Medical Engineering & Physics 36 (2014) 110–118

Contents lists available at ScienceDirect

Medical Engineering & Physics

journal homepage: www.elsevier.com/locate/medengphy

Pediatric bed fall computer simulation model: Parametric sensitivity analysis

Angela Thompson[a], Gina Bertocci[a,b,c,*]

[a] Department of Bioengineering, University of Louisville, Louisville, KY, USA
[b] Mechanical Engineering, University of Louisville, Louisville, KY, USA
[c] Pediatrics, University of Louisville, Louisville, KY, USA

ARTICLE INFO

Article history:
Received 6 April 2012
Received in revised form
20 September 2013
Accepted 11 October 2013

Keywords:
Pediatric falls
Biomechanics
Computer simulation
Injury assessment

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

* Corresponding author at: Department of Bioengineering, University of Louisville, 204 Research Tower, 500 S Preston Street, Louisville, KY 40202, USA. Tel.: +1 502 852 0296; fax: +1 502 216 7859.
E-mail address: g.bertocci@louisville.edu (G. Bertocci).

1350-4533/$ – see front matter © 2013 IPEM. Published by Elsevier Ltd. All rights reserved.
http://dx.doi.org/10.1016/j.medengphy.2013.10.006

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