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Pediatric bed fall computer simulation model: Parametric sensitivity analysis

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ABSTRACT

Falls from beds and other household furniture are common scenarios that may result in injury and may also be 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 bed fall computer simulation model of an anthropomorphic test device representing a 12-month-old child 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, stiffness for other body segments) on fall dynamics and outcomes related to 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) 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, initiating force) than surrogate parameters (head stiffness, neck stiffness, body segment stiffness).

1. Introduction

Falls from beds and other household furniture are common scenarios that may result in injury and may also be stated to conceal child abuse [1–4]. Identification of important factors related to injury potential in short-distance falls may aid clinicians in history-taking and improve assessments of injury and history compatibility when 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 [5–11]. However, many of these studies have been limited by the biofidelity of anthropomorphic surrogates used to represent the fall victim [5,7–11]. Mechanical response requirements for pediatric surrogates are often based on scaled adult cadaver or primate data and may not accurately represent a human child.

Computer simulation modeling is a tool that can be used to investigate injury-producing events, and to study the effect of changing event parameters on injury potential. Within the model, parameters that can be altered include fall environment parameters (such as fall height and impact surface) and child surrogate parameters (such as mass and mechanical properties of joints and tissues) which are difficult to alter experimentally. 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 [12–18]. 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 [19]. The purpose of this study was to use the validated model to investigate the effect of altering fall environment and surrogate parameters on biomechanical measures related to injury potential. This will serve to identify key factors that may increase a child's risk of injury in a given fall scenario.

2. 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) [19]. The model depicts the CRABI in a side-lying initial position on the edge of a horizontal surface 24 in. (61 cm) above the

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

Parameters	Injury outcome measures
Horizontal surface (bed) height	Peak resultant linear head acceleration
Impact surface (floor) stiffness	Peak resultant angular head acceleration
Actuator velocity/force (to initiate fall)	Peak resultant upper neck force
Surrogate mass	Peak resultant upper neck moment
Surrogate skull stiffness	
Surrogate neck stiffness (4 orientations):	
Axial compression	
Flexion/extension bending	
Lateral bending	
Torsional bending	
Surrogate neck damping	
Surrogate body segment stiffness	

ground. In the experiments, a pneumatic actuator was used to push the ATD from the bed surface with a repeatable force. This actuator was replicated in the model. Validation of the model entailed a visual comparison of fall dynamics and quantitative comparison of outcome measure time histories between the model and experimental results. Additionally, the predictive capability of the model was assessed by changing the floor (impact surface) properties and verifying the model outcomes matched experimental results.

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 outcome measures related to injury potential. Fall environment and surrogate parameters were varied in the model, and the sensitivity of injury outcome measures to model parameters was determined.

2.1. Model parameters

Eleven parameters were evaluated (Table 1). Each parameter was varied individually within the model 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 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 identified from the scientific literature 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.

2.1.1. Horizontal surface (bed) height

Height has been shown in biomechanical studies to influence injury risk in pediatric falls [5–7,9–11]. A clinical study of pediatric falls from horizontal surfaces was used to provide a real-world range of fall heights [6]. 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 ±50% and ±25% of the baseline bed height. The baseline surface height in the validated model was 608 mm.

2.1.2. Impact surface (floor) stiffness

Impact surface has been shown in biomechanical studies to influence injury potential in pediatric falls [5,7,9–11]. The surface stiffness in the baseline model was specified to match that of playground foam (206 N/mm) as the stiffness for this material was able to be measured directly. The playground foam is a 2 in. thick stiff rubber made from recycled tires. Surface stiffness was adjusted to ±50% and ±25% of the baseline value for analysis. As an additional reference, the stiffness was adjusted to that for linoleum over a wood subfloor (simulated as part of model validation) which is 867 N/mm [19].

2.1.3. Initial velocity/force

To initiate the fall in both the model and physical experiments with the surrogate [19], an actuator impacted the posterior torso of the surrogate (approximately the center of mass location). The impact velocity of the actuator was measured in the experiments and replicated in the computer simulation. For the parametric analysis, actuator velocity was adjusted to assure that the target force value (actuator contact with ATD) was attained. Since actuator force was directly related to velocity in the model, both force and velocity values were reported. The baseline velocity was 0.52 m/s and baseline force was 140 N. As initial force and velocity are not measurable parameters in most household falls, no information was found to establish a real-world range for simulation.

2.1.4. 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 by changing the mass of each body segment proportionally (i.e. no changes to mass distribution or body segment geometries). In addition to the predetermined incremental mass changes (±50% and ±25% of the baseline value), the 5th (8.3 kg) and 95th (11.9 kg) percentile mass values for a 12-month-old child [20] were evaluated.

2.1.5. Surrogate skull stiffness

The surrogate in the computer model represents the CRABI 12-month-old ATD. Some have questioned the biofidelity of the CRABI head particularly in low-energy impacts such as falls [11,21]. 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. [22] conducted compression tests on three skulls (ages 1–11 days) in two orientations (anterior-posterior compression and lateral compression). Similarly, Loyd measured head stiffness at various loading rates and orientations in compression tests of human pediatric skulls including that of an 11-month-old child [23]. Since the neonate skull stiffness was less than that of the 11-month-old, the mean of the dynamic stiffness curves measured by Prange was used as the lower bound of head stiffness in the parametric analysis. Yoganandan et al. [24] tested six adult skulls in compression under quasi-static loading and dynamic loading (7.1–8.0 m/s). The mean (dynamic values only) of the adult stiffness curves (Yoganandan et al.) was used as an upper bound of head stiffness properties for analysis. Fig. 1 shows the head force-displacement curve used in the validated bed fall model (baseline) compared to experimentally determined cadaver data.

2.1.6. Surrogate neck stiffness

Just as skull stiffness is expected to play a major role in head injury potential, surrogate 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 motor vehicle 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

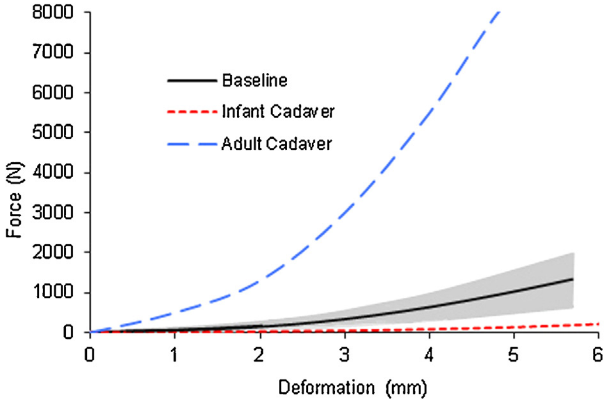

Fig. 1. Head stiffness values for baseline (validated) bed fall model, adult cadaver experimental data [24], and infant cadaver experimental data [22]. Shaded region represents the ±50% of baseline range simulated as a part of the sensitivity analysis.

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, the properties used in the analysis represent a lower bound of neck stiffness.

a. Flexion/extension – Wheeldon et al. [25] reported load–displacement curves for seven healthy adult subjects (Fig. 2). Panjabi et al. [26] and Schwab et al. [27] report similar or lower adult flexion/extension stiffness compared to those by Wheeldon. Therefore, the Wheeldon stiffness properties were used as the upper bound for neck flexion/extension stiffness in the parametric analysis. Ouyang et al. [28] reported load-displacement properties in flexion/extension for ten pediatric cervical spine cadaveric specimens (ages 2–12 years). Data for the youngest specimen (age 2 years) is shown in Fig. 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. [29] 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 (Fig. 2).

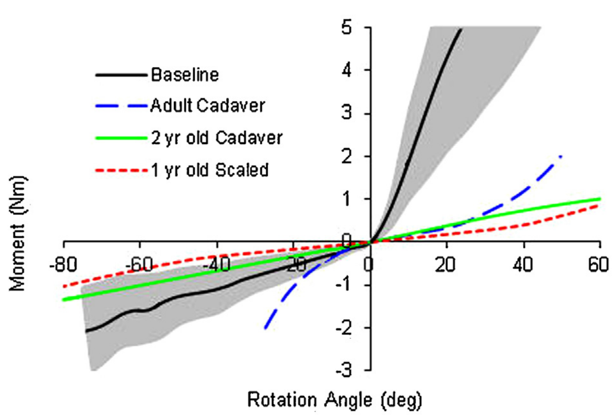

Fig. 2. Neck flexion/extension stiffness properties for baseline (validated) bed fall model and cadaver experimental data for an adult [25], 2 year-old child [28], and scaled results for a 1 year-old child. Positive rotation angles indicate flexion motion and negative angles indicate extension. Shaded region represents the ±50% of baseline range simulated as a part of the sensitivity analysis.

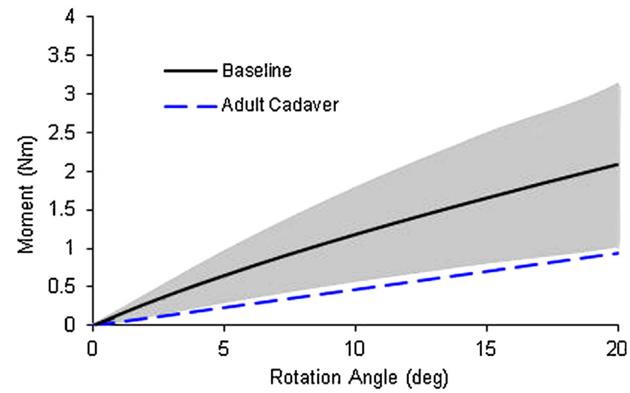

Fig. 3. Neck lateral bending stiffness properties for baseline (validated) bed fall model and adult cadaver experimental data [27]. Shaded region represents the ±50% of baseline range simulated as a part of the sensitivity analysis.

b. Lateral bending – Schwab et al. [27] described lateral stiffness for the adult neck (Fig. 3). No pediatric data or scaling factors were found for lateral stiffness. Therefore, only adult stiffness properties (in addition to the predefined incremental values) were evaluated in the parametric analysis.
c. Torsion – Schwab et al. [27] described stiffness for the adult neck in torsion (Fig. 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.
d. Axial compression – Shea et al. [30] described adult neck stiffness in axial compression (Fig. 5). Additionally, Kumaresan et al. [29] estimated in a finite element study that the neck stiffness of a 1-year-old child is 500% softer than adult neck stiffness 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 (Fig. 5). Both the adult and scaled infant properties were included in the analysis.

2.1.7. Surrogate neck damping coefficient

In the computer simulation model, joint properties are represented by both stiffness and damping coefficient parameters. Unlike neck stiffness properties, damping coefficients for cadaveric neck specimens have not been measured. However, damping

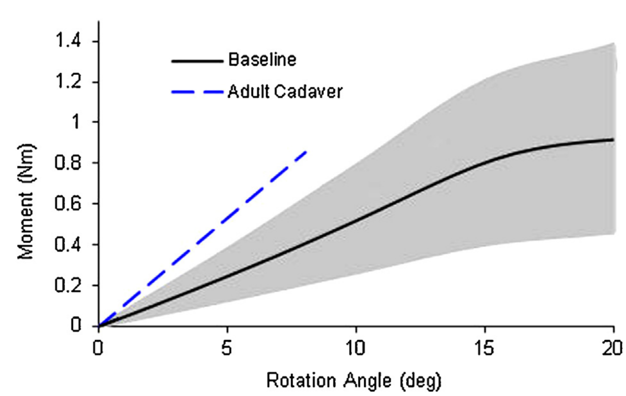

Fig. 4. Neck torsional stiffness properties for baseline (validated) bed fall model and adult cadaver experimental data [27]. Shaded region represents the ±50% of baseline range simulated as a part of the sensitivity analysis.

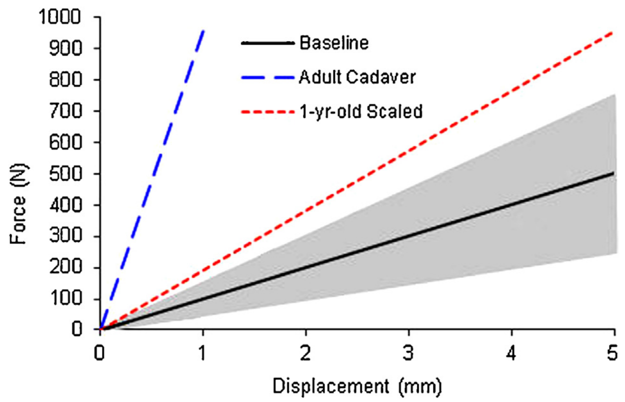

Fig. 5. Neck compression stiffness properties for baseline (validated) bed fall model, adult cadaver experimental data [30], and scaled results for a 1-year-old child. Shaded region represents the ±50% of baseline range simulated as a part of the sensitivity analysis.

properties are an important component in mathematical or computer models to define rate-dependent material behavior. The neck damping coefficient was altered to ±50% and ±25% of the baseline value (0.4 Ns/m). The damping coefficient is uniform for all neck bending orientations within the model.

2.1.8. Surrogate body segment 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. [6] 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 given that the ATD was designed to withstand repeated impact tests and soft tissue injuries were not of interest in this type of testing. To investigate the effect of soft tissue stiffness on injury potential, the stiffness of body segments (other than the head/neck) was altered to ±50% and ±25% of the baseline value (Fig. 6). A few studies have measured soft tissue stiffness of adult subjects using indentation tests [31–33]. However, these tests were done for small skin indentations (displacements < 5 mm). The results of the skin indentation tests were not extrapolated for the parametric analysis because of the non-linear nature of soft tissue stiffness properties.

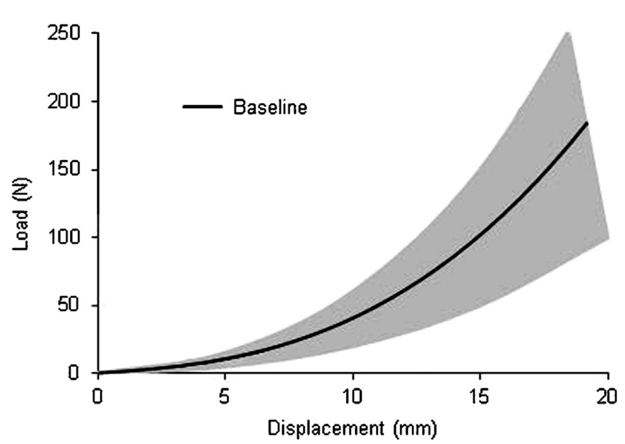

Fig. 6. Body segment stiffness for the baseline (validated) bed fall model. Shaded region represents the ±50% of baseline range simulated as a part of the sensitivity analysis.

2.2. Outcome measures

Changes in fall dynamics due to changing input parameters were qualitatively assessed. Additionally, four outcome measures relating to head and neck injury potential were assessed (Table 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). Outcome measures were filtered according to SAEJ211 standards [34].

2.3. Sensitivity analysis

Sensitivity index 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 given change in a specific 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 indices were calculated for each parameter with each of the 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 index for each parameter was determined and used for parameter sensitivity comparisons.

3. Results

3.1. Fall dynamics

Changes in bed height, the initial velocity/force, and surrogate mass produced considerable changes in fall dynamics. Fig. 7 illustrates variations in the impact orientation of the surrogate with changing height, force, and mass. 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 fall initiation force affected kinematics with which the surrogate left the bed surface. In the baseline model, the surrogate was impacted with a level of force to initiate a 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). This may be due to a greater fraction of surrogate mass and thus greater friction with the bed surface superior to the point of actuator impact. 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 enough force to push the surrogate from the bed surface. Therefore, this simulation was not included in the results.

Surrogate mass changes also affected impact orientation. Simulations with increasing mass resulted in a steeper angle of impact (feet at higher elevation at the moment of impact), and simulations with decreasing mass resulted in a shallower angle of impact (feet

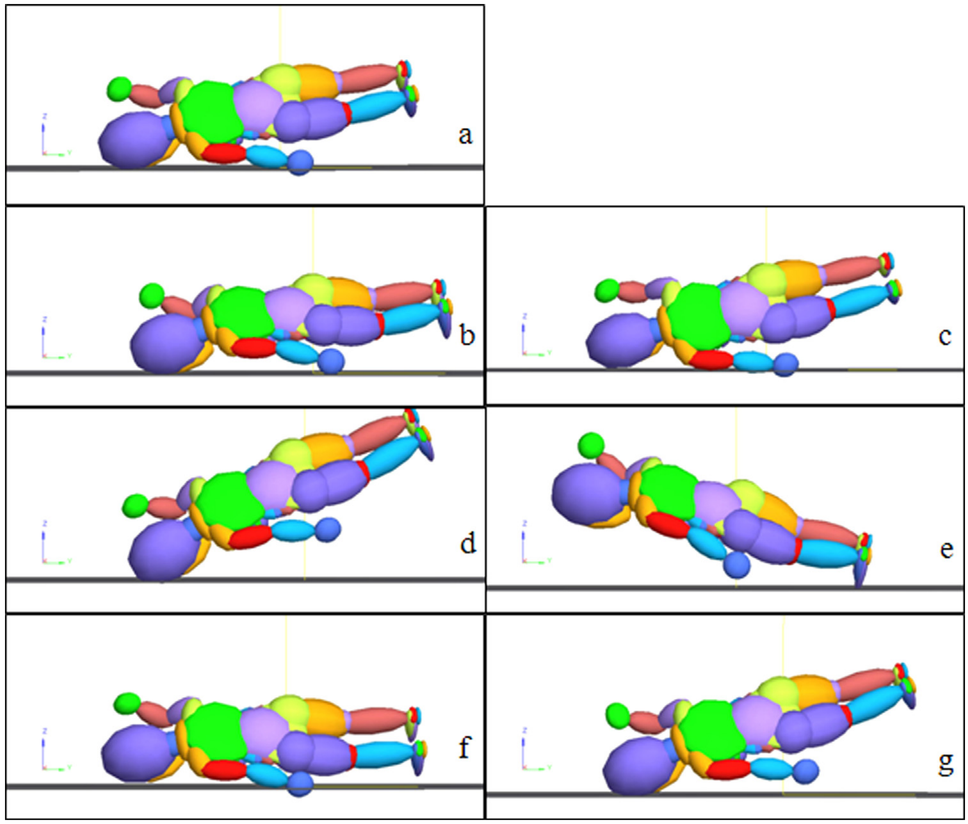

Fig. 7. Orientation of the surrogate upon impact with the floor surface for parameters that substantially altered fall dynamics: (a) baseline (validated) model, (b) model with bed height set at −25% of the baseline, (c) model with bed height set at +25% of the baseline, (d) model with initial velocity/force set at −25% of the baseline, (e) model with initial velocity/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.

relatively lower 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 body stiffness).

3.2. Head injury measures

Peak linear head acceleration values were most sensitive to changes in surrogate mass (Fig. 8). Additionally, there was an inverse relationship between mass and head injury outcome measures (Table 2). Increasing the surrogate's mass resulted in decreasing peak linear accelerations and peak angular head accelerations. Angular head accelerations were most sensitive to actuator velocity/force; increasing the initial force resulted in increasing peak angular head accelerations (Fig. 9). The influence of initial force on linear head accelerations was less pronounced. In all simulations, peak linear and angular head accelerations occurred during head impact with floor surface. Bed fall height, surface stiffness, and surrogate head stiffness had direct relationships with head injury outcome measures. Altering neck properties and body segment stiffness had little influence on head injury outcome measures.

3.3. Neck injury measures

Peak resultant neck force was most sensitive to changes in the initial velocity/force and peak neck moment was most sensitive to neck damping coefficient (Figs. 10 and 11). Unlike the head injury measures, however, initial velocity/force had an inverse

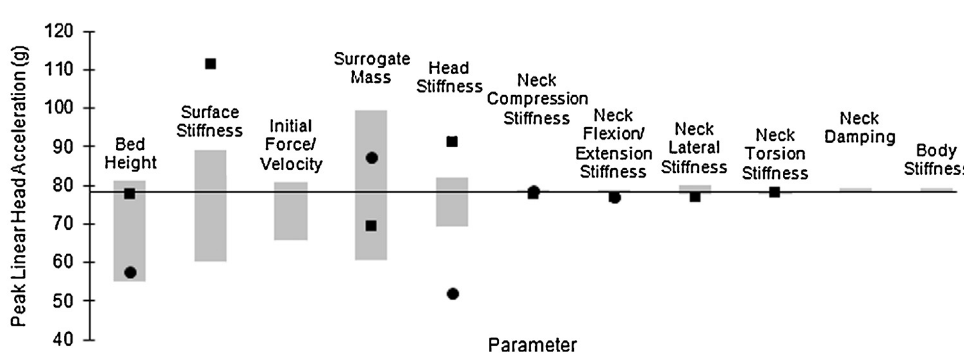

Fig. 8. Peak resultant linear head acceleration for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the outcome value range for parameter values ±50% of the baseline; the square and circle markers indicate the outcome values associated with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

Table 2
Mean sensitivity index for each outcome measure to each model input parameter.

Parameters	Outcome measures			
	Peak resultant head linear acceleration	Peak resultant head angular acceleration	Peak resultant neck force	Peak resultant neck moment
Fall height	0.31	0.49	0.05	−0.16
Surface stiffness	0.36	0.22	0.17	0.07
Actuator velocity/force	0.11	2.83	−0.50	−0.39
Surrogate mass	−0.56	−0.79	0.33	0.28
Head stiffness	0.15	0.09	0.08	0.05
Neck compression stiffness	0.00	0.01	0.20	0.04
Neck flexion/extension stiffness	0.01	0.05	−0.01	0.04
Neck lateral stiffness	0.03	0.10	0.00	0.05
Neck torsion stiffness	0.01	−0.03	0.00	0.02
Neck damping coefficient	−0.02	−0.06	0.13	0.58
Body stiffness	−0.02	−0.03	0.05	0.07

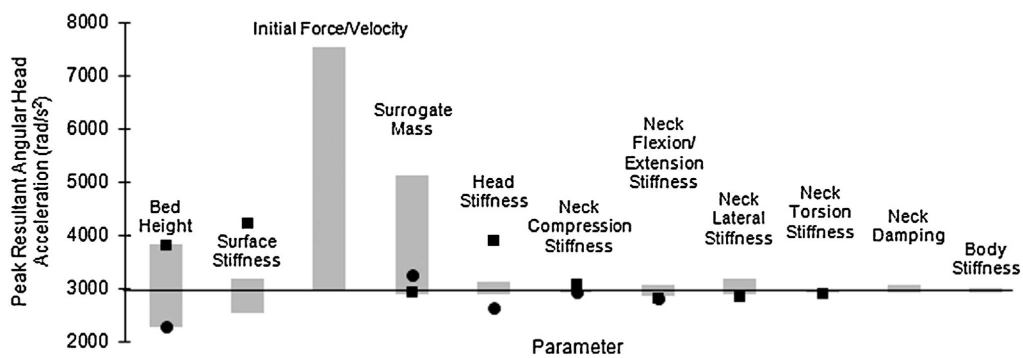

Fig. 9. Peak resultant angular head acceleration for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the outcome value range for parameter values ±50% of the baseline; the square and circle markers indicate the outcome values associated with the maximum and minimum parameter values from the literature, respectively.

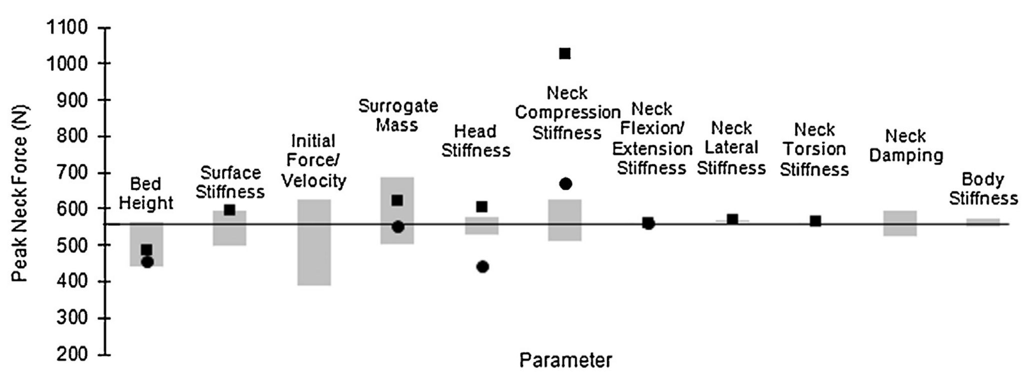

Fig. 10. Peak resultant neck force for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the outcome value range for parameter values ±50% of the baseline; the square and circle markers indicate the outcome values associated with the maximum and minimum parameter values from the literature, respectively.

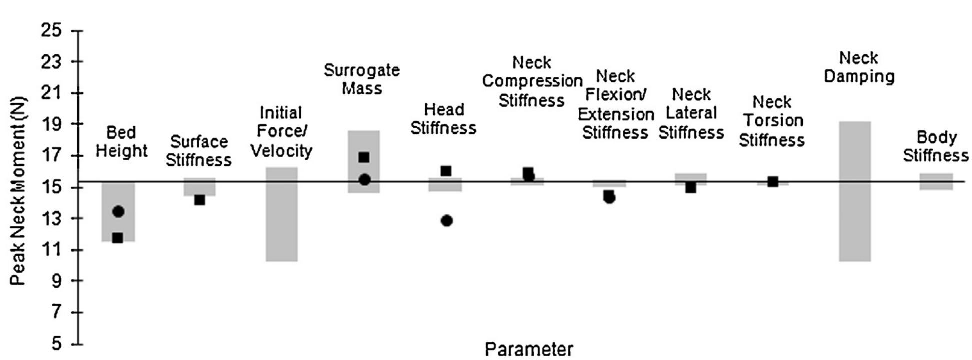

Fig. 11. Peak resultant neck moment for varying input parameter ranges: the horizontal line represents the baseline value; the shaded bar represents the outcome value range for parameter values ±50% of the baseline; the square and circle markers indicate the outcome values associated with the maximum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

relationship with neck forces and moments (Table 2). Surrogate mass had a direct relationship with neck loads. 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. Bed height, surface stiffness, and body segment stiffness also influenced peak neck forces and neck moments.

4. Discussion

4.1. Sensitivity analysis

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 velocity/force) than surrogate parameters (head stiffness, neck stiffness, body segment stiffness). Increasing bed height and increasing surface stiffness led to increases in 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 [5,7–11]. Increasing the actuator velocity/force tended to increase head injury measures, but decrease 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 was transferred through the neck as the left arm and torso impacted the ground sooner. Relationships between the force to initiate the fall and injury potential have not been studied previously. Factors that could increase the initial velocity of the child in an actual fall could include the child being pushed from the surface or the child being active 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 velocity/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 ATD [5]. Thompson et al. found that fall height changes led to differing fall dynamics which were seen to influence head injury outcomes.

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. Two factors contributed to this finding. First, the initial velocity of the surrogate (after contact with the actuator but just prior to the fall) was reduced. 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 steeper angle (feet higher above ground at moment of impact). The impact force was transferred primarily from the head through the neck, and with a steeper impact angle, there were increased neck compression loads and decreased neck bending loads. The head impact duration was also increased (Fig. 12). Larger impact durations are associated with reductions in peak linear and angular head accelerations. This has been demonstrated in previous fall studies [5]. A similar effect occurred in simulations with varying initial velocity/force. In falls with increased initial force, the surrogate landed feet-first so little energy was absorbed through neck compression. However, in falls with a lower initial force, the surrogate landed head-first leading

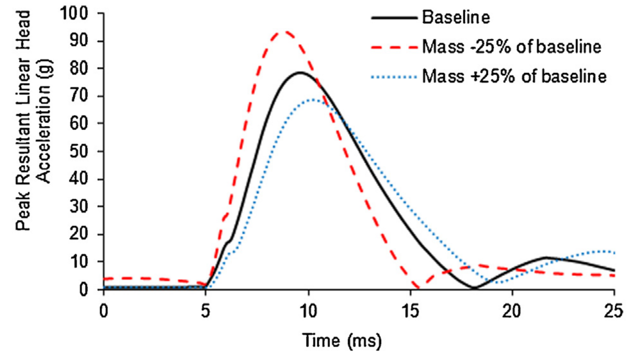

Fig. 12. Peak resultant linear head acceleration time-histories for baseline (validated) bed fall model and simulations with surrogate mass set at ±25% of the baseline value.

to greater neck compression forces, increased head impact durations, and decreased head accelerations (relative to falls in which the surrogate landed feet-first).

Despite decreases in head acceleration measures with increasing surrogate mass, the head contact force increased with increasing mass (Table 3). These results suggest that acceleration alone may not be sufficient for predicting head injury potential in impacts. Higher impact forces suggest an increased risk of head injury (particularly skull fracture) while decreasing peak accelerations are typically associated with a lower risk of head injury. Acceleration measures alone 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 ±50% of the baseline (50th percentile 12-month-old child) mass (see Figs. 8–11). Therefore, the influence of surrogate mass on injury potential may be exaggerated in this study.

Surrogate head stiffness influenced peak linear head accelerations, 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 ±50% of the baseline head stiffness (see Figs. 8–11).

Neck parameters, with the exception of axial compression stiffness and neck damping coefficient, and body segment 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 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 neck damping coefficient led to increases in 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

Table 3
Peak resultant head impact force versus surrogate mass.

Surrogate mass (kg)	Peak resultant head impact force (N)
4.9 (−50% of baseline)	1800
7.4 (−25% of baseline)	2406
9.9 (baseline)	2771
12.3 (+25% of baseline)	3131
14.8 (+50% of baseline)	3449

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 [25–28,30], however, only neck stiffnesses are measured. Future work investigating rate-dependent neck properties is needed to improve accuracy in modeling neck properties.

A few studies have investigated the effect of fall parameters on injury risk using computer simulation [12,14,18]. 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. [12] simulated four falls using MADYMO® (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 Gilchrist [14] 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.

4.2. Injury potential

Head accelerations and neck loads can be compared to injury thresholds for assessment of injury potential. In general, higher accelerations and loads indicate a greater risk of injury. Several studies have reported head injury thresholds based on peak linear and angular head acceleration (Table 4). Neck injury thresholds are more scarce. The National Highway Traffic Safety Administration has established Neck Injury Criteria (N_{ij} values) for use with the anthropomorphic dummies (including the CRABI) [35]. However, these criteria are based on combined axial and rotational loading in the sagittal plane. In this study, the primary neck loading was lateral bending. Therefore, our results should not be compared to N_{ij} thresholds. A single study of lateral neck injury in adult cadavers found that specimens could withstand up to 75 Nm lateral bending moment combined with low axial loads (less than 300 N) without injury [36].

4.3. Limitations

Due to limitations of the computer model, the results should not be used to make absolute predictions of injury occurrence in pediatric falls. Rather, relationships between model parameters and injury outcome measures 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. It should also be noted that computer models are discrete representations of real-world events, and thus our model may lack accuracy in its depiction of the CRABI in a bed fall. As an example, floor surface properties were represented in the model by a linear stiffness parameter. During the model validation process, a linear stiffness was found to be sufficient in predicting head and neck outcome measures for the simulated fall scenario. Other surfaces, such as carpet, may not be accurately represented by a linear stiffness. Future work is needed to determine if changing floor damping or other nonlinear properties significantly affect outcome measures. Additionally, it should be noted that joint properties 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, and other body segments may influence injury outcome measures but were not investigated in this study. Results of changing surrogate mass are limited in that they did not include any changes in anthropometrics, overall size or mass distribution. Future simulations should investigate changes in the anthropometrics of the child along with variations in overall mass. 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. This study found that fall dynamics and impact orientation played a significant role in outcome measures related to injury potential. Thus, the results of this study regarding sensitivity to varying surrogate

Table 4
Head injury thresholds reported in the literature.

Injury threshold	Description	Reference
Linear head acceleration		
51 g	Tolerance for CRABI 12-month-old ATD representing 5% risk of significant head injury	Nahum and Melvin, 2002 [37]
83 g	Tolerance for AIS2+ injury for 6–7 yr-old children based on reconstructions of pedestrian accidents	Sturtz 1980 [38]
200–250 g	Tolerance for children (age not specified) based on reconstructions of free falls	Mohan et al., 1979 [18]
46–128 g	Tolerance for skull fracture using 6-month-old CRABI ATD	Klinich et al. [39]; Van Ee et al. [40]
Angular head acceleration		
6500/10,000 rad/s^2	Tolerance for concussion in young child/infant (800/400 gm brain mass, respectively)	Ommaya et al. [41]
18,000/30,000 rad/s^2	Tolerance for mild diffuse axonal brain injury in young child/infant (800/400 gm brain mass, respectively)	Ommaya et al. [41]
10,000 rad/s^2	Tolerance for subdural hematoma in adult (impact durations < 10 ms)	Depreitere et al. [42]

AIS: abbreviated injury scale, ATD: anthropomorphic test device.

and fall environment parameters are applicable to the simulated fall scenario only (rolling from an elevated horizontal surface), and any significant deviations from this scenario (for example, feet-first falls or falls from a seated position) require further validation of the model for those scenarios.

5. 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 outcome measures. 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) 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, initiating force) than surrogate parameters (head stiffness, neck stiffness, body segment 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, especially in comparison to fall environment parameters.

Competing interests

None declared.

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Ethical approval

Not required.

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