 falls. The playground foam surface consisted of rubber tiles 61 x 61 cm, 5.1 cm thick. The linoleum surface was self-adhesive vinyl flooring 0.1 cm thick. The linoleum was adhered to a wood subfloor (1.5 cm thick plywood), while the playground foam was placed over concrete.

The ATD was instrumented with tri-axial accelerometers (Endevco, Model 7264-2000) at the center of mass of the head, overall ATD center of mass located at the midline within the torso, and the pelvis. Two angular rate sensors (ATA Sensors, Model ARS-06) were also positioned in the head to measure head angular velocity in the anterior-posterior and medial-lateral directions. Additionally, a six-axis load cell (First Technology Safety Systems, Model IF-954) was located at the superior aspect of the neck (approximately the C1 vertebrae location). Accelerometer and load cell data were sampled at 10,000 Hz and filtered according to the SAE J211 standards.¹⁰⁵ Data were

filtered using a 4th order low-pass Butterworth filter with cutoff frequencies of 1000 Hz (accelerations, angular velocities, and neck forces) and 600 Hz (neck moments).

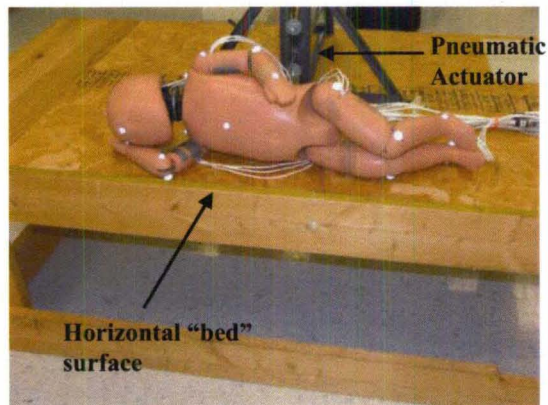


Figure 4-1. CRABI anthropomorphic test device (ATD) in side-lying initial position for bed fall experiments. The pneumatic actuator (used to deliver a force to the posterior torso of the ATD to push it from the surface) is shown behind the ATD.

Each fall experiment was videotaped and captured using a three-dimensional digital motion capture system (Motion Analysis Co., Santa Rosa, CA) to record fall dynamics. This system uses five infrared cameras at a 100 Hz frame rate. Forty-eight reflective markers (4-5 per body segment) were placed on the ATD, and one marker was placed on the actuator to determine actuator kinematics.

Model Development

Fall Environment

The fall environment used in the ATD experiments was recreated in the computer simulation model using rigid body planes and ellipsoids to represent the bed surface and impact surface. Appropriate geometry and surface properties were specified in the model. Initially, the model was created using properties of playground foam as the impact surface. A rigid ellipsoid was created to represent the actuator. The velocity and acceleration of the actuator were specified to match that measured in the experiments.

ATD Properties

The 12-month-old CRABI ATD ellipsoid model from the MADYMO® database was imported into the model and positioned on the bed surface as in the experiments. The CRABI model consists of 32 bodies and was created with geometric and inertial properties to match the physical ATD. The CRABI model, developed by TNO-MADYMO®, was created by scaling down the Hybrid III 50th percentile adult ATD model. Anthropometric measurements on the physical ATD were also included in model development by TNO-MADYMO®. The Hybrid III adult ATD model within the MADYMO® database was previously validated through both component tests and full-body sled impact tests¹¹⁶. However, no specific validation was performed by TNO-MADYMO® for the CRABI model after scaling. Due to the lack of validation of the CRABI model and the poor performance of the ATD model (in comparing model outcome measures with results from ATD fall experiments) without any modifications,

head and neck properties were measured through component testing of the physical ATD to improve the CRABI model. Additionally, segment masses in the original CRABI model differed from those of the physical ATD and were updated accordingly. Head contact properties used in our model were determined using an experimental head drop test as a part of our study. In this test, the instrumented head of the ATD was dropped from a height of 61 cm (same fall height used in experiments with the full ATD) onto a concrete surface. The head was positioned so that the impact orientation was similar to that found in the ATD fall experiments and the model (impacting on the left parietal aspect of the head). Three trials were conducted. A computer model of the head drop test was created using MADYMO®, and head stiffness properties were adjusted until the resultant head acceleration from the head drop model matched those in the experiments. The resulting load-deformation curve for the head (Figure 4-2) was then imported into our bed fall model.

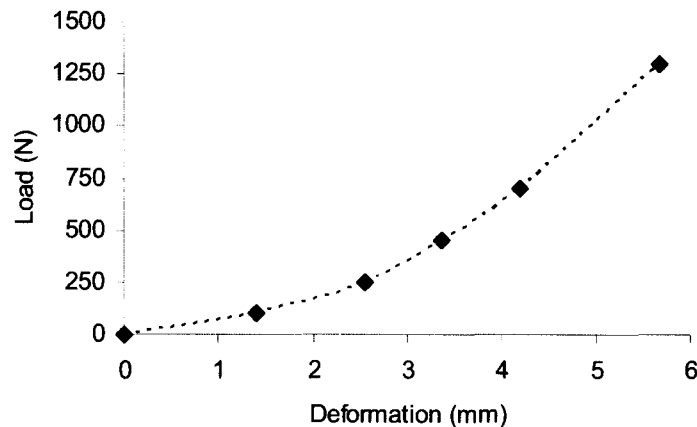


Figure 4-2. Load-deformation characteristic for CRABI head used in our computer model based upon head drop experiments.

Neck stiffness was determined using static testing whereby the neck was adjusted to a known angle and the bending moment was recorded using a load cell positioned at the superior aspect of the neck. The base of the neck was fixed and rotation angles were recorded using a goniometer (positioned at the center of the superior aspect of the neck). Stiffness was determined for flexion, extension, lateral bending, and torsion (Figure 4-3). The MADYMO® CRABI ATD model includes two spherical joints (three rotational degrees of freedom) at the superior and inferior aspects of the neck. Due to the head-first nature of the fall, an additional translation joint was added to the neck in our computer model to allow for neck compression.

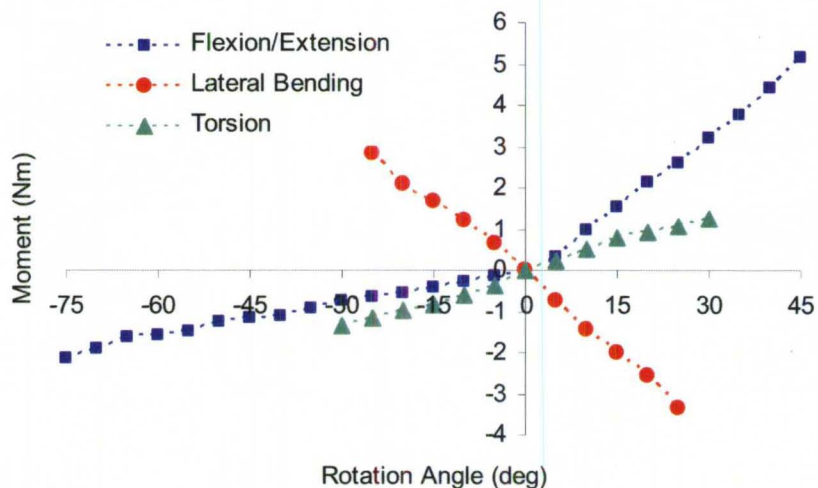


Figure 4-3. Moment vs. rotational displacement characteristics for CRABI neck used in our model based upon experimental evaluation.

Impact Surface Properties

In order to determine contact properties between the evaluated impact surfaces and the ATD, additional head drop experiments were performed. Head drop tests were

used because the ATD impacted the ground head-first in each of the experimental trials. Three head drop tests were performed onto each of the impact surfaces (playground foam and linoleum). The stiffness and damping properties of the impact surface in the model were then adjusted until the resulting head acceleration time histories matched those from the physical head drop experiments. The resulting surface stiffness values were 206 N/mm for playground foam and 867 N/mm for linoleum. A constant damping coefficient was insufficient to describe the interaction between the head and impact surface. Therefore damping was specified as a nonlinear function of both the velocity and penetration (deformation of the surface upon contact). The resulting damping force was calculated using

$$F_d = c \cdot k \cdot x \cdot v \quad (1)$$

where c is the damping coefficient (0.15 for playground foam, 0.30 for linoleum), k is the combined contact stiffness for the head and impact surface, x is the penetration, and v is the velocity. The resulting damping characteristics for the two impact surfaces are shown in Figure 4-4. Note that damping properties were determined for the ATD head-impact surface interaction and do not necessarily represent properties of the surfaces alone. Stiffness and damping properties resulting from the head drop tests were imported into the bed fall model. Friction coefficients were set to 0.88 and 0.87 for playground foam and linoleum, respectively, as previously measured.⁹³

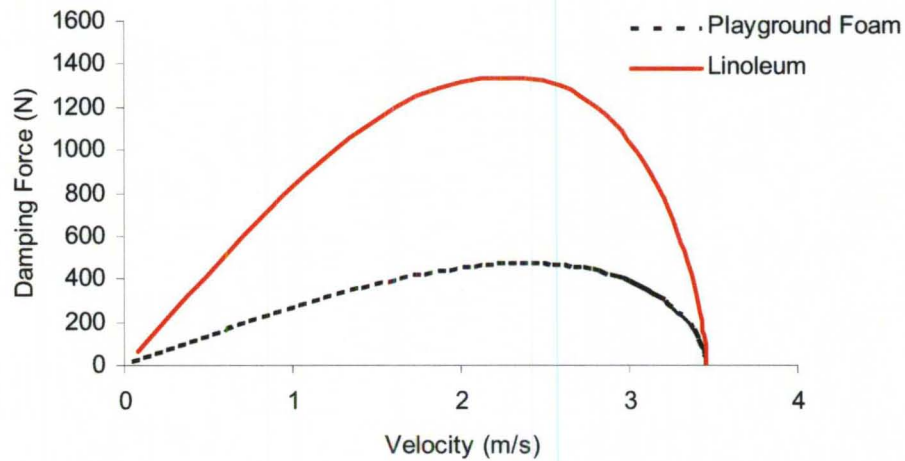


Figure 4-4. Damping force vs. velocity for head impact onto playground foam and linoleum surfaces.

Model Validation

The first step in the validation process was a visual comparison of the fall dynamics. The initial position of the ATD was adjusted until the ATD dynamics in the model matched those seen in the experiments. Next, outcome measures from the model were compared to those from the experiments. The measures selected for comparison were the head, torso, and pelvis resultant linear accelerations, head angular velocity in the anterior-posterior and medial-lateral directions, and upper neck resultant force and resultant moment. Only the primary impact event was investigated. For each outcome measure, the time-history curves from the nine experiments were used to create a min-max corridor. The model time-history curve was then overlaid onto this corridor to compare general curve profiles. The model was tuned until the curve profiles and peaks were similar. Parameters that were tuned include ATD position and orientation, stiffness

properties for body segments (other than the head) and joints, including neck stiffness and damping properties. Although neck bending stiffnesses were measured for the physical ATD, these were measured under static (rather than dynamic) loading conditions and were therefore used only as a starting point in the model.

The model outcome measures were statistically compared to the mean of the nine experimental trials using the playground foam surface. Four statistical tests were chosen to evaluate different aspects of the time-history comparison:

1. Mean value comparison – The mean value of the model over the time window of the primary impact was compared to that of the mean of the experimental trials. The percent difference between the two mean values was determined.
2. Peak value comparison – The peak value and time occurrence of the peak value (in relation to the start of the primary impact) were compared between the model and experimental mean. The percent difference in magnitude and the time difference between the two peak values were determined.
3. Relative error – The mean relative error, standard deviation of the relative error, and maximum relative error were computed to assess the error magnitude between the model and experimental mean over the entire duration of the primary impact.
4. Correlation coefficient – The extent of a linear relationship between the model and experimental time-history curves was determined over the time window of the primary impact.

For each statistical test, criteria for validation were determined based on the range of variation measured in the bed fall experiments. Each of the nine experimental trials was compared to the mean of the nine trials using the four tests described above. The maximum percent difference in mean value, maximum percent difference in peak value, maximum relative error, and minimum correlation coefficient for the nine trials were used as acceptance criteria for model validation. This was repeated for each of the seven outcome measures. Then the model was compared to the experimental mean using the same four statistical tests. If the results of the statistical tests between the model and the experimental mean were as good or better than the acceptance criteria, the model was considered valid. Statistical comparison was performed for the primary impact event only. This began at the moment of impact and ended when the signal leveled off near zero (change in signal magnitude beyond this end point was less than 1% of the peak value).

Assessment of Model Predictive Capability

Once the model was validated using playground foam impact surface properties, the surface contact properties were altered to represent the linoleum surface. Without making any additional changes to the model, the model outcomes were statistically compared to the mean of the experimental bed fall trials conducted onto linoleum using the four statistical tests described above. As with the validation tests performed for falls onto playground foam, the acceptance criteria for linoleum falls were determined by comparing each experimental trial to the mean of the experimental trials. If the results of

the statistical tests between the model and the experimental mean were as good or better than the results of the statistical comparisons between experimental trials and the experimental mean, the model was considered valid.

Results

The first step in model validation was to visually compare fall dynamics between the model and ATD experiments. Figure 4-5 shows a time sequence of one of the experimental falls onto the playground foam surface along with the corresponding sequence generated from the computer model. Fall dynamics were found to be comparable between the computer model and experiments. In both the model and experiments, the ATD was initially in a side-lying position with the right arm placed beneath the head. The ATD rolled off the horizontal “bed” surface and impacted the floor surface on the lateral aspect of the head first followed by shoulder contact with the surface.

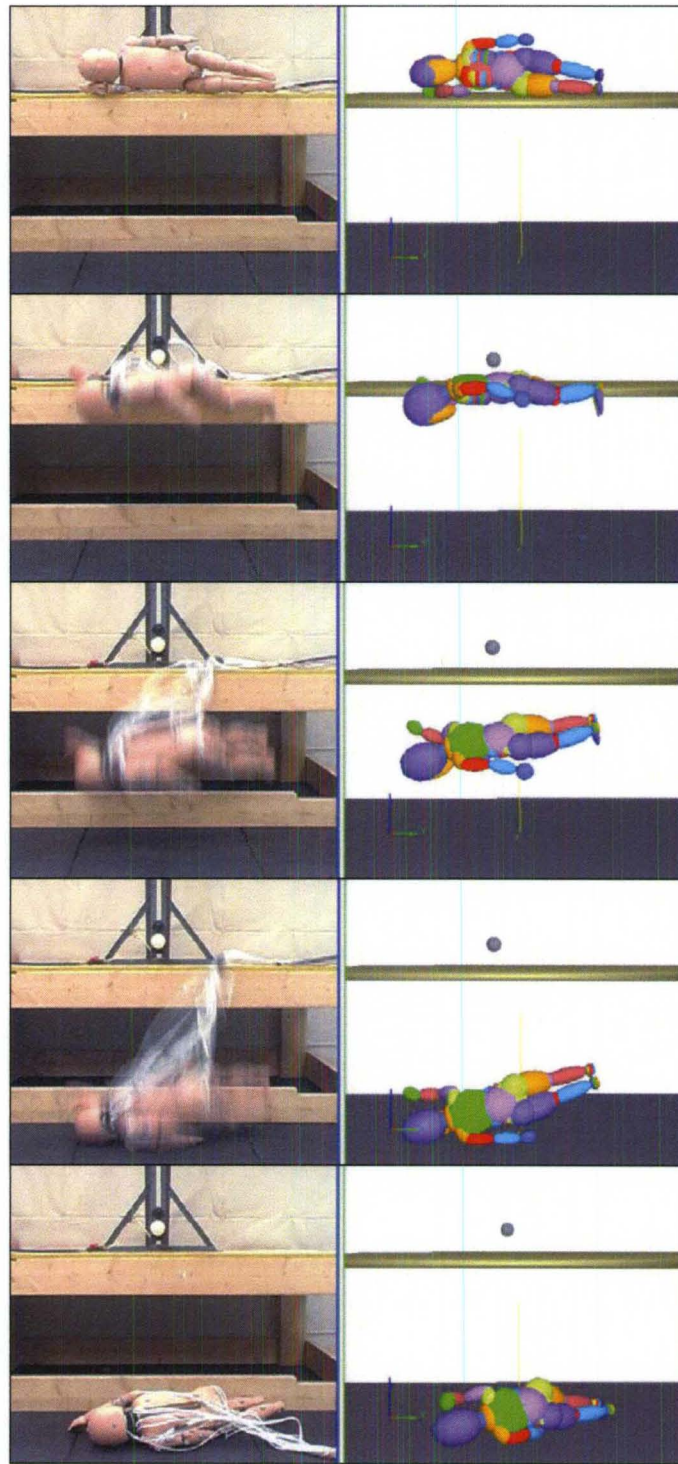


Figure 4-5. Time sequence comparison of ATD bed fall experiment and computer simulation model fall dynamics.

After visual comparison of the fall dynamics, outcome measures were compared both qualitatively and quantitatively between the simulation and experimental mean. Simulation model output, experimental mean, and experimental min-max corridor time histories of the seven outcome measures for falls onto the playground foam surface were compared (Figures 4-6 through 4-8).

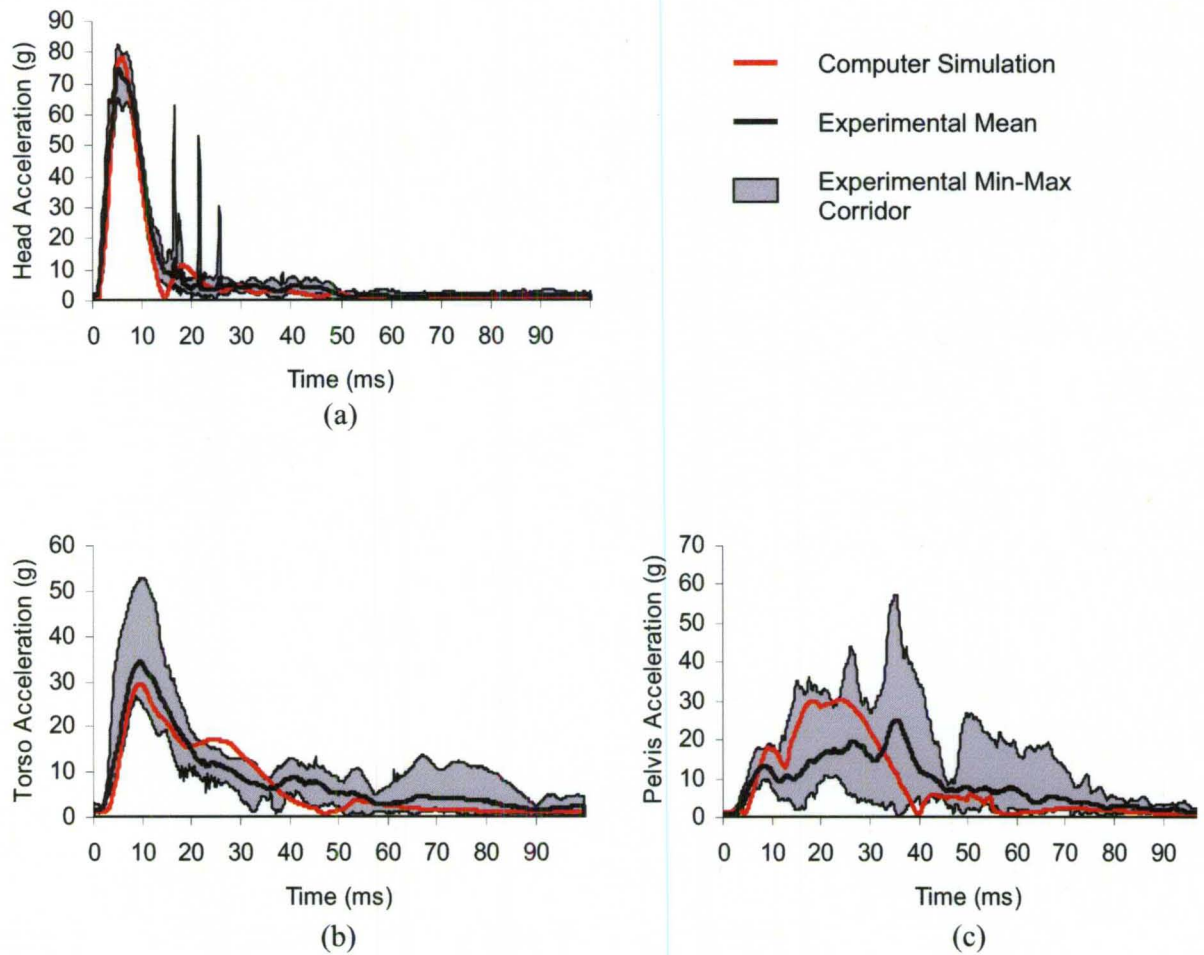


Figure 4-6. Computer simulation model and experimental time history comparisons for falls onto playground foam: (a) resultant linear head acceleration, (b) resultant linear torso acceleration, (c) resultant linear pelvis acceleration.

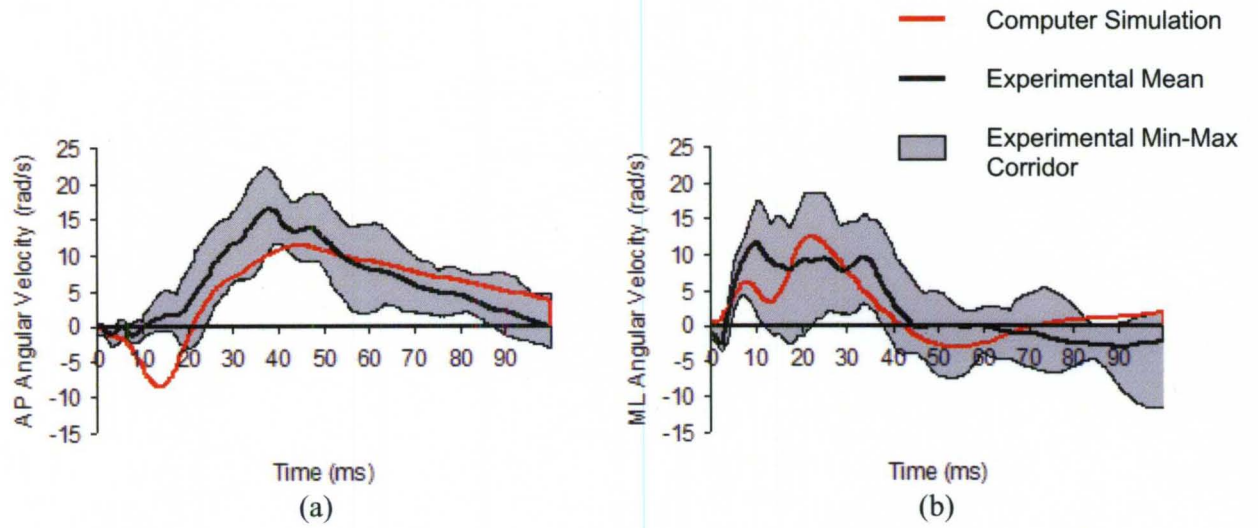


Figure 4-7. Computer simulation model and experimental time history comparisons for falls onto playground foam: (a) anterior-posterior (AP) angular head velocity, (b) medial-lateral (ML) angular head velocity.

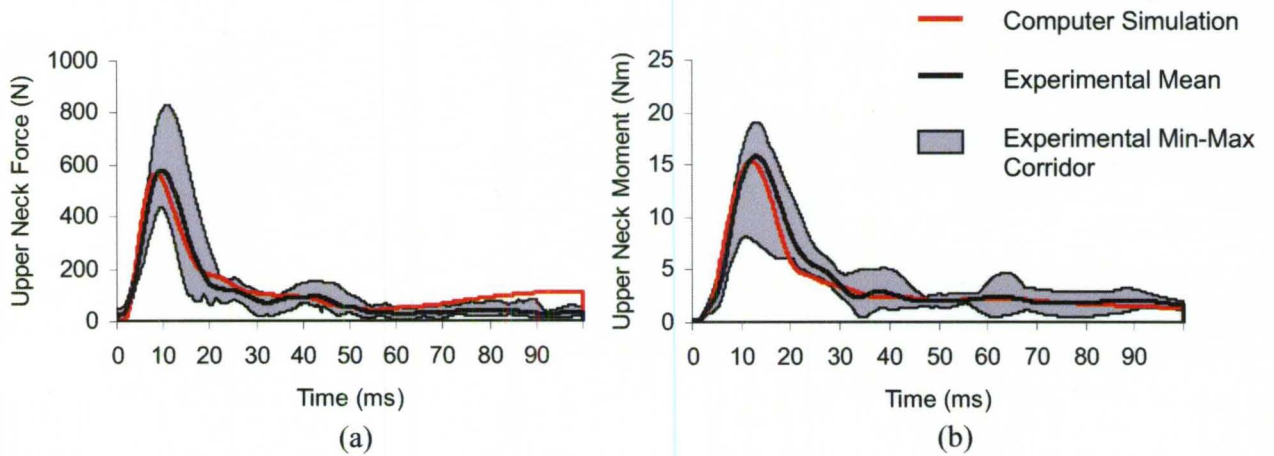


Figure 4-8. Computer simulation model and experimental time history comparisons for falls onto playground foam: (a) resultant upper neck force, (b) resultant upper neck moment.

Table 4-1 shows the results of the validation statistical tests comparing the model and experimental means along with acceptance criteria for falls onto playground foam. The model outcomes passed the acceptance criteria for each of the statistical tests.

Outcome Measure	Statistical Test and Acceptance Criteria*						Correlation Coefficient
	Mean Value	Peak Value		Relative Error			
	Difference (%)	Difference (%)	Time Difference (ms)	Mean (%)	Standard Deviation (%)	Maximum (%)	
Head Acceleration	8.8 (12.2)	4.9 (12.7)	0.6 (1.5)	35.0 (35.3)	34.0 (58.8)	149.2 (464.6)	0.98 (0.97)
Torso Acceleration	15.5 (15.8)	14.3 (54.3)	0.0 (4.9)	42.2 (76.0)	22.0 (76.0)	86.8 (235.4)	0.92 (0.87)
Pelvis Acceleration	1.3 (9.5)	20.7 (129.1)	11.2 (31.2)	54.4 (80.6)	26.6 (74.6)	131.4 (366.8)	0.71 (-0.27)
Head Anterior-Posterior Angular Velocity	19.7 (39.3)	32.1 (34.9)	6.9 (8.7)	289.5 (304.0)	1283.4 (2220.4)	22958.4 (55285.3)	0.76 (0.70)
Head Medial-Lateral Angular Velocity	2.6 (88.4)	8.5 (63.4)	12.3 (25.6)	722.6 (1637.1)	4020.2 (9675.0)	58897.65 (138761.9)	0.77 (0.41)
Upper Neck Force	2.5 (36.8)	2.8 (43.6)	1.2 (1.2)	21.0 (50.2)	18.5 (31.4)	89.6 (115.4)	0.98 (0.93)
Upper Neck Moment	5.6 (30.4)	3.2 (48.6)	0.9 (2.0)	12.0 (34.5)	11.5 (28.1)	56.8 (110.6)	0.97 (0.91)

* Acceptance criteria shown in parentheses

Table 4-1. Results of statistical tests to evaluate model validation; computer model vs. experimental mean for fall onto playground foam surface.

Simulation model output, experimental mean, and experimental min-max corridor time histories for each outcome measure were compared for falls onto the linoleum surface (Figures 4-9 through 4-11). Table 4-2 shows the results of the validation

statistical tests between the model and experimental means along with the acceptance criteria. The comparison of the model outcomes with the experimental means passed all statistical tests except one (the torso acceleration mean value test).

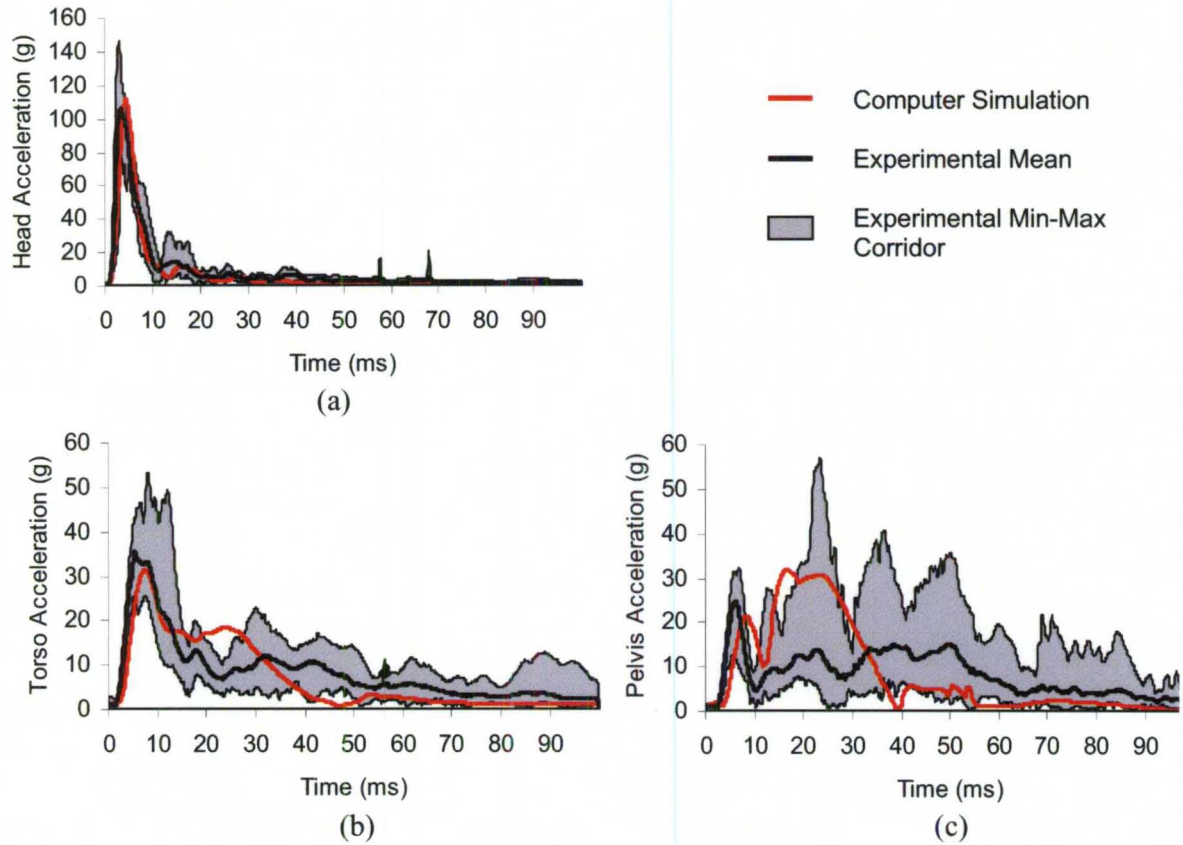


Figure 4-9. Computer simulation model and experimental time history comparisons for falls onto linoleum: (a) resultant linear head acceleration, (b) resultant linear torso acceleration, (c) resultant linear pelvis acceleration.

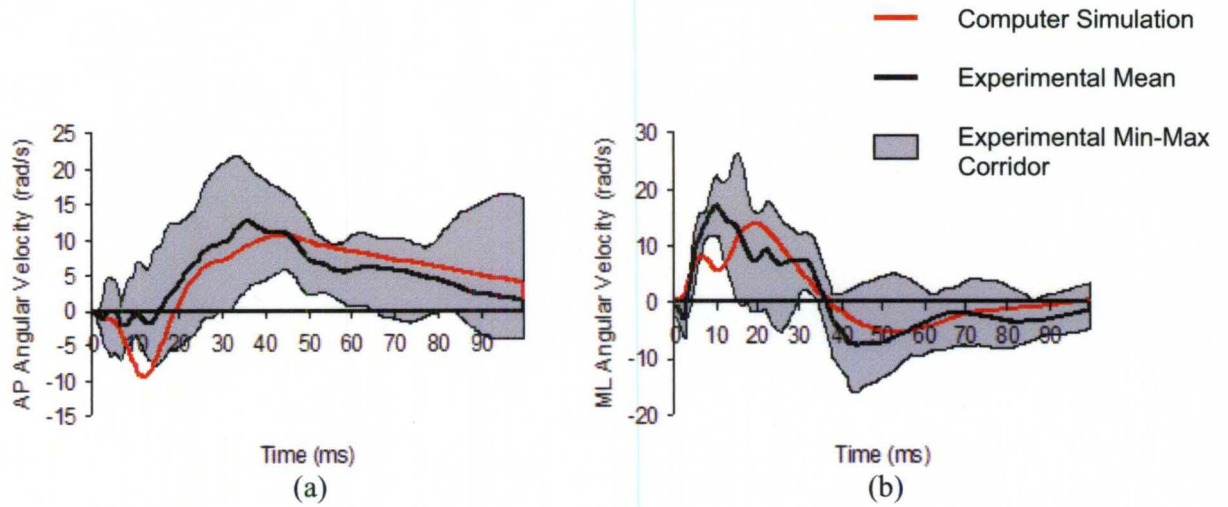


Figure 4-10. Computer simulation model and experimental time history comparisons for falls onto linoleum: (a) anterior-posterior (AP) angular head velocity, (b) medial-lateral (ML) angular head velocity.

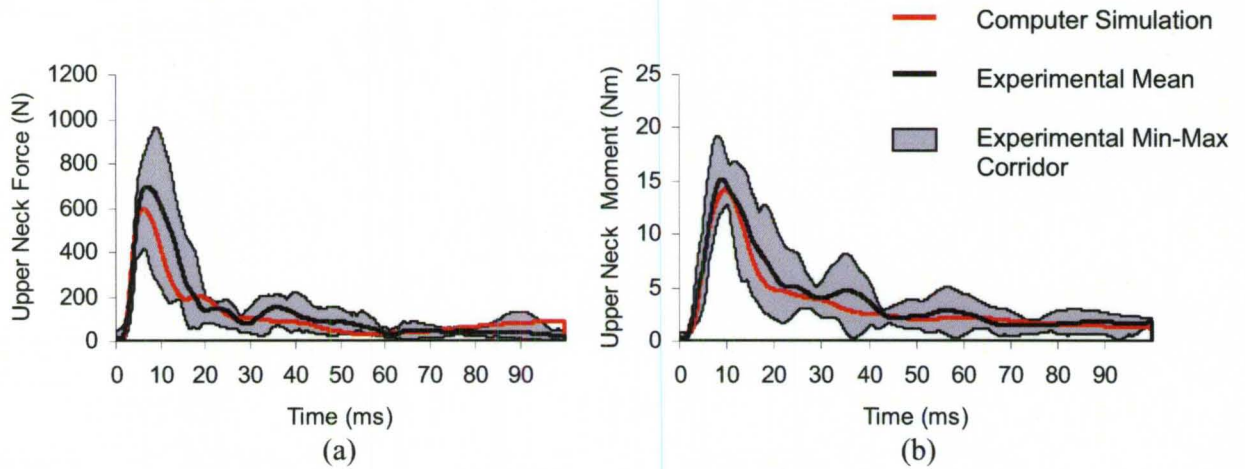


Figure 4-11. Computer simulation model and experimental time history comparisons for falls onto linoleum: (a) resultant upper neck force, (b) resultant upper neck moment.

Outcome Measure	Statistical Test and Acceptance Criteria*						Correlation Coefficient
	Mean Value	Peak Value		Relative Error			
	Difference (%)	Difference (%)	Time Difference (ms)	Mean (%)	Standard Deviation (%)	Maximum (%)	
Head Acceleration	7.3 (11.4)	5.2 (38.0)	1.1 (1.8)	26.2 (55.7)	16.9 (40.1)	82.3 (144.9)	0.94 (0.82)
Torso Acceleration	16.8 (14.7)	11.7 (50.0)	2.1 (6.7)	54.0 (73.3)	29.1 (83.3)	168.6 (327.8)	0.83 (0.83)
Pelvis Acceleration	0.9 (8.5)	28.2 (129.1)	10.5 (44.7)	84.0 (88.5)	51.4 (80.0)	273.6 (319.4)	0.42 (0.16)
Head Anterior-Posterior Angular Velocity	2.8 (91.4)	17.3 (73.1)	7.9 (9.2)	186.5 (281.2)	1603.1 (2558.1)	49693.0 (79399.8)	0.82 (0.57)
Head Medial-Lateral Angular Velocity	103.8 (252.6)	18.3 (54.8)	9.1 (13.0)	64.4 (161.8)	210.1 (1597.3)	5910.0 (50354.5)	0.87 (0.43)
Upper Neck Force	20.5 (31.8)	14.2 (40.5)	1.0 (2.2)	30.2 (45.4)	16.1 (39.5)	78.0 (270.7)	0.95 (0.87)
Upper Neck Moment	13.0 (30.9)	6.4 (27.1)	0.8 (4.4)	17.5 (43.6)	12.3 (28.7)	55.8 (122.4)	0.98 (0.85)

* Acceptance criteria shown in parentheses

Note: Shaded cell indicates validation criteria not met.

Table 4-2. Results of statistical tests to evaluate model predictive capability; computer model vs. experimental mean for fall onto linoleum surface.

Discussion

In this study, a computer simulation of a pediatric bed fall was developed and validated using experiments with a pediatric ATD. To the authors' knowledge, this is the first study that developed a computer simulation model of a short-distance fall using a 12-month-old ATD to represent the fall victim. The model was validated using both

qualitative and quantitative methods, and the predictive capability of the model was assessed by altering surface properties and verifying model outcome measures. A validated computer model of pediatric falls will be useful for future investigation of the influence of model parameters on injury outcome measures. Findings from such a study can provide an improved understanding of the relationships between fall parameters (including both child and environment characteristics) and injury potential in these falls.

During the model validation process, it was necessary to make several modifications to the 12-month-old CRABI model available within the MADYMO® database for use in our simulation model. In our simulated falls, the ATD rolled laterally off the “bed” surface and landed head-first with the lateral aspect of its head impacting the floor. Because of the head-first impact, a compression joint was added to the neck. Additionally, since the CRABI model was developed for use in high-energy motor vehicle crashes, head and neck properties were adjusted to more accurately represent the properties of the CRABI in short-distance falls (a relatively low-energy event). Components tests of the head and neck were conducted to determine more accurate mechanical properties for fall simulations.

Our pediatric bed fall model was validated following a rigorous procedure, and was based on those originally described by Dsouza and Bertocci ¹¹⁷ and Salipur and Bertocci ¹¹⁸. This validation procedure first qualitatively compared the event dynamics, followed by statistical methods to compare outcome measures between the simulation and physical experiments. Four statistical tests were used to compare different aspects of the simulation and experimental time-history curves. Validation criteria for each statistical test were based on the experimental range. This study used unique criteria for

each test and each outcome measure based on experimental variation in the fall scenario being modelled. Although the model passed each of the statistical validation tests, it was still necessary to assess of the predictive capability of the model. This was done by altering impact surface properties, running the fall simulation, and repeating the validation statistical tests. The results of this predictive assessment showed that the model was valid for all outcome measures except one (torso acceleration). The difference in the mean value of the torso acceleration did not meet the acceptance criteria for the model simulating a fall onto linoleum. However, the difference between the model result and criteria was fairly small (2.1 %). For the purposes of this study, the peak value, relative error and correlation tests represent more important aspects of comparison than the mean value test. The peak value is an important factor in assessing injury potential, and the relative error and correlation tests compare the outcome measure time histories over the entire impact duration. Since the torso acceleration (in the linoleum fall) passes the peak value, relative error, and correlation tests, and the time history profiles are in reasonable agreement (Figure 4-9), we consider this outcome measure valid along with the others that were assessed. In terms of the seven model outcome measures evaluated (head linear acceleration, torso linear acceleration, pelvis linear acceleration, head anterior-posterior angular velocity, head medial-lateral angular velocity, upper neck force, upper neck moment), our model provided a reasonable prediction of a 12-month-old CRABI fall onto a linoleum surface.

Although several studies have evaluated falls using computer simulation, most have focused on reconstructions of real-world fall events. Forero Rueda and Gilchrist¹¹², O’Riordain et al.⁷², Doorly and Gilchrist¹¹⁴, and Adamec et al.¹¹¹ reconstructed falls in

MADYMO® based on eye-witness accounts and information collected from the scene of the fall. The subjects in these studies ranged in age from 6 to 76 years. These studies use human body (non-ATD) models within MADYMO® to represent the subjects. Within MADYMO®, these human body models have been validated. However, no additional validation was performed by the authors of those studies for the fall scenario being modelled. After initial reconstruction of the fall event, the sensitivity of the model to initial conditions was investigated. These studies provide useful information about fall dynamics and model sensitivity to input parameters. However, the results are limited because no validation was performed of the specific scenario being modelled.

In a study by Schulz, a bed fall model of a Hybrid III adult ATD was created using LifeMOD software (LifeModeler, Inc; San Clemente, California), and the results were compared to a physical bed fall experiment with the ATD. The ATD was initially lying supine on a bed, and was rolled from the bed surface so that it impacted the floor head-first. Although the outcomes of the computer model were compared to the experimental outcomes, no validation process was conducted. Rather, several simulations were performed to determine the effect of 2-dimensional versus 3-dimensional modelling techniques as well as simulations beginning just before impact versus simulations of the entire fall. It was found that 3-dimensional simulations of the entire fall event provided head acceleration results most similar to those measured in the physical experiments.

Our computer simulation model has several limitations. Most importantly, the model was based on ATD experiments and thus retains any biofidelity limitations of the ATD in terms of representing a human child. This model is not intended to provide

absolute predictions of injury in pediatric falls. Rather the model was developed to study relationships between fall environment and ATD parameters and measures related to injury potential. Although the predictive capability of our model was assessed by altering a single parameter (impact surface), the model's predictive capability may be diminished with simultaneous changes in multiple input parameters. Additionally, our model's predictive capabilities are specific to the investigated scenario and are not generalizable to all types of pediatric falls or to children of varying ages experiencing a bed fall. In this study, seven outcome measures (head acceleration, torso acceleration, pelvis acceleration, head anterior-posterior angular velocity, head medial-lateral angular velocity, neck force, and neck moment) were used to validate the model. These outcome measures were selected because fall dynamics and head and neck injury measures will be investigated in future parametric sensitivity studies. In order to study other outcome measures (for example, extremity loading), those measures must also be included in the validation process. Lastly, it should be noted that computer simulations are simplified and discretized representations of real world events, and therefore may lack accuracy in predicting these events.

Conclusions

A computer simulation model of a 12-month-old child surrogate falling from a horizontal surface representing a bed has been developed. The model was validated using data from physical fall experiments conducted using a 12-month-old CRABI ATD to represent the fall victim. General comparison of fall dynamics, statistical comparison

of key outcome measures, and assessment of the model predictive capabilities were included in the validation process. This model will serve as a useful tool for studying relationships between fall parameters and injury potential. In future sensitivity analyses, fall environment and ATD parameters will be varied to investigate their effect on injury outcome measures (Chapter V). In particular, altering ATD properties within the model may lead to an improved understanding of child (fall victim) characteristics as they relate to injury risk in short-distance falls.

CHAPTER V
PEDIATRIC BED FALL COMPUTER SIMULATION MODEL PART II:
PARAMETRIC SENSITIVITY ANALYSIS

Overview

Falls from beds and other household furniture are common scenarios stated to conceal child abuse. Knowledge of the biomechanics associated with short-distance falls may aid clinicians in distinguishing between abusive and accidental injuries. In this study, a validated pediatric bed fall computer simulation model was used to investigate the effect of altering fall environment parameters (fall height, impact surface stiffness, initial force used to initiate the fall) and child surrogate parameters (overall mass, head stiffness, neck stiffness, soft tissue stiffness) on injury potential. The sensitivity of head and neck injury outcome measures to model parameters was determined. Parameters associated with the greatest sensitivity values (fall height, initiating force, and surrogate mass) significantly altered fall dynamics and impact orientation. This suggests that fall dynamics and impact orientation play a key role in head and neck injury potential. With the exception of surrogate mass, injury outcome measures tended to be more sensitive to changes in environmental parameters (bed height, impact surface stiffness, and initiating force) than surrogate parameters (head stiffness, neck stiffness, soft tissue stiffness).

Introduction

Falls from beds and other household furniture are common scenarios stated to conceal child abuse. A better understanding of the true injury potential associated with these falls is needed to aid clinicians in distinguishing between abusive and accidental injuries. Fall environment and child (fall victim) factors have been shown in previous studies to be related to injury potential in short falls^{64-68, 93}. However, many of these studies have been limited by the biofidelity of anthropomorphic surrogates used to represent the fall victim^{64-68, 93}. Moreover, little information is available regarding the injury tolerance and biomechanical response of children. Therefore, most pediatric surrogates are based on scaled adult cadaver or primate data and may not accurately represent a human child, particularly in low-energy events such as falls.

Computer simulation is a useful tool that can be used to investigate injury-producing events, and to study the effect of changing event parameters on injury potential. Parameters that can be altered include not only fall environment parameters (such as fall height and impact surface) but also child surrogate parameters (such as mass and mechanical properties of joints and tissues) which are not easily altered experimentally. By altering surrogate properties, this study will take a first step at addressing the issue of surrogate biofidelity. Computer simulation has been widely used by the automotive industry to study motor vehicle crash events, and has also been used in a few studies to investigate falls^{71-72, 111-115}. A computer simulation model of a 12-month-old child surrogate falling from an elevated horizontal surface such as a bed was previously developed and validated (Chapter IV). The purpose of this study was to use

the validated model to investigate the effect of altering fall environment and surrogate (fall victim) parameters on biomechanical measures and potential for injury.

Methods

A computer simulation model of a pediatric bed fall was previously developed using MADYMO® version 7.0 (MATHematical DYnamic Modeling; TNO, Netherlands) and validated using results from physical bed fall experiments with the Child Restraint Air-Bag Interaction (CRABI) 12-month-old anthropomorphic test device (ATD) (Chapter IV). In this study, the validated model was used to conduct a parametric sensitivity analysis. The purpose of this analysis was to investigate relationships between model parameters and injury potential. Fall environment and surrogate parameters were varied in the model, and the sensitivity of injury outcome measures to model parameters was determined.

Model Parameters

Eleven parameters were selected for variation (Table 5-1). Each parameter was varied individually in MADYMO® while all other parameters were held constant at their initial values from the validated model (baseline level). For the sensitivity analysis, each parameter was altered to +50%, +25%, -25%, and -50% of the baseline value. Once the parameter was altered, the computer simulation was run with the new values. This resulted in four simulation runs for each parameter (in addition to the baseline run which

was the original validated model). Additionally, parameter values from clinical and human cadaver studies were determined and the maximum and minimum values were used for additional computer simulation runs. This was done to include a real-world range of parameter values in the analysis. Details regarding each parameter are presented below.

Parameters	Injury Outcome Measures
Horizontal surface (bed) height	Peak resultant linear head acceleration
Impact surface (floor) stiffness	Head Injury Criterion (HIC)
Initial force (to initiate fall)	Peak resultant angular head acceleration
Surrogate mass	Peak resultant upper neck force
Surrogate head stiffness	Peak resultant upper neck moment
Surrogate neck stiffness (4 orientations):	
Axial compression	
Flexion/extension bending	
Lateral bending	
Torsional bending	
Surrogate neck damping	
Surrogate soft tissue stiffness	

Table 5-1. Altered computer model parameters and outcome measures used in sensitivity analysis.

1. Horizontal surface (bed) height – Height has been shown in biomechanical studies to influence injury risk in pediatric falls^{64-66, 68, 93, 109}. A clinical study of pediatric falls from horizontal surfaces was used to provide a real-world range of fall heights for simulation¹⁰⁹. The minimum (330 mm) and maximum (890 mm) surface heights measured in the clinical study were input into the model in addition to runs with $\pm 50\%$ and $\pm 25\%$ of the baseline bed height. The baseline surface height in the validated model was 608 mm.

2. Impact surface (floor) stiffness – Impact surface has been shown in biomechanical studies to influence injury potential in pediatric falls^{64-66, 68, 93}. The surface stiffness in the baseline model was specified to match that of playground foam (206 N/mm). Surface stiffness was adjusted to +50%, +25%, -25%, -50% of the baseline value for analysis.

3. Initial Force (to initiate fall) – To initiate the fall in both the model and physical experiments with the surrogate, an actuator impacted the posterior torso of the surrogate. The impact velocity of the actuator was measured in the experiments and replicated in the computer simulation. As initial force and velocity are not measurable parameters in most clinical falls, no information was found to establish a clinical range (based on real-world falls) for simulation. Therefore, initial force was only simulated at +50%, +25%, -25%, -50% of the baseline value. The baseline force was 140 N.

4. Surrogate mass – In the computer simulation, the surrogate represents a 50th percentile 12-month-old child (overall mass of 9.9 kg). For the sensitivity analysis, the overall mass was adjusted without any changes to mass distribution or body segment geometries. Realistically, mass distribution and body size would likely change with increasing or decreasing mass. However, for the purposes of this study, the effect of mass changes alone was investigated. In addition to the predetermined incremental mass changes ($\pm 50\%$ and $\pm 25\%$ of the baseline value),

the 5th (8.3 kg) and 95th (11.9 kg) percentile values for a 12-month-old child¹¹⁹ were also evaluated.

5. Surrogate head stiffness – The surrogate in the computer model is based on the CRABI 12-month-old ATD. Some have questioned the biofidelity of the CRABI head particularly in low-energy impacts such as falls^{66,91}. The biomechanical properties of the head and skull (represented in the model by a stiffness or force-displacement curve) are important when considering injury potential, particularly in head-first falls. In addition to the predefined incremental values, cadaveric studies reporting skull properties were used to define head stiffness values for analysis. Prange et al.¹¹⁰ conducted compression tests on three heads (ages 1-11 days) in two orientations (anterior-posterior compression and lateral compression). Skull stiffness did not appear to be dependent on orientation, but was found to be dependent on loading velocity (maximum velocity tested was 50 mm/s). Yoganandan et al.¹²⁰ tested six adult heads in compression (multiple orientations) under quasi-static loading and dynamic loading (7.1-8.0 m/s). The mean (dynamic values only) of the infant stiffness curves (Prange et al.) and adult stiffness curves (Yoganandan et al.) were used as minimum and maximum head stiffness properties for analysis. Figure 5-1 shows the head force-displacement curve used in the validated bed fall model (baseline) compared to experimentally determined cadaver data.

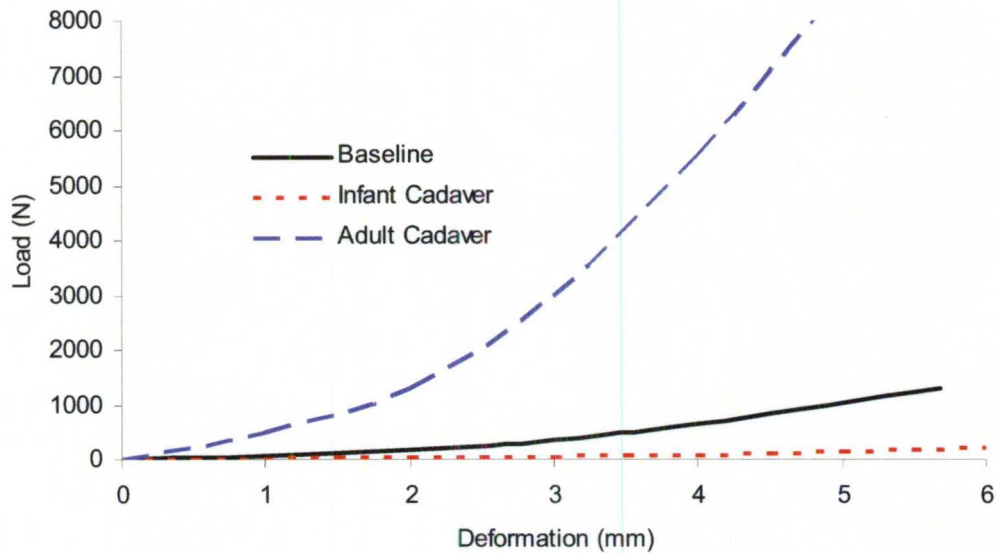


Figure 5-1. Head stiffness comparison for baseline (validated) bed fall model with adult cadaver experimental data ¹²⁰ and infant cadaver experimental data ¹¹⁰.

6. Surrogate Neck Stiffness – Just as head stiffness is expected to play a major role in head injury potential, neck stiffness is expected to affect neck injury potential. The baseline neck properties in the validated model match the stiffness properties of the CRABI neck. The CRABI neck is likely stiffer than a 12-month-old child’s neck, particularly in low-energy events such as short-distance falls (the CRABI was designed to study injury in high-energy automobile crashes). The computer model neck stiffness properties are represented by force-displacement and moment-rotation curves for four orientations: axial compression, flexion/extension, lateral bending, and torsion. Each neck parameter was varied independently. In addition to the predefined incremental values, human cadaveric data were used to define neck stiffness values for analysis. It should be noted that

cadaveric data presented below were measured quasi-statically. The dynamic neck stiffness would likely be greater than static stiffness due to the visco-elastic nature of human tissues. Therefore, it should be noted that the properties used in the analysis represent a lower bound of neck behavior.

- a. Flexion/Extension –Wheeldon et al.¹²¹ reported load-displacement curves for seven healthy adult subjects (ages 20-51 years) (Figure 5-2). Studies by Panjabi et al.¹²² and Schwab et al.¹²³ report similar or lower adult flexion/extension stiffnesses compared to those by Wheeldon. Therefore, the Wheeldon adult stiffness properties were used as the upper bound for neck flexion/extension stiffness in the parametric analysis. Ouyang et al.¹²⁴ reported load-displacement properties in flexion and extension for ten pediatric cervical spine cadaveric specimens (ages 2-12 years). Data for the youngest specimen (age 2) is shown in Figure 5-2. No other studies were found that report measured pediatric neck properties. However, several studies have used scaling parameters to study pediatric neck behavior. Kumaresan et al.¹²⁵ used a finite element model to study age differences in neck stiffness due to size, structure and material differences. This study estimated that the neck of a 1-year-old child is 175% more flexible than an adult neck in flexion and 400% more flexible in extension. Using this information, the adult properties (Wheeldon et al.) were scaled for a 1-year-old child. The scaled 1-year-old data is more flexible than the 2-year-old

cadaver data and was therefore used as a lower bound of neck stiffness in the parametric analysis (Figure 5-2).

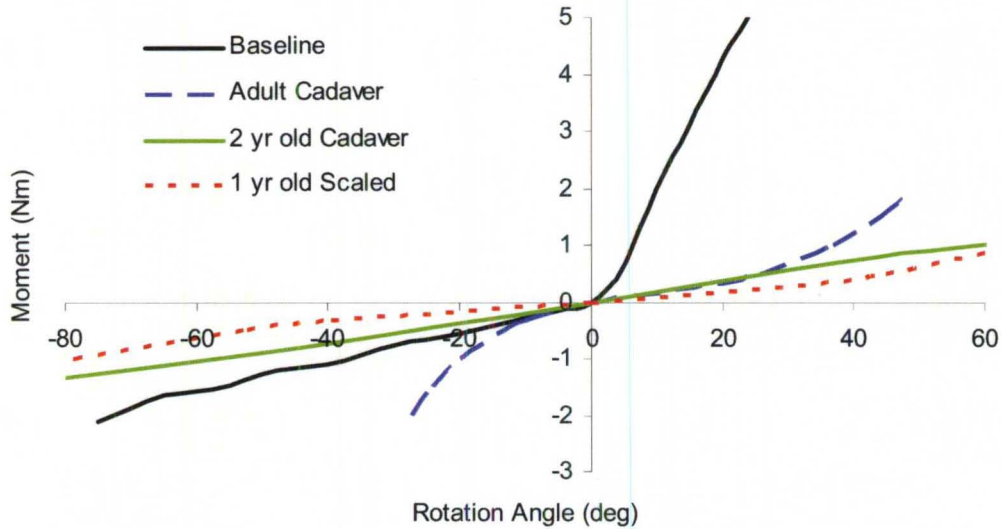


Figure 5-2. Neck flexion/extension stiffness properties for baseline (validated) bed fall model and cadaver experimental data for an adult ¹²¹, 2 year-old child ¹²⁴, and scaled results for a 1 year-old child.

- b. Lateral bending –Schwab et al. ¹²³ describes stiffness for the adult neck in lateral motion (Figure 5-3). No pediatric data or scaling factors were found for lateral motion. Therefore, only adult stiffness properties (in addition to the predefined incremental values) were evaluated in the parametric analysis.

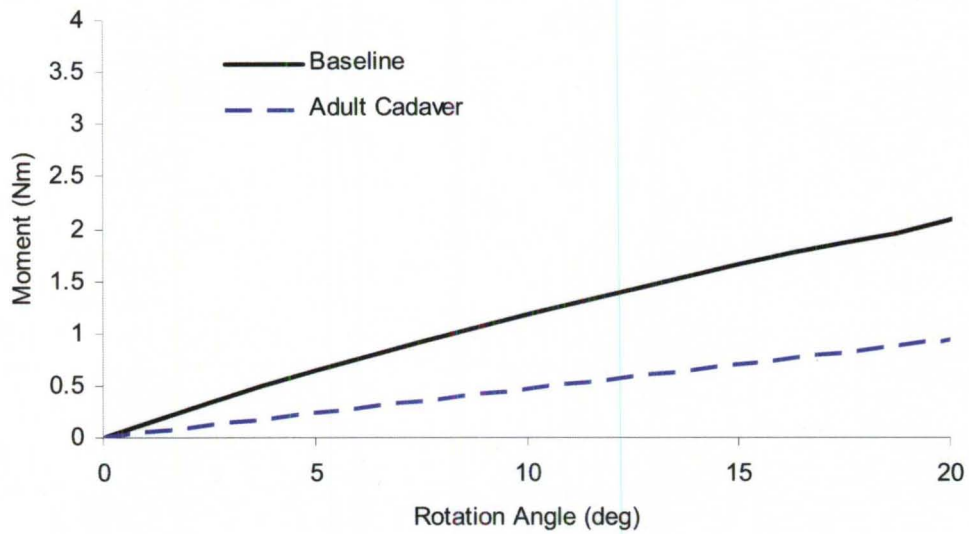


Figure 5-3. Neck lateral bending stiffness properties for baseline (validated) bed fall model and adult cadaver experimental data ¹²³.

- c. Torsion - Schwab et al. ¹²³ describes stiffness for the adult neck in torsion (Figure 5-4). No pediatric data or scaling factors were found for torsional loading. Therefore, only adult stiffness properties (in addition to the predefined incremental values) were evaluated in the parametric analysis.

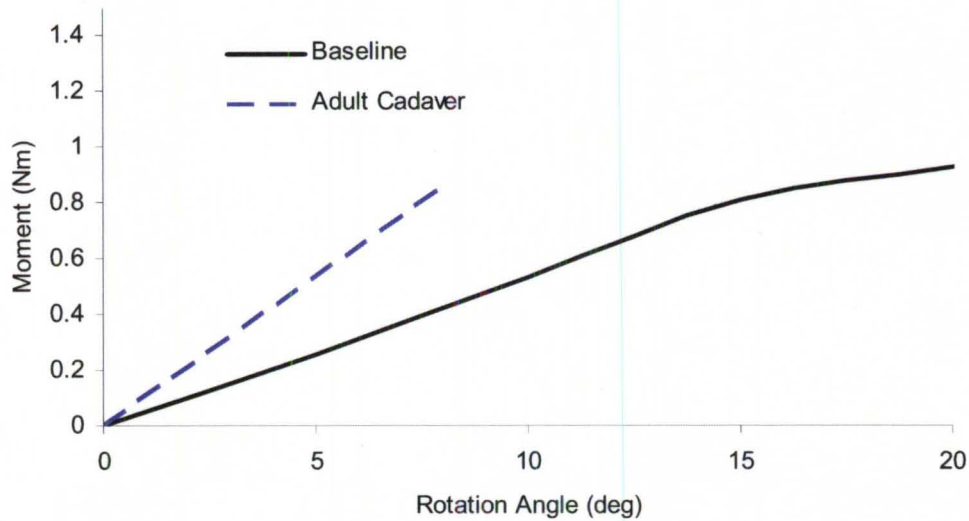


Figure 5-4. Neck torsional stiffness properties for baseline (validated) bed fall model and adult cadaver experimental data ¹²³.

- d. Axial Compression – Shea et al. ¹²⁶ describes adult neck stiffness in axial compression (Figure 5-5). Additionally, the finite element study by Kumaresan et al. ¹²⁵ estimated that the neck of a 1 year-old child is 500% more flexible than an adult neck in compression. Using this scaling factor, the stiffness properties found by Shea et al. were scaled to estimate a 1-year-old child’s neck compression stiffness (Figure 5-5). Both the adult and scaled infant properties were included in the analysis.

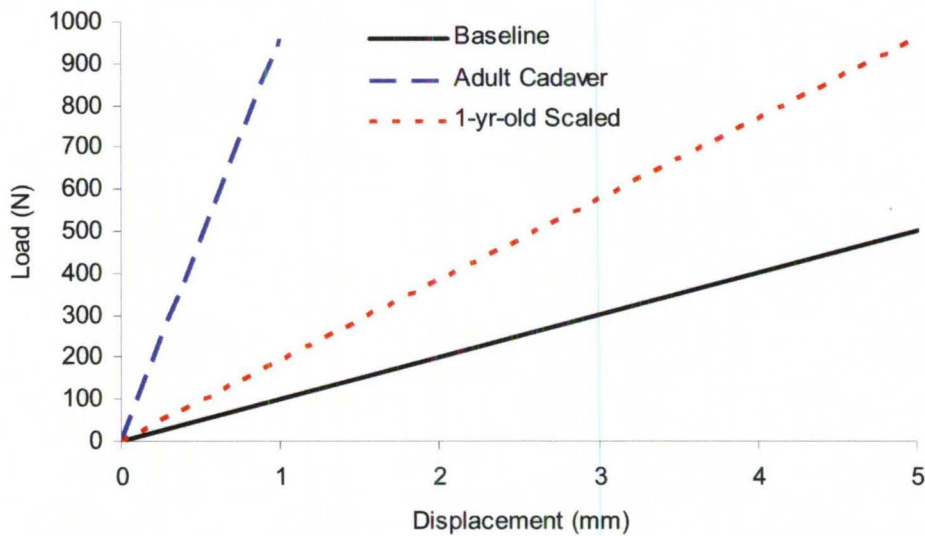


Figure 5-5. Neck compression stiffness properties for baseline (validated) bed fall model, adult cadaver experimental data ¹²⁶, and scaled results for a 1-year-old child.

7. Surrogate Neck Damping Coefficient – In the computer simulation model, joint properties (including neck properties) are represented by both stiffness and damping coefficient parameters. Stiffness relates force to the amount of displacement in joint, and damping relates force to the joint velocity. The damping coefficient is a dynamic property that creates a rate-dependent force opposing joint motion. Unlike neck stiffness properties, damping coefficients for cadaveric neck specimens have not been measured. However, damping properties are an important component in mathematical or computer models to define rate-dependent material behavior. The neck damping coefficient was altered to +50%, +25%, -25%, -50% of the baseline value which was 0.4. Note that in the validated model, the damping coefficient is uniform for all neck bending orientations.

8. Surrogate Soft Tissue Stiffness – Obesity is a growing problem in children, but the effect of child weight and body fat content on injury risk in falls is unclear. Thompson et al.¹⁰⁹ reported that in short-distance falls, children with more severe injuries had a significantly lower body mass index (BMI) than children with minor injuries. It is likely that these differences were due in part to soft tissue stiffness. Additionally, the soft tissue stiffness of the CRABI ATD is greater than that of a human child. This is because the ATD was designed to withstand repeated impact tests and soft tissue injuries were not of interest in this type of testing. A few studies have measured soft tissue stiffness of adult subjects using indentation tests¹²⁷⁻¹²⁹. However, these tests were done for small skin indentations/displacements (< 5 mm). The results of the skin indentation tests could not be extrapolated for the parametric analysis because of the non-linear nature of soft tissue stiffness properties. The baseline soft tissue stiffness properties used in the validated bed fall model are shown in Figure 5-6.

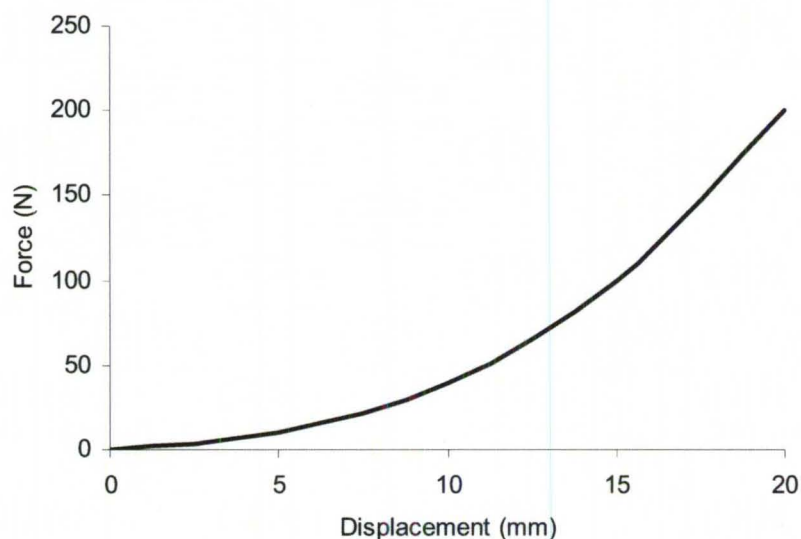


Figure 5-6. Soft tissue stiffness for the baseline (validated) bed fall model.

Outcome Measures

Changes in fall dynamics due to changing input parameters were qualitatively assessed. Additionally, five outcome measures relating to head and neck injury potential were assessed (Table 5-1). Head linear and angular accelerations were measured at the center of mass of the head. Neck forces and moments were measured at the superior aspect of the neck (approximately the C1 vertebrae location). The Head Injury Criterion (HIC) is a measure of head injury risk in impacts. HIC_{15} values are calculated using the linear head acceleration time-history.

Sensitivity Analysis

Sensitivity was defined as the ratio of change in the outcome measure over the change in the input parameter. Because several of the input parameters are represented by curves rather than single values, the changes were specified as a percentage of the baseline value. Greater sensitivities indicate a greater change in the outcome measure for a particular parameter. Additionally, a positive sensitivity indicates a positive or direct relationship between the parameter and outcome measure (e.g. increasing parameter resulted in increasing outcome measure). Conversely, a negative sensitivity indicates a negative or inverse relationship between the parameter and outcome measure (e.g. increasing parameter resulted in decreasing outcome measure). Sensitivity values were calculated for all combinations of parameters and outcome measures (except fall dynamics). Since each parameter was associated with multiple sensitivity values (for

simulation runs at +50%, +25%, -25%, and -50% of the baseline value), the mean sensitivity for each parameter was determined and used for parameter sensitivity comparisons.

Results

The results of all simulation runs are shown in Figures 5-7 through 5-12. Additionally, the sensitivity values of the outcome measures to each parameter are shown in Table 5-2.

Fall Dynamics

Changes in bed height, the initial force to initiate the fall, and surrogate mass produced considerable changes in fall dynamics (Figure 5-7). With increasing bed height, the surrogate had more time to rotate about its longitudinal (superior-inferior) axis before impact and thus, landed more on its side. In falls with bed heights less than the baseline value, the surrogate landed in a more prone position.

The initial force of the fall affected the manner in which the surrogate left the bed surface. In the baseline model, the surrogate was impacted with enough force to initiate the rolling motion, but once the surrogate reached the edge of the bed surface, the actuator was no longer in contact with the torso, and the force of gravity caused the surrogate to fall from the bed. In simulations with initial forces greater than the baseline value, the increased force applied at the mid-torso caused the legs of the surrogate to lead

in the fall, so that the surrogate landed feet-first (rather than head-first). In the simulation with an initial force set at -25% of the baseline value, the surrogate landed head-first at a slightly greater angle of impact relative to the ground (feet were higher at moment of impact). In the simulation of -50% of the baseline initial force, there was not a great enough force to push the surrogate from the bed surface. Therefore, this simulation was not included in the results.

Surrogate mass changes affected the impact orientation. Simulations with increasing mass resulted in a greater angle of impact (feet higher at the moment of impact), and simulations with decreasing mass resulted in a smaller angle at impact (feet closer to the ground at the moment of impact). In the simulation with the smallest mass (-50% of baseline), the surrogate's feet impacted the ground before the head.

No visible changes in fall dynamics were present for variations in any of the other parameters (surface stiffness, head stiffness, neck stiffnesses, neck damping coefficient, and soft tissue stiffness).

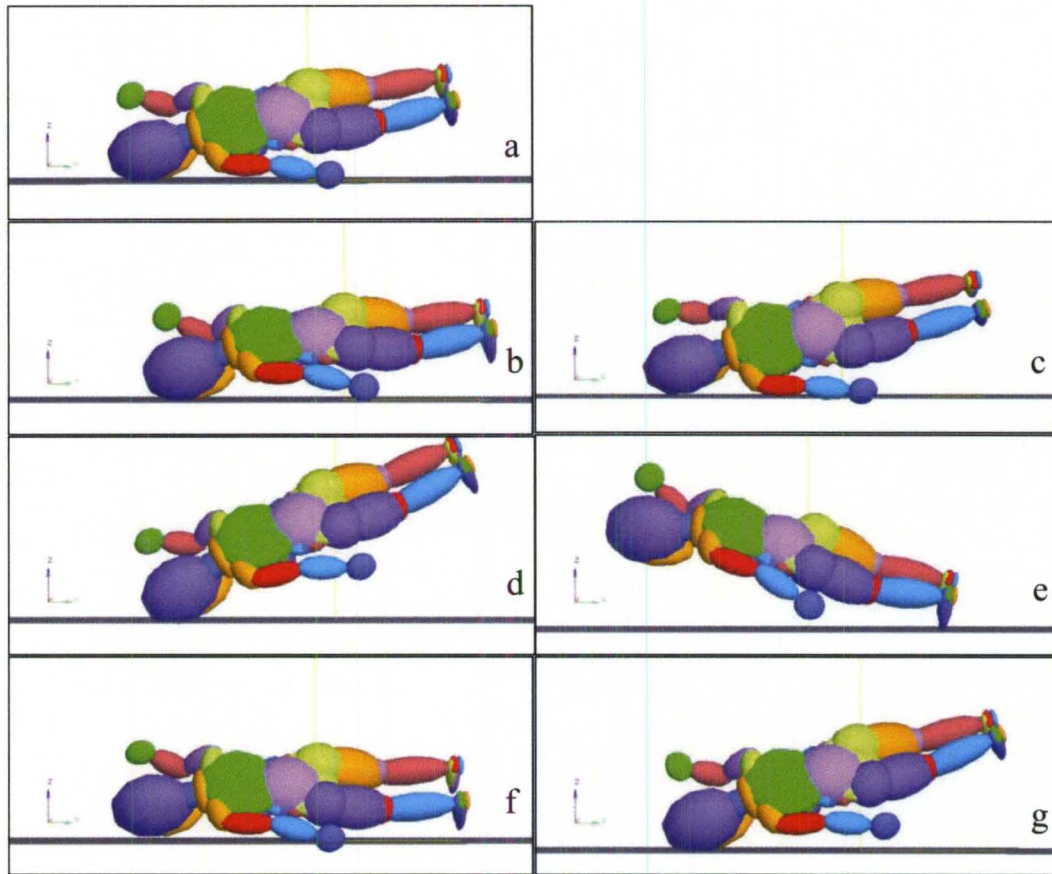


Figure 5-7. Orientation of the surrogate upon impact with the floor surface for parameters that significantly altered fall dynamics: (a) baseline (validated) model, (b) model with bed height set at -25% of the baseline, (c) model with bed height set at +25% of the baseline, (d) model with initial force set at -25% of the baseline, (e) model with initial force set at +25% of the baseline, (f) model with surrogate mass set at -25% of the baseline, (g) model with surrogate mass set at +25% of the baseline value.

Head Injury Measures

Peak linear head acceleration and HIC_{15} values were most sensitive to changes in surrogate mass (Table 5-2). Additionally, there was an inverse relationship between mass

and head injury outcome measures. Increasing the surrogate's mass resulted in decreasing peak linear accelerations, peak angular head accelerations and HIC_{15} values. Angular head accelerations were most sensitive to the initial force used to initiate the fall; increasing the initial force resulted in increasing peak angular head accelerations. The influence of initial force on linear head accelerations, however, was less pronounced. Bed fall height, surface stiffness, and surrogate head stiffness had direct relationships with head injury outcome measures (increasing fall height, increasing surface stiffness, and increasing head stiffness resulted in increasing peak linear head accelerations, peak angular head accelerations, and HIC_{15} values). Altering neck properties and soft tissue stiffness had little influence on head injury outcome measures.

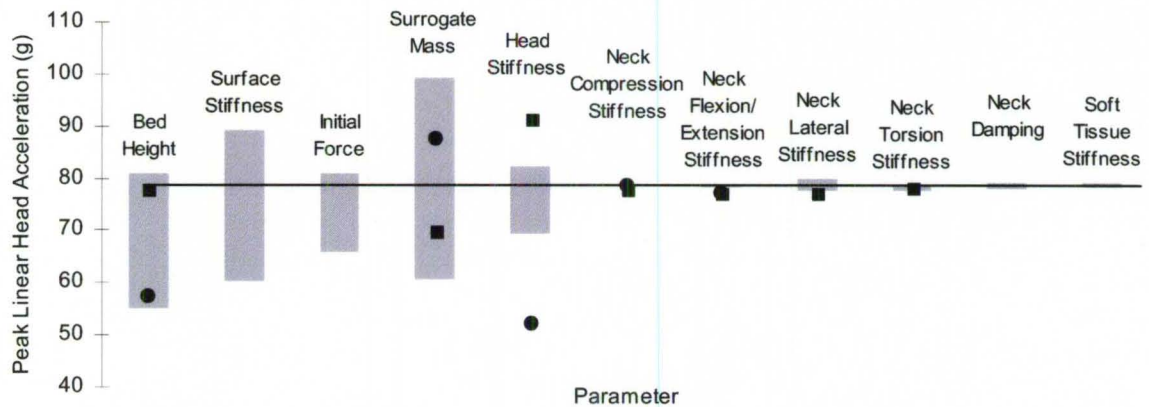


Figure 5-8. Peak resultant linear head acceleration for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the range for parameter values +/-50% of the baseline; the square and circle markers indicate the outcome values associated with the maximum and minimum parameter values from the literature, respectively.

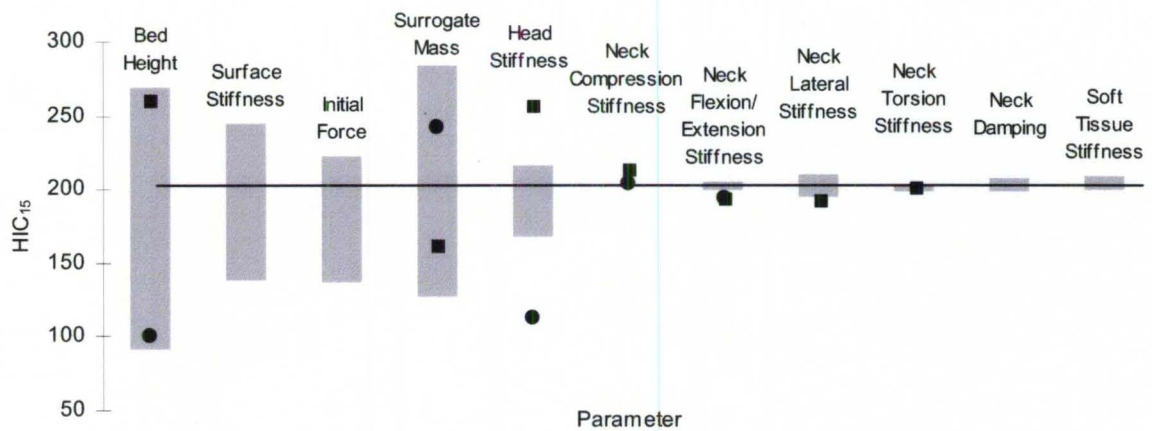


Figure 5-9. Peak HIC₁₅ for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the range for parameter values +/-50% of the baseline; the square and circle markers indicate the outcome values associated with the maximum and minimum parameter values from the literature, respectively.

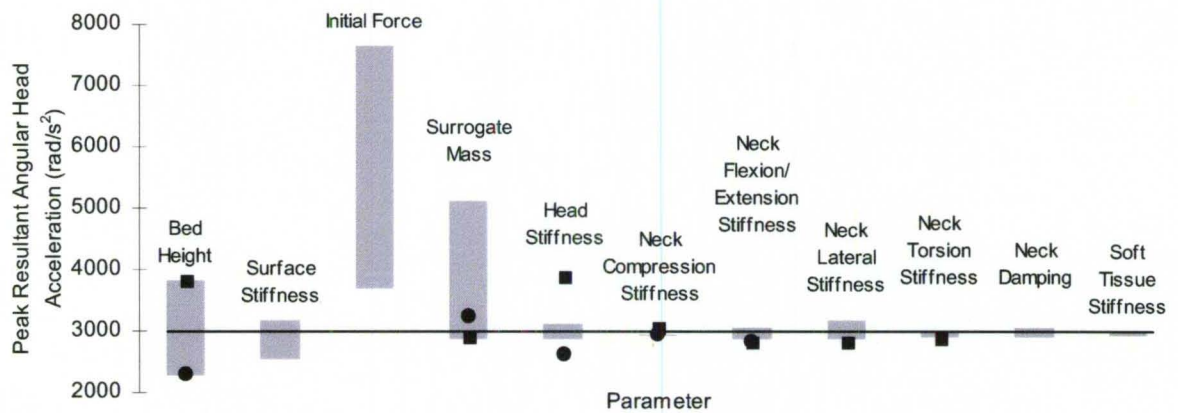


Figure 5-10. Peak resultant angular head acceleration for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the range for parameter values +/-50% of the baseline; the square and circle markers indicate the outcome values associated with the maximum and minimum parameter values from the literature, respectively.

Neck Injury Measures

Peak resultant neck force was most sensitive to changes in the initial force used to initiate the fall and peak neck moment was most sensitive to neck damping coefficient. Unlike the head injury measures, however, initial force had an inverse relationship with neck forces and moments (increasing initial force resulted in decreasing neck forces and neck moments). Surrogate mass had a direct relationship with neck loads (increasing mass resulted in increasing neck forces and neck moments). With the exception of neck compression stiffness, which had a direct relationship with peak resultant neck force, and neck damping coefficient, which had a direct relationship with peak resultant neck moment, neck parameters had little influence on neck loads. Additionally, bed height, surface stiffness, and soft tissue stiffness had small influences on neck forces and neck moments.

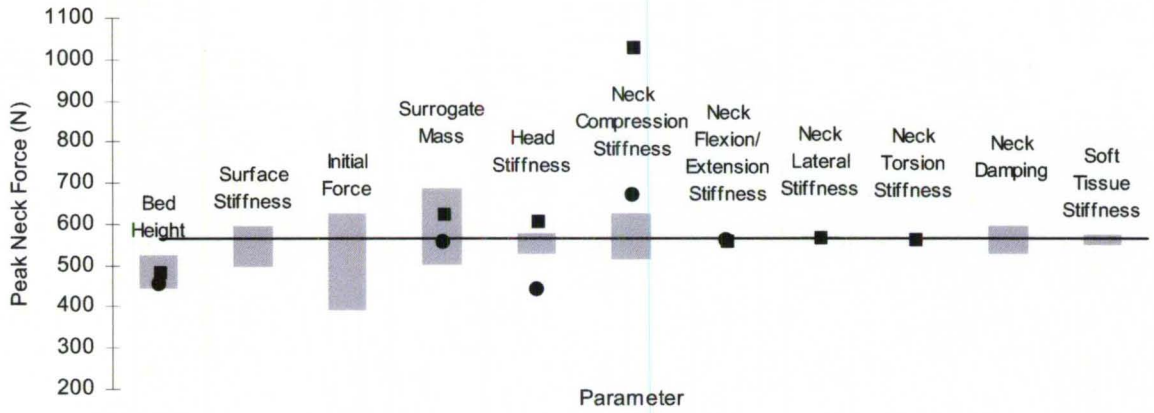


Figure 5-11. Peak resultant neck force for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the range for parameter values +/-50% of the baseline; the square and circle markers indicate the outcome values associated with the maximum and minimum parameter values from the literature, respectively.

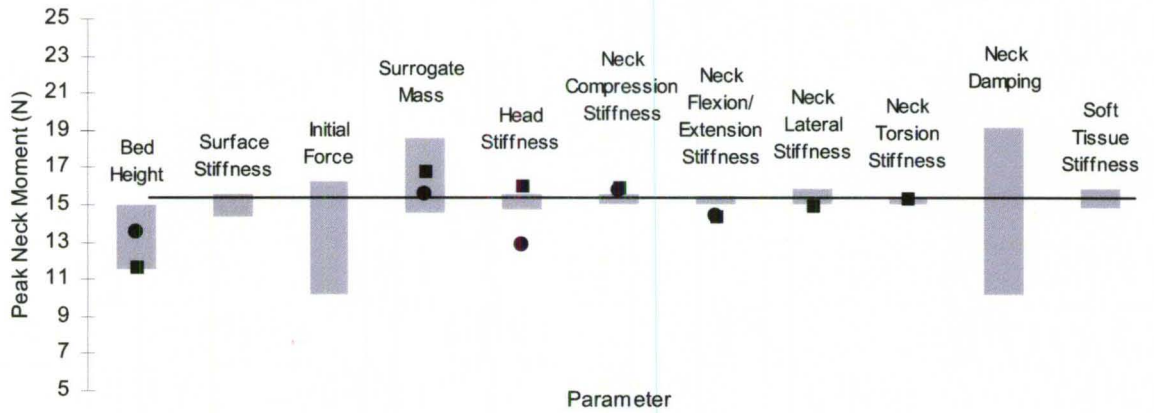


Figure 5-12. Peak resultant neck moment for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the range for parameter values +/-50% of the baseline; the square and circle markers indicate the outcome values associated with the maximum and minimum parameter values from the literature, respectively.

Parameters	Outcome Measures				
	Peak Resultant Head Linear Acceleration	HIC ₁₅	Peak Resultant Head Angular Acceleration	Peak Resultant Neck Force	Peak Resultant Neck Moment
Fall Height	0.31	0.90	0.49	0.05	-0.16
Surface Stiffness	0.36	0.52	0.22	0.17	0.07
Initiating Force	0.11	0.30	2.83	-0.50	-0.39
Surrogate Mass	-0.56	-0.95	-0.79	0.33	0.28
Head Stiffness	0.15	0.23	0.09	0.08	0.05
Neck Compression Stiffness	0.00	0.00	0.01	0.20	0.04
Neck Flexion/ Extension Stiffness	0.01	0.03	0.05	-0.01	0.04
Neck Lateral Stiffness	0.03	0.08	0.10	0.00	0.05
Neck Torsion Stiffness	0.01	0.02	-0.03	0.00	0.02
Neck Damping Coefficient	-0.02	-0.05	-0.06	0.13	0.58
Soft Tissue Stiffness	-0.02	-0.05	-0.03	0.05	0.07

Table 5-2. Mean sensitivity of outcome measures to each model input parameter.

Discussion

With the exception of surrogate mass and neck damping coefficient, injury outcome measures tended to be more sensitive to changes in environmental parameters (bed height, impact surface stiffness, initial force) than surrogate parameters (head stiffness, neck stiffness, soft tissue stiffness). Increasing bed height and increasing surface stiffness led to increases in the head injury measures. This is consistent with previous studies that have shown fall height and impact surface to significantly affect head injury risk in short-distance falls^{64-68, 93}. Increasing the initial force or initial velocity of the child prior to the fall tended to increase head injury measures, but decrease the neck injury measures. The neck loads were likely reduced in falls with increasing initial force due to changes in impact dynamics. With a more horizontal impact orientation, less force is transferred through the neck as the left arm and torso impact the

ground sooner. Relationships between initial force or initial velocity of the child prior to the fall and injury potential have not been studied previously. Factors that could increase the initial velocity of the child in a real fall could include the child being pushed from the surface or the child playing/moving around on the bed (or other elevated surface). Increases in initial force resulted in substantial increases in peak head angular acceleration (up to 160%) and should therefore be considered in future assessments of head injury potential.

Three parameters were found to influence fall dynamics: bed height, initial force, and surrogate mass. These three parameters also tended to have the largest influence on the outcome measures. This suggests that fall dynamics, particularly the orientation of the surrogate upon impact with the ground, play a significant role in head and neck injury potential in falls. This has been shown previously in free fall experiments with a 12-month-old ATD⁹³. Thompson et al. found that slight changes in fall dynamics due to changes in the overall fall height significantly influenced head injury risk.

Of the surrogate parameters varied, mass had the largest influence on head and neck injury outcome measures. Increasing surrogate mass tended to decrease head injury measures but increase neck injury measures. This is counterintuitive because increasing mass generally results in acceleration increases. . Two factors contributed to this finding. First, in all simulations with changing mass, actuator kinematics were held constant. Therefore, the increased mass of the surrogate likely reduced the load transmitted from the actuator to the surrogate. This, in combination with increased friction between the surrogate and bed surface, reduced the initial velocity of the surrogate (after contact with the actuator but just prior to the fall). The second factor

contributing to the inverse mass/head acceleration relationship was impact orientation. In falls with increasing mass, the surrogate impacted the ground at a greater angle (feet higher above ground at moment of impact). With this greater impact angle, the impact force was transferred primarily from the head through the neck (as no other body segments were in contact with the ground) which also explains the increased neck loads. The neck stiffness is much lower than the head stiffness, and this effectively increased the head impact duration (Figure 5-13). Larger impact durations have been shown to be associated with reductions in peak linear and angular head accelerations in falls⁹³. It should be noted that despite decreases in head acceleration measures with increasing mass, the head contact force increased with increasing mass (Table 5-3). These results suggest that acceleration alone may not be sufficient for predicting head injury potential in impacts. Acceleration measures and HIC do not account for variations in head or surrogate mass. It should also be noted that the range of surrogate mass used in the sensitivity analysis exceeds the normal range for a 12-month-old child. Simulations of mass values for a 5th percentile and 95th percentile 12-month-old child resulted in a smaller range for all outcome measures than results indicated by the simulations with mass $\pm 50\%$ of the baseline (50th percentile 12-month-old child) mass. Therefore, the influence of surrogate mass on injury potential may be exaggerated in this study.

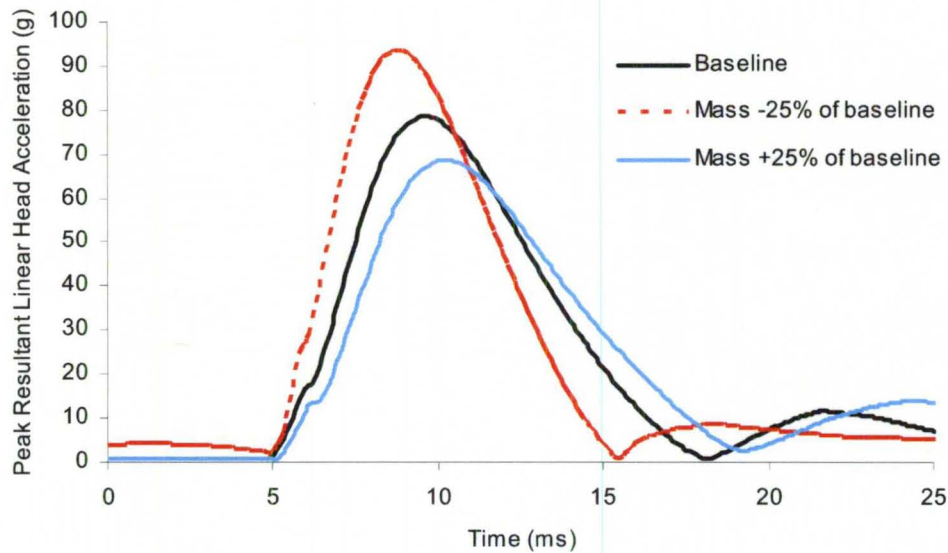


Figure 5-13. Peak resultant linear head acceleration time-histories for baseline (validated) bed fall model and simulations with surrogate mass set at $\pm 25\%$ of the baseline value.

Surrogate Mass (kg)	Peak Resultant Head Impact Force (N)
9.9 (baseline)	2771
4.9 (-50% of baseline)	1800
7.4 (-25% of baseline)	2406
12.3 (+25% of baseline)	3131
14.8 (+50% of baseline)	3449

Table 5-3. Peak resultant head impact force versus surrogate mass.

Surrogate head stiffness influenced peak linear head accelerations and HIC_{15} values, but had little influence on peak angular head accelerations and neck injury measures. As expected, increases in head stiffness resulted in increases in peak linear head accelerations. Head stiffness properties from the literature describing skull stiffness of infant and adult cadaver specimens were included in the analysis. This resulted in a

much larger range for all outcome measures than results of the analysis with $\pm 50\%$ of the baseline head stiffness. This suggests the influence of head stiffness on injury potential may be underestimated in this study.

Neck parameters, with the exception of axial compression stiffness and neck damping coefficient, and soft tissue stiffness had little effect on head and neck injury outcome measures. Increases in neck compression stiffness led to increases in the peak neck force. Because of the head-first impact orientation in the baseline (validated) model, the forces transmitted through the neck were primarily in the axial direction. Thus, compression of the neck dominated the resultant neck force. Increases in the neck damping coefficient led to increases in the peak neck moment. Because the damping load opposes joint motion, increasing the damping coefficient effectively reduced neck bending motion. The reduced neck motion led to increases in the neck moments. In the computer simulation model, neck bending moments were more sensitive to neck damping parameters than neck stiffness parameters. In experimental studies of neck properties, however, only neck stiffnesses are measured. Future work investigating rate-dependent neck properties is needed to improve accuracy in modeling surrogate neck properties.

A few studies have investigated the effect of fall parameters on injury risk using computer simulation^{71-72, 112}. Mohan et al. reconstructed seven real-world head-first free falls (six subjects were children ages 1-10 years and one subject was a 21-year-old adult) using a 2-D computer model. Impact angles were varied over 20 degrees, but were found to have a minimal effect on head impact response outcomes in the children, and a more pronounced effect in the adult fall simulation. These results differ from our study, but the Mohan surrogate model was much more simplistic (body represented by nine masses

separated by ten linkages and detailed anthropometric measurements such as head geometry were not included). Additionally, Mohan et al. reported reduced head impact response outcomes with reduced surface stiffness. O’Riordain et al. and Forero Rueda and Gilchrist reconstructed real-world falls in MADYMO®. O’Riordain et al.⁷² simulated four falls (subjects aged 11-76) with varying head stiffness properties and initial velocities. As with our study, reducing the head stiffness led to reductions in peak head linear and angular accelerations. Effects of initial velocity were less pronounced than those of head stiffness. Initial velocities were adjusted by ± 0.1 m/s (linear) and ± 0.1 rad/s (angular), but actual velocities were not presented. Therefore, it is possible that the changes in initial velocity simulated by O’Riordain were less than the 25% and 50% changes used in our study. O’Riordain et al found that increasing initial velocities led to decreases in the peak linear head accelerations. This was attributed to changes in fall dynamics and energy absorption by other parts of the body. Forero Rueda and Gilchrist¹¹² simulated a fall by a 6-year-old child from a playground frame. Surface properties and impact orientation parameters were varied, and both were found to have a significant effect. Reductions in surface stiffnesses reduced head injury outcome measures. Impact orientations with the surrogate in a horizontal prone position were associated with a greater head injury risk than side-lying, supine, or feet-first postures. Orientations with the head leading were not simulated. No studies were found that investigated the effect of neck properties or soft tissue properties on injury risk.

This study has many limitations. First, the results should not be used to make absolute predictions of injury occurrence in pediatric falls. Rather, relationships between model parameters and injury potential were of interest. Due to the lack of information

regarding pediatric injury tolerance and biomechanical response of pediatric tissues, the model simulates an anthropomorphic test device (CRABI) representing a child but with limited biofidelity. The CRABI is anthropometrically similar to a 12-month-old child, but the head and neck are stiffer than an actual child's. This study attempted to address concerns about CRABI biofidelity by investigating the effect of varying head and neck properties on injury outcome measures. Results of changing surrogate mass are limited in that they did not include any changes in anthropometrics, overall size or mass distribution. Additionally, it should be noted that joint properties (as with the neck) and contact characteristics (as with head and other body segments contacting the ground surface) are defined by both stiffness and damping parameters. Neck loads were influenced by damping properties, and the combination of stiffness and damping effects should be studied further. Similarly, damping coefficients of head, soft tissue, and surface properties may influence injury outcome measures but were not investigated in this study. Finally, parameters in this study were varied individually, and thus, no interaction effects between parameters were determined. However, multiple parameter changes simultaneously may affect the model validity, and were therefore not simulated in this study.

Conclusion

In this study, a validated computer simulation model of an anthropomorphic surrogate representing a 12-month-old child rolling off of a bed or other horizontal surface was used to investigate the influence of fall environment and child surrogate

parameters on injury potential. The sensitivity of head and neck injury outcome measures to model parameters was determined. Parameters associated with the greatest sensitivity values (fall height, initiating force, and surrogate mass) significantly altered fall dynamics and impact orientation. This suggests that fall dynamics and impact orientation play a key role in head and neck injury potential. With the exception of surrogate mass and neck damping, injury outcome measures tended to be more sensitive to changes in environmental parameters (bed height, impact surface stiffness, initiating force) than surrogate parameters (head stiffness, neck stiffness, soft tissue stiffness). This has important implications for ATD biofidelity. Differences in head, neck, and soft tissue properties between the CRABI ATD and an actual human child may play a smaller role in injury risk assessments of short falls than previously thought, especially in comparison to fall environment parameters.

CHAPTER VI
SUMMARY AND CONCLUSIONS

Key Findings and Clinical Implications

The purpose of this study was to provide objective information about injury potential in short-distance household falls that can aid clinicians in distinguishing between inflicted and non-inflicted injuries in children. This study involved three methodological components. The first component was a prospective case-based biomechanical assessment of children who presented to the emergency department of a metropolitan children's hospital with a history of a fall from a bed or other similar furniture. Descriptions of fall dynamics and fall environment were obtained through interviews with the caregivers and in-depth scene investigations. The second component utilized an anthropomorphic test device (ATD), or human surrogate, representing a 12-month-old child, to experimentally simulate falls from furniture surfaces in a laboratory setting. The final component involved development of a validated computer model based upon the ATD experiments. The computer model extended beyond the experiments by allowing variation in fall parameters and ATD characteristics.

Overall, the risk of severe or life-threatening injury in short-distance household falls is low. Fractures of the skull and extremities may result from these falls (21.5% of falls resulting in Emergency Department visits). 2 of 79 fall cases involved small,

contact-type subdural hematomas (SDH). It should be noted that the clinical presentation and course for these children was benign. Very few studies have reported intracranial hemorrhages resulting from short-distance falls.^{52, 55} The 2 cases with SDH in our study both had unique fall dynamics that contributed to their injuries. Both resulted from fall heights greater than 1 m. One child (initially seated on the back of sofa) rotated rearward off the back of a sofa and landed directly on her head. The second child struck his head on a hard object (humidifier) during the fall.

Results of ATD experiments regarding injury potential in short-distance falls support those from the clinical study with the exception of neck injury potential. Based on the experimentally measured neck loads, published pediatric neck injury thresholds suggest a substantial risk of AIS 3 neck injury. However, this is not consistent with epidemiological studies that suggest neck injuries in short-distance falls are rare. Limitations in surrogate neck biofidelity and published pediatric neck injury thresholds likely contribute to this discrepancy. Future studies are needed to both improve ATD neck biofidelity and determine more accurate pediatric neck injury thresholds.

In each study component (clinical, anthropomorphic surrogate experiments, computer simulation), relationships between fall environment and child/surrogate parameters and injury potential were investigated. As with previous biomechanical studies of falls^{64-67, 93}, fall environment parameters (fall height and impact surface type) were found to influence injury potential. To our knowledge, this is the first study to investigate the influence of child or surrogate parameters on injury potential. Child/surrogate body mass index, overall mass, head stiffness, and neck properties influenced injury potential in these falls.

Through the parametric sensitivity analysis, it was found that fall environment and surrogate parameters that altered fall dynamics had the greatest influence on head and neck injury potential. In a previous study of feet-first free falls with a 12-month-old anthropomorphic surrogate⁹³, differences in fall dynamics due to changing fall height resulted in a unique finding regarding head injury potential; increasing fall heights were associated with reduced head accelerations. Similarly in the present study, changes in fall dynamics produced results that were initially counterintuitive. In the parametric sensitivity analysis, increasing surrogate mass resulted in changes to fall dynamics that effectively reduced head accelerations. This implies that increasing surrogate mass reduces head injury potential (as greater accelerations are generally associated with a greater risk of head injury). However, despite reduced head accelerations, the head contact force with the ground surface increased with increasing mass. These results suggest that head accelerations alone may not be sufficient in predictions of head injury potential in impacts. New pediatric head injury criteria are needed that account for the mass of the child/surrogate.

The results of this study may aid clinicians in assessing compatibility between a child's injuries and the stated cause when the scenario provided is a short-distance fall, thus improving accuracy in child abuse diagnoses. Additionally, results highlight the need for detailed case histories when making injury assessments that include fall environment factors (fall height and impact surface type), child factors (age, mass, body mass index), and descriptions of the fall dynamics and impact orientation of the child.

Limitations and Recommendations for Future Work

This study applied biomechanical techniques and knowledge to investigations of injury potential in short-distance pediatric falls. However, the approaches used have limitations in their applicability. Future studies are recommended to address some of the limitations of this work.

The first component of this work involved case-based assessments of real-world pediatric falls. The prospective design of this study allowed detailed biomechanical assessments including fall scene investigations. This built upon previous epidemiological fall studies that were limited to information contained in medical records. However, the sample size in this study was relatively small (79 cases). Few biomechanical measures (fall height, impact velocity, and child body mass index) were significantly related to injury severity outcomes. With a greater sample size, additional relationships between biomechanical measures and injury severity could emerge. Additionally, fall scene investigations were not possible for all cases. Therefore, results were dependent upon estimates of fall height in some cases. A larger sample size could also allow for a multifactor analysis in which interactions between fall variables could be investigated.

Fall experiments with the anthropomorphic surrogate expanded upon results from the case-based study because fall environment and surrogate parameters could be controlled. Additionally, biomechanical measures relating to injury potential (e.g. acceleration) were obtained. Therefore, specific relationships between parameters and injury outcome measures could be investigated. However, these experiments were limited by surrogate biofidelity. Further development of a more biofidelic

anthropomorphic surrogate is needed to improve accuracy in results. A few studies have developed child surrogates using skull and neck properties obtained from pediatric cadaver specimens.⁶⁵⁻⁶⁶ However, more information is needed to development surrogates with full-body biofidelity. For example, pediatric joint properties should be investigated as they would likely affect fall dynamics.

Results of the case-based assessments suggested that BMI plays a significant role in injury potential. Children with more severe injuries tended to have higher BMI values. The extra soft tissue in children with higher BMI values likely has a cushioning or protective effect. Results of the parametric sensitivity analysis indicated that soft tissue stiffness has a very small or negligible effect on head and neck injury risk. However, most of the moderate and serious injuries in the case-based assessments were extremity fractures which were not investigated in the computer model. Future studies should further investigate the role of soft tissue in pediatric injury potential. Anthropomorphic surrogates with more realistic soft tissue properties should be developed. Additionally, the computer model of a pediatric bed fall should be expanded to include investigation of extremity injury potential. This could be accomplished through validation of extremity loads in the model by comparing results to those obtained experimentally with the CRABI ATD.

In addition to surrogate biofidelity, the assessments of injury potential in the fall experiments are limited by the injury criteria used in comparisons. Much of the published pediatric injury thresholds are scaled from adult or primate data. In particular, neck injury thresholds and fracture thresholds for the extremities are questionable due to limited information on material properties of the pediatric neck and long bones. Further

work is needed to obtain more accurate pediatric injury criteria. Due to the rare availability of pediatric tissue specimens, studies should focus on the use of animal models and computer modeling techniques to better understand age-related changes in pediatric tissue structure and properties.

Results of the parametric sensitivity analysis suggested that head acceleration alone may be insufficient in predictions of head injury potential in impacts. Thus, more accurate head injury criteria are needed. Incorporation of impact force, impact energy, and the head mass into head injury models should be considered.

It should also be noted that in the fall experiments, only one initial position was simulated. Changes in initial position may affect fall dynamics and subsequently, injury potential. Simulations of additional positions using both surrogate experiments and computer modeling should be conducted for comparisons of fall dynamics and injury potential. In this study, the ATD was initially positioned on its side causing the ATD to also land primarily on its side. Simulations with the ATD initially positioned in a prone or supine position should be investigated to better understand the sensitivity of injury outcome measures to impact orientation. Additionally, simulations with the ATD initially seated or standing on the horizontal surface will give further insight into fall dynamics.

A digital motion capture system was used to track fall dynamics in the ATD experiments. However, data was insufficient to allow a detailed quantitative description of fall dynamics (for example, joint angles and positions and segment velocities throughout the fall). Future studies should attempt to collect more accurate data with additional cameras (a five-camera system was used in this study). Quantitative

descriptions of fall dynamics will enable more detailed and more accurate comparisons of dynamics and impact orientations across fall scenarios to further understand relationships between fall dynamics and injury potential.

The computer simulation model contributed important information about the biomechanics of short-distance pediatric falls. In particular, variations in surrogate parameters (that would be difficult to achieve experimentally) were investigated. Although a rigorous validation process was used, validation was only conducted for one fall scenario, and results obtained from deviations from the validated scenario are limited in their accuracy. Additionally, further validation is recommended that includes additional outcome measures (for example, head angular accelerations and extremity loads). This will improve model accuracy and enable investigation of model parameters on injury potential of the extremities.

In this dissertation, several fall and child characteristics relating to injury potential have been identified. This work may serve as first steps toward development of an injury prediction model for short-distance pediatric falls. The injury prediction model could serve as a clinical tool to determine the likelihood of injury associated with a particular fall scenario and thus increase accuracy in diagnoses of abuse or accidental injury. An injury prediction model could also be used in the medico-legal community for more case-specific injury assessments.

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CURRICULUM VITA

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I. PERSONAL

Education

- Exp. 5/11 Ph.D. in Mechanical Engineering, University of Louisville
Dissertation Title – *“Biomechanics and Injury Assessment of Household Falls in Children: Clinical, Anthropomorphic Surrogate, and Computer Simulation Studies”*
Advisor – Gina Bertocci
- 2007 M.Eng. in Mechanical Engineering, University of Louisville
Thesis Title – *“Investigation of Head and Neck Injury Risk Associated with Short-Distance Falls in 12-Month-Old Children”*
Advisor – Gina Bertocci
- 2005 B.S in Mechanical Engineering, University of Louisville

Teaching-Related Education

- 8/09 – 12/09 College Teaching Course, University of Louisville (ELFH 683)
- 8/08 – 5/09 Future Faculty Program, University of Louisville
- 10/08 Celebration of Teaching and Learning, University of Louisville

Experience

8/05 – Present	Graduate Research Assistant Injury Risk Assessment and Prevention Lab, University of Louisville
5/05 – 8/05	Undergraduate Research Co-op Injury Risk Assessment and Prevention Lab, University of Louisville
6/04 – 5/05	Undergraduate Researcher Bioengineering Department, University of Louisville

Non-Academic Positions

1/03 – 1/04	Technology Engineer Co-op GE Consumer Products, Louisville, KY
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Professional Affiliations

2007 – Present	Society of Women Engineers, Member
2005 – Present	Biomedical Engineering Society, Member
2004 – Present	Pi Tau Sigma, Member President, University of Louisville Student Chapter, 2006-2007 Secretary, University of Louisville Student Chapter, 2005-2006

Honors and Awards

2008	Hsing Chuang Award for Excellence in Graduate Study, University of Louisville
2007	ASME - Pi Tau Sigma Award, University of Louisville
2006	University of Louisville Fellowship
2006	J.B. Speed School of Engineering Outstanding Student, University of Louisville
2006	Lewis Streng Award – Highest academic honor for M.Eng. graduate at J.B. Speed School of Engineering, University of Louisville
2005	Alfred T. Chen Award – Merit-based award given to student pursuing M.Eng. degree at J.B. Speed School of Engineering, University of Louisville
2001 – 2006	University of Louisville Provost Hallmark Scholarship
2002 – 2004	Mechanical Engineering Academic Achievement Award, University of Louisville

2002 – 2003

Tau Beta Pi Outstanding Freshman/Sophomore Award, University of Louisville

II. RESEARCH

Publications

Peer Reviewed Journal Articles

- **Thompson A**, Bertocci G. Pediatric Bed Fall Computer Simulation Model Part I: Development and Validation. Submitted to *Computer Methods in Biomechanics and Biomedical Engineering*. Nov 2010.
- **Thompson A**, Bertocci G, Pierce MC, Rice W. Pediatric Short-Distance Household Falls: Biomechanics and Injury Severity. *Accident Analysis and Prevention*. 2011; v 43:143-150.
- **Thompson A**, Bertocci G, Pierce MC. Assessment of Head Injury Risk Associated with Feet-First Free Falls in 12-Month-Old Children using an Anthropomorphic Test Device. *Journal of Trauma*. 2009; v 66:1019-1029.

Conference Proceedings

- **Thompson A**, Bertocci G, Rice W, Pierce MC. Pediatric Short-Distance Household Falls: Biomechanics and Associated Injury Severity. Eleventh International Conference on Shaken Baby Syndrome/Abusive Head Trauma, Atlanta, GA. Sept 2010.
- Rice W, Pierce MC, **Thompson A**, Kim IK, Bertocci G. Investigation of Injury Risk Associated with Common Household Falls in Young Children. National Pediatric Academic Society Conference, St. Louis, MO, Mar 2010.
- **Thompson A**, Bertocci G, Pierce MC, Rice W. Biomechanics of Pediatric Short-Distance Household Falls: Relationships with Injury Severity. *Research Louisville*, Louisville, KY. Oct 2009.
- Rice W, **Thompson A**, Pierce MC, Kim IK, Bertocci G. Investigation of Injury Risk Associated with Common Household Falls in Young Children. *Research Louisville*, Louisville, KY. Oct 2009.
- Sison S, **Knight A**, Bertocci G. Assessing Kinematics of Pediatric Falls using an Anthropomorphic Test Device. *Society of Women Engineers National Conference*, Nashville, TN. Oct 2007. **Collegiate Poster Competition Finalist**.
- **Knight A**, Bertocci G, Pierce MC, Campbell-Kyureghyan N. Head Injury Risk Associated with Falls from Standing in Children and the Influence of Joint Stiffness. *Injury Biomechanics Symposium*, Columbus, OH. May 2007. **Travel Award Recipient**.
- **Knight A**, Bertocci G, Pierce MC. Effect of Joint Stiffness on Head Injury Risk in Pediatric Falls. *University of Louisville Engineering Expo*, Louisville, KY. Mar 2007.
- **Knight A**, Bertocci G, Pierce MC, Bialczak K. Head Injury Risk Associated with Feet-First Free Falls in Children and Influence of Impact Surface Type. *ASME Summer*

Bioengineering Conference, Amelia Island, FL. Jun 2006. **2nd Place in Masters Level Student Paper Competition.**

- Bialczak K, Bertocci G, Pierce MC, **Knight A**. Pediatric Bed Fall Computer Simulation Model Development and Validation. *ASME Summer Bioengineering Conference*, Amelia Island, FL. Jun 2006. **1st Place in Masters Level Student Paper Competition.**
- Bertocci G, Pierce MC, **Knight A**, Bialczak K, Kaczor K, Deemer E. Head Injury Risk Associated with Free Falls from Varying Heights in Children. *Pediatric Academic Society Conference*, San Francisco, CA. May 2006.
- **Knight A**, Bertocci G, Pierce MC, Bialczak K, Deemer E. Head Injury Risk Associated with Free Falls in Children. *Research Louisville*, Louisville, KY. Oct 2005.
- Bialczak K, Bertocci G, **Knight A**, Deemer E, Spivack B. A Child Abuse Investigation: Evaluation of Exersaucer Accelerations Attained During Spinning. *Research Louisville*, Louisville, KY. Oct 2005.
- **Knight A**. Formation of Stable Lipid Bilayers on MEMS Devices. *University of Louisville Engineers Days*, Louisville, KY. Mar 2005.

Invited Presentations

- Thompson A. Using Biomechanics to Aid in the Detection of Child Abuse. HSS 387 Biomechanics, University of Louisville, Louisville, KY. Dec 2009.
- Thompson A. Using Physics Principles to Aid in the Detection of Child Abuse. Summer Medical and Dental Education Program, University of Louisville, Louisville, KY. Jun 2008, Jun 2009 and Jun 2010.
- Knight A, Bertocci G, Pierce MC, Bialczak. Head Injury Risk Associated with Feet-First Free Falls in Children. ASME-ASHRAE Local Chapter Meeting, Louisville, KY. Mar 2006.
- Knight A, Bertocci G, Pierce MC, Bialczak. Head Injury Risk Associated with Feet-First Free Falls in Children. University of Louisville Pediatric Research Luncheon, Louisville, KY. Mar 2006.

Research Experience

- Department of Justice Grant. Oct 2009 – Sept 2012. Development of Scientific and Objective Methods to Detect Physical Child Abuse. Role: Graduate Student Researcher.

Responsibilities:

- a. Conduct pediatric fall experiments using an anthropomorphic test device.
- b. Develop and validate a computer simulation model of a pediatric fall.
- c. Conduct a parametric sensitivity study to determine the effect of various fall environment and child characteristics on injury risk in short falls.

