JRRD Volume 45, Number 8, 2008
Pages 1227–1236
Journal of Rehabilitation Research & Development

Estimation, simulation, and experimentation of a fall from bed

Brian W. Schulz, PhD;[1*] William E. Lee III, PhD;[2] John D. Lloyd, PhD[1]
[1]Rehabilitation Research and Development Research Enhancement Award Program, James A. Haley Veterans' Hospital, Tampa, FL; [2]College of Engineering, University of South Florida, Tampa, FL

Abstract—Computer simulations using multibody models have been extensively applied to vehicular crash testing but have rarely been used to investigate falls. This article investigated planar and three-dimensional simulations of a single physical test of a Hybrid III anthropomorphic test dummy falling from a bed and compared them with a common estimation method. The effects of initial model position and velocity on simulated peak resultant head deceleration and head impact criterion (HIC) were determined while all contact and model parameters were held constant. Improving body position at impact and impact velocity direction both improved results. Simulating the entire fall instead of only the impact further improved simulation output, but HIC was consistently overestimated because of inaccurate contact parameters. These results show that accurate kinematics are crucial to accurate simulation output but improved contact parameters and thorough validation of experimental data are required before any fall simulation should be used to extrapolate findings beyond what is experimentally practical or possible.

Key words: accelerometer, anthropomorphic test dummy, bed, biomechanics, deceleration, fall, head impact, head injury, LifeMOD, simulation.

INTRODUCTION

The majority of injuries to older adults result from falls [1–2], and an estimated one-third of adults aged 65 and over fall one or more times each year [3]. Those living in institutions fall at about 4.5 times this rate (1.5 falls per bed per year [4]), and about 15 percent of falls result in serious injury [4].

Although the majority of hip factures in the elderly (89% in men and 93% in women) resulted from falls from a standing height, falls from bed or in the bedroom accounted for 36 percent of 380 falls experienced in the home [5] and 46 percent of 865 falls experienced in an institutional setting [6]. Forty percent of the institutional bedroom falls were classified as a fall from bed, and the head was the most frequent body part injured (41% of 238 injuries) [6].

These and many other studies have determined who is at risk of a fall, what circumstances predicated the fall, and what injuries resulted from the fall [7], but less is known about the characteristics of the fall itself. Fall direction [8–10] and impact site [9,11–13] are important factors to consider that have been studied [14], but the self-reported nature of these data has recently been shown to be unreliable [15] and may not be valid. Recent recordings of actual falls from a standing position by young, nondisabled subjects in a laboratory setting [15–16] provide definitive fall kinematics and impact sites but not actual impact forces or accelerations because these human subjects must land on thick mats to prevent injury. Data from such studies may be merged with computer simulation techniques to calculate

Abbreviations: 2-D = two-dimensional, 3-D = three-dimensional, ATD = anthropomorphic test dummy, HIC = head impact criterion, VA = Department of Veterans Affairs.

***Address all correspondence to Brian W. Schulz, PhD; Biomechanics Researcher, Research Service, Rehabilitation Research and Development Research Enhancement Award Program, 8900 Grand Oaks Circle, Tampa, FL 33637; 813-558-3944; fax: 813-558-3990. Email: Brian.Schulz@va.gov**
DOI: 10.1682/JRRD.2007.06.0092

the impact forces and injury risk these subjects would have sustained were they to have fallen on a hard surface. These simulations could then be manipulated to evaluate potential effects of aging by increasing reaction times and reducing the maximum strength and speed of the simulated faller or to evaluate the injury reduction potential of energy absorbing flooring materials.

Accurate fall simulations would thus enable the evaluation of external protective mechanisms (e.g., hip protectors, floor mats, and break-away flooring) and internal protective mechanisms (e.g., faster reactions times and strengthened muscles or bones) without the need for physical testing of anthropomorphic test dummies (ATDs) or human subjects with their inherent intertrial variability and limitations. Fall simulation holds the potential to identify the most effective measure or combination of measures to reduce or eliminate injuries due to falls, but any simulation must be validated to some gold standard (generally experimental data) before its output can be used with confidence for any clinical or scientific purpose.

While computer simulations of falls have been used to reconstruct real-life head injury accidents [17], these simulations were based on injury and eye-witness estimates that may not be reliable. In contrast, this study takes an initial step toward accurate fall simulations by validating simulation output against experimental data of a Hybrid III ATD (Denton ATD, Inc; Milan, Ohio) falling from a bed onto its forehead. The data from a single physical test of this ATD will be used to establish the effects of increasing simulation complexity and biofidelity.

METHODS

A single physical test trial of a Hybrid III ATD falling from a bed onto its forehead was used as the gold standard of comparison. Impact severity of this physical test was estimated using the following methods, listed here in order of increasing complexity and biofidelity:

1. Estimation from equations of basic Newtonian physics.
2. Two-dimensional (2-D) simulation of virtual Hybrid III ATD.
3. Three-dimensional (3-D) simulation of virtual Hybrid III ATD.

The first method, while not a simulation per se, was included here as a point of reference because it is commonly performed as a quick “back of the envelope” calculation to estimate fall severity [18–19]. The second method was a simulation that was restricted to the sagittal plane, while the physical test involved substantial motion out of the sagittal plane. The third method was a full 3-D simulation that was used to investigate the effects of model posture and velocity.

Many variables that are predictive of a wide range of injuries may be compared among these methods. This analysis was restricted to the head because, in general, the head is sensitive to injury and in the specific fall scenario evaluated here—a fall from bed—initial ground contact was made with the right temporal aspect of the head. We also restricted our analysis to only two variables: peak resultant head deceleration and the head injury criterion (HIC) [20]. HIC is the most widely used head injury severity index and was calculated using the following formula:

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

In the above formula, $a(t)$ is the resultant head acceleration (measured in g) and t_1 and t_2 are the initial and final times (in seconds) of the interval during which the HIC attains a maximum value. The current guidelines from the National Highway Traffic Safety Administration limit the maximum time interval between t_1 and t_2 to 15 ms [21]. To determine the HIC value, we evaluated the portion of **Equation (1)** within the “{}” brackets for all possible values of t_2 (i.e., from t_1 + 0.083 ms to t_1 + 15 ms) as well as for all possible values of t_1 for the equation estimation, all simulations, and the physical test using a custom MATLAB script (The MathWorks, Inc; Natick, Massachusetts). The maximum values across this range for the estimation, simulations, and test were reported as the HIC values. Peak resultant head deceleration was also used as an outcome variable because it is related to HIC and is used for playground safety guidelines [22], but it does not similarly incorporate impact duration and was thus more straightforward to interpret.

Instrumentation

A Hybrid III pedestrian ATD was used for the physical test. This ATD is 1.7 m tall and 77 kg in mass. Joint properties and body segment properties of this ATD have been shown to be comparable to human data in automotive occupant restraint testing [23]. While the ATD is a passive device that cannot react to a fall like a living human, it does effectively model a patient who falls from a bed with little to no awareness of the imminent fall and subsequent impact event. Head accelerations were recorded using a triaxial accelerometer (PCB 356A02, PCB Piezotronics; Depew, New York) and integrated with Vicon MX system hardware and workstation software (Vicon; Los Angeles, California). Resultant head deceleration was calculated in multiples of the acceleration of gravity ($g = 9.81\ \text{m/s}^2$) and used to calculate HIC according to **Equation (1)**. Since no motion capture equipment was used to determine accelerometer orientation, we assumed that the majority of the resultant head deceleration due to floor impact was in the opposite direction of gravity, and the acceleration of gravity was subtracted from the resultant head deceleration value. All physical and simulated data were collected or computed at 12 kHz and filtered using a 4th-order low-pass Butterworth filter with a cutoff frequency of 1,650 Hz using MATLAB in accordance with automotive industry standards [24].

LifeMOD/BodySIM 2007 (beta version 4, LifeModeler, Inc; San Clemente, California, http://lifemodeler.com) was used for all simulations. LifeMOD is a human modeling plug-in to Adams (version 2005r2, MSC Software Corporation; Santa Ana, California, http://www.mscsoftware.com/products/adams.cfm), which is a family of motion simulation software. Within LifeMOD, the GeBod database option was used to create a virtual ATD that matched our physical ATD in height and weight. The body segment lengths and masses and joint properties used by this model were based on the studies compiled by Backaitis and Mertz [23]. The only modification made to the default parameters of this virtual ATD was alteration of the head ellipsoid shape scaling factors (to $x = 17$ cm, $y = 16$ cm, and $z = 24$ cm) to more closely match the caliper-measured shape of the physical ATD headform at the point of impact. The default units of meters, kilograms, and seconds were used. The solid-solid contact algorithm was used with the dummy ellipsoids because contacts with the dummy solids do not yet work properly in LifeMOD. The contact parameters used were determined via manual iteration such that accelerometer data of a drop of the detached Hybrid III headform from waist height onto the floor in approximately the same orientation and impact site as the simulation trial most closely matched that of a simulation of this drop created in Adams (**Figure 1**). Friction coefficients were set to 1 as per data from an MC3A force transducer (Advanced Mechanical Technologies, Inc; Watertown, Massachusetts) that was used to record trials of the headform hanging (to determine normal force) and dragging across the floor (to determine friction force).

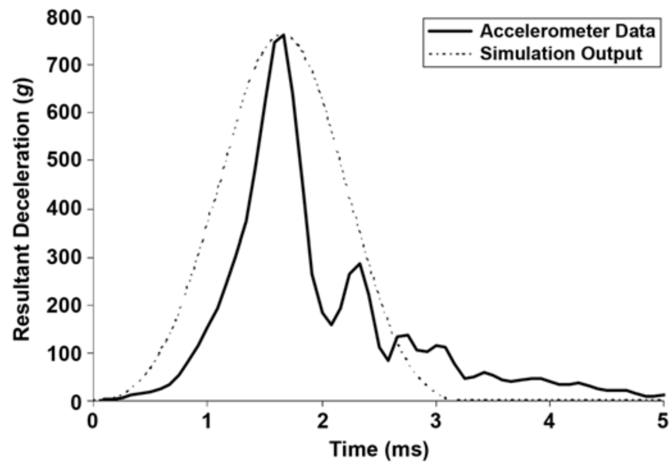

Figure 1.
Accelerometer data and simulation output for calibration trial of waist-high Hybrid III anthropomorphic test dummy headform drop used to determine contact parameters. Simulation contact parameters were stiffness = 1×10^{11}, stiffness exponent = 3, damping = 250, full damping depth = 0.005, and friction coefficients = 1.

Protocol

Physical Testing of ATD Fall

The ATD was rolled out of a typical hospital bed at its maximum height a single time. The distance from the lowest point on the ATD head to the floor, while it lay supine on the mattress, was 0.94 m. The data from this trial were used as input parameters for equation estimation, as initial conditions for all simulations, and as the gold standard against which these estimations and simulations would be measured.

Equation Estimation of Average Deceleration

Microsoft Excel 2003 (Microsoft Corp; Redmond, Washington) was used to calculate the average deceleration of the ATD assuming that the ATD was a single nonrotating lumped mass that was slowed by a constant ground reaction force (constant deceleration). Since the stopping distance was not known without experimental data, the deceleration was plotted as a function of possible stopping distance. The derivation of this formula is shown here. From basic physics, the kinetic energy of a nonrotating falling body is equal to the work performed to bring the falling ATD to rest [19]:

$$\text{Kinetic Energy} = 1/2mv^2 \ , \qquad \textbf{(2)}$$

where m was the mass of the body and v was the velocity of the body.

$$\text{Work Performed} = Fd \ , \tag{3}$$

where F was the average impact force and d the stopping distance.

$$\text{Since } F = ma \ , \tag{4}$$

where a is average acceleration (or deceleration) due to impact, and **Equations (2)**, **(3)**, and **(4)** can be written as

$$1/2v^2 = ad \ . \tag{5}$$

Thus,

$$a = \frac{v^2_{\text{impact}}}{2d} \ , \tag{6}$$

where v_{impact} is velocity at impact.

The impact velocity required for **Equation (6)** was determined by three distinct methods: (1) estimated from basic physics, (2) calculated from accelerometer data, and (3) determined from simulation output of the entire fall. Thus, **Equation (6)** allows the relationship between average deceleration due to impact to be plotted against stopping distance for these three impact velocities.

The estimated impact velocity from basic physics (method 1) would result from a free-fall instantaneously starting at bed height and neglecting air resistance. As such, this is a simplification of the physical test and the direction of the impact velocity must be assumed to be directly downward. The estimated impact velocity resulting from the fall height (h_{fall} = 0.94 m) evaluated here is calculated as follows:

$$\text{Estimated Impact Velocity} = \sqrt{2gh_{\text{fall}}} = \sqrt{2 \times 9.81 \text{ m/s}^2} = 4.29 \text{ m/s} \quad \text{downward [19]} \ . \tag{7}$$

For method 2, the accelerometer data from the physical test were used to calculate the resultant head velocity at impact (4.20 m/s) by integrating the resultant accelerations using MATLAB. These calculated velocity data were then integrated to determine the displacement. The stopping distance and time were determined to be the minimum displacement of the head accelerometer after ground impact and the duration required to reach it. This measured impact velocity was used for the simulations, and the measured stopping distance and time were used to assess the validity of the equation estimation. Note that this measured impact velocity is a resultant value (a scalar). The exact direction of the velocity vector can only be determined from data not collected here; thus, we initially assumed the direction of the impact velocity to be directly downward—i.e., in the same direction as the gravitational acceleration—and later altered this assumption as indicated by the simulation output. The third method of determining impact velocity was to extract resultant head velocity from the simulation of the entire fall at the instant of ground impact.

2-D Simulation

LifeMOD was used to simulate a virtual ATD impacting the floor in the sagittal plane using the "contacts optimized" integrator settings. The direction of the impact velocity vector was assumed to be directly downward, and its magnitude was determined via integration of the resultant accelerometer data. All joints were in the neutral position, and the orientation of the dummy was 40° with respect to the floor to roughly correspond to the impact event shown in **Figure 2**. While this model is technically 3-D, it is effectively 2-D because all motions and forces are restricted to the sagittal plane. The 2-D simulation was of the impact event alone because this simulation was not intended to replicate the exact circumstances of the fall.

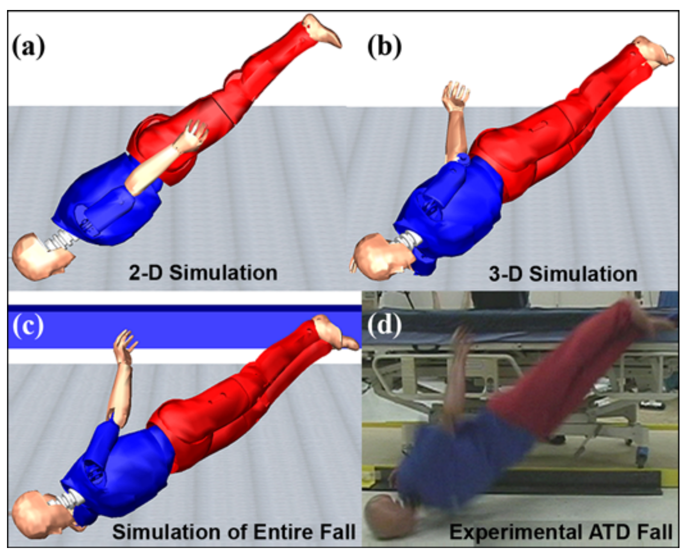

Figure 2.
Simulation of **(a)** two-dimensional (2-D), **(b)** three-dimensional (3-D), and **(c)** entire fall. **(d)** Experimental impact event. Blue rectangle in **(c)** is mattress off of which virtual anthropomorphic test dummy (ATD) was rolled.

3-D Simulation

LifeMOD was used to simulate a virtual ATD impacting the floor in as similar a manner as possible given the available data. The virtual ATD used in the 2-D simulation was further rotated 30º about its longitudinal axis, and all joint positions were approximated from video data of the impact event shown in **Figure 2**. Simulations of the impact event alone and the entire fall were evaluated. Initial conditions for the physical test and all simulations are shown in **Table 1**. Animations and further details of all simulations are in the **Appendix**, which is available online at www.rehab.research.va.gov/jour/08/45/8/pdf/contents.pdf or from the corresponding author upon request.

RESULTS

Resultant head accelerations with respect to time for all three methods of determining impact velocity are shown in **Figure 3**. Peak accelerations and HIC values from these curves are summarized in **Table 2**, along with percentage of deviation from accelerometer-measured values. Note that the equation estimation has no specific peak but is a constant value from the instant of impact until the head is no longer moving downward. When the curves for the initial 2-D and 3-D simulations (indicated by "linear") proved to be higher than the measured data, we reassessed the assumption that the impact velocity vector was directed downward. Upon reexamination of the video footage of the impact event, we determined that the body tended to rotate about a point near the ankle and did not fall directly downward. To account for this, we ran the 3-D simulation again with the impact velocity applied as a rotation about a point near the right ankle (indicated by "rotating"). In an attempt to further improve the simulation results, we simulated the entire fall event rather than the impact event alone (indicated by "entire fall"). This required substantial iteration of the starting position and initial ATD velocities in order to

Table 1.
Initial conditions of physical test and all simulations.

Test or Simulation	Initial Position	Initial Velocity
ATD Test	Supine on bed.	ATD rolled from bed by experimenter manipulating hips and shoulders.
2-D (linear)	40° head downward from prone; arms at sides.	4.20 m/s downward.
3-D (linear)	As in 2-D, but additional 30° rotation about left to right axis in **Figure 2**.	4.20 m/s downward.
3-D (rotating)	As in 3-D (linear).	155 °/s counter-clockwise rotation about point near right ankle in the viewing plane of **Figure 2**.*
3-D (entire fall)	On right side at edge of bed, similar to ATD position just prior to fall.	150 °/s longitudinal rotation about edge of bed and 50 °/s anterior rotation about point near right ankle.†

*Rotation point selected to approximate rotation point of physical test. Angular velocity calculated by applying 4.20 m/s resultant head velocity as tangent velocity about rotation point. This resulted in downward component of velocity vector of 3.44 m/s.
†Determined via manual iteration to match kinematics of fall. Resultant impact velocity was 4.00 m/s, and downward component of impact velocity was 3.86 m/s.
2-D = two-dimensional, 3-D = three-dimensional, ATD = anthropomorphic test dummy.

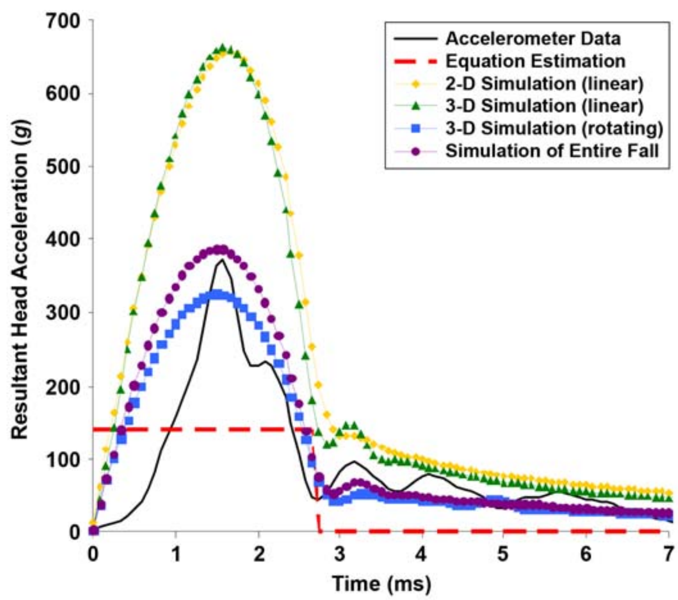

Figure 3.
Resultant head accelerations plotted against time. Accelerometer data are from single physical test: equation estimation (red dashed line) shows average deceleration from **Equation (2)** using impact velocity calculated from fall height using **Equation (7)** (4.29 m/s), 2-D (yellow symbol/line) indicates simulation that is symmetrical about sagittal plane, 3-D indicates virtual ATD positioned to replicate pose of physical ATD at impact, linear (green symbol/line) indicates impact velocity applied directly downward, rotating (blue symbol/line) indicates that direction of resultant impact velocity is acting to pivot body about ankles, simulation of entire fall (purple symbol/line) indicates that entire fall, rather than just instant of impact, was simulated. 2-D = two dimensional, 3-D = three-dimensional, ATD = anthropomorphic test dummy.

Table 2.
Results by physical test or simulation.

Parameter	Peak Resultant Head Deceleration		Head Impact Criterion	
	Value (g)	Actual (%)	Value	Actual (%)
Accelerometer Data from Physical Test	373	100	1,324	100
Equation Estimation	139	37	529	40
2-D Simulation (linear)	407	109	4,435	335
3-D Simulation (linear)	392	105	3,723	284
3-D Simulation (rotating)	282	75	11,570	119
3-D Simulation (entire fall)	371	99	33,015	228

2-D = two-dimensional, 3-D = three-dimensional.

approximate the dynamics of the fall and increased processing time from 16 to 132 seconds, but the peak resultant head deceleration proved to be closest to the actual value (**Table 1**). However, the timing of the peak acceleration, the shape of the deceleration-time curve, and the HIC value remained substantially different from the actual values (**Figure 3**).

EQUATION ESTIMATION OF AVERAGE DECELERATION

Estimations of average deceleration using **Equation (6)** are shown in **Figure 4**. Curves are shown for resultant impact velocities as calculated from fall height using **Equation (7)**, from numerical integration of resultant accelerations during the physical testing, and from the simulation of the entire fall. The peak deceleration from the experimental data is also shown for reference (thin horizontal dashed line), and the average deceleration values predicted by the equation estimation using **Equation (6)** for the stopping distance calculated from numerically double-integrating the accelerometer data are also indicated. Note that the stopping distance of the center of mass of the head may include rotation of the head and compression of both the floor and the head itself.

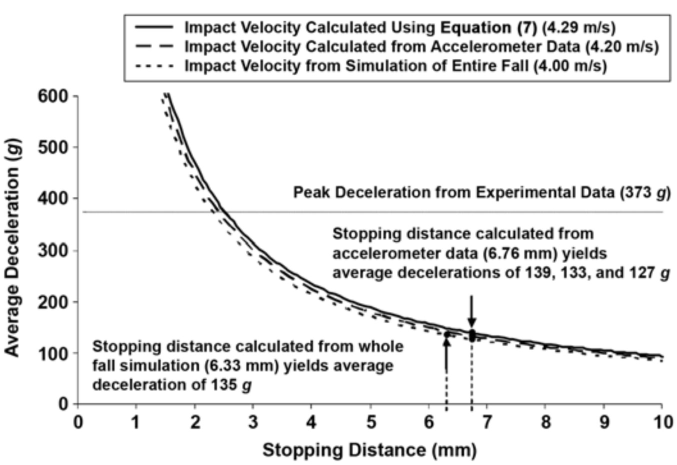

Figure 4.
Estimations of average deceleration calculated from **Equation (6)** using estimated, experimental, and simulated impact velocities. Peak accelerometer deceleration and stopping distances calculated from accelerometer data and from simulation output shown for reference.

DISCUSSION

The head-first impact explored here was initially chosen to simplify the simulation and analysis, but later anecdotal data indicated that injuries caused by falls from a bed do occur at the impact site reported here. While we could find no documented data of impact location or order in falls of human patients from a bed, recent data of multiple Hybrid III falls from a bed collected in our laboratory [25] demonstrated that the head would be likely to sustain impacts of similar magnitudes even if the pelvis or feet impact the ground first. These data also indicated that physical tests of ATD falls in different scenarios result in considerable intertrial variability due to slight differences in initial conditions that are difficult to control in a realistic fall scenario. Computer simulations were explored here in an attempt to eliminate this source of variability while retaining the realism of the fall scenario and enhancing the data accuracy and versatility.

The National Highway Traffic Safety Administration has adopted a maximum allowable HIC of 700 over a maximum time span of 15 ms for all adult ATDs in automobile collisions [21]. Consumer Product Safety Commission playground guidelines limit peak head decelerations to 200 g [22]. The HIC and peak resultant head deceleration from the accelerometer data in this study (1,324 and 373 g, respectively) are well beyond these limits, while the HIC and peak head deceleration from the equation estimation are well below them (529 and 139 *g*, respectively). Such considerable underestimations can be explained by the assumptions (e.g., single deceleration value, single-body system) required to estimate average impact deceleration from **Equation (6)**. This average (not instantaneous) impact deceleration does not account for the transient nature of the impact event (**Figure 4**). One or more peaks are generally evident during an impact event, and this simplified calculation will always underestimate the actual peak deceleration because the implicit assumption of a constant impact force is invalid. Additionally, **Equation (6)** only applies to a single lumped-mass nonrotating body traveling along a linear path, while the ATD is a multisegment system in which each segment moves relative to the ground and to each other in 3-D space during the fall and at ground impact. For example, the impact forces applied to the head during the test fall caused the head to rotate in relation to the torso and were partially transmitted through and partially damped out by the neck.

The method of estimating impact deceleration using basic physics equations is included here because many investigators use it to determine impact severity in injury causation analysis [18–19]. The demonstrated inaccuracies of this method contraindicate its use, even when stopping distance can be accurately measured or estimated, but additional concerns arise when stopping distance is not known. Here, the stopping distance was determined from accelerometer data and simulation output, but it is often assumed. Moreover, the asymptotic relationship between average deceleration and stopping distance (**Figure 4**) demonstrates that large differences in average deceleration may result from small changes in the assumed stopping distance, particularly when the stopping distance is small. This sensitivity to frequently assumed data creates the potential to manipulate the assumed stopping distance in order to obtain a desired average deceleration. This potential indicates that such estimations should be interpreted with extreme caution.

The small differences in impact velocities calculated from the fall height, accelerometer data, and whole fall simulation did not result in large differences in average deceleration (**Figure 4**). However, the direction of the impact velocity appears to be more important than its magnitude as demonstrated by the differences between the first three simulations (**Figure 3**). Significant alterations in impact position (i.e., from 2-D to 3-D) while maintaining the downward impact velocity slightly decreased the overestimations for peak deceleration, while changes in the direction of the impact velocity vector (while maintaining the 3-D body position) based on observed fall dynamics from video data reduced the simulated peak deceleration from a 5 percent overestimation to a 25 percent underestimation. Thus, errors induced by the rough approximation of ATD segment and joint positions at impact were unlikely to have significantly influenced the results, while errors in the segmental velocities of the ATD were a substantial source of error that should be addressed by recording segmental kinematics of the falls and incorporating them into future simulations.

Simulating the entire fall event further improved the peak resultant head deceleration, but the general shape of the curve remained similar to the other simulations and dissimilar to the actual data. This broader curve shape explains why the HIC value for the simulation of the entire fall was 228 percent of the experimental value and the peak head deceleration was 99.4 percent of the accelerometer data and why the direction of the changes in HIC value mirrored that of the peak resultant deceleration but the actual values were consistently overestimated. Thus, simulating the entire fall most closely replicated fall kinematics and impact decelerations, but more accurate ground contact parameters and possible model properties are needed to further improve the accuracy of the head deceleration and HIC data.

The contact parameters used here were determined by a combination of experimentation and manual iteration only for this specific simulation, yet they were still not ideal. While some kind of optimization algorithm may have improved them, the ideal would be a method to generate contact parameters analytically from known or measured material properties. The contact parameters used by Adams and LifeMOD are the combined properties of the interaction between the two contacting surfaces that are related to, but distinct from, their material properties. If a method to calculate these contact parameters from material properties were developed and validated, the properties of the rubber layer over the aluminum ATD headform that contacts the vinyl tile-covered concrete slab could be used to calculate a single set of contact parameters for this interaction. These parameters could then be entered into various simulations and run with confidence for a variety of fall scenarios. The variable thickness of the rubber layer over different ATD segments could also be taken into account to create segment-specific contact parameters. Ultimately, the properties of actual human body segments obtained from cadaver testing, protective responses from motion capture data, and fall protection devices (e.g., hip protectors, fall protection mats, and break-away flooring) could be used to calculate segment-specific contact properties and run simulations that would be able to predict the injury attenuation abilities of these devices in real-world fall situations.

This is not the first study to use multibody simulations to examine falls. O'Riordain and colleagues reconstructed five actual falls from eyewitness reports and fall site reviews [17] using MADYMO (TASS; Delft, the Netherlands, http://www.tass-safe.com), a multibody simulation package typically used for automotive crash testing. Their results appeared to be more sensitive to changes in head contact characteristics than to changes in initial conditions, but with only the injuries of the fall victims, they could not know which simulations most accurately replicated the impact dynamics. In contrast, this study examined the differences between computer simulations using LifeMOD and a single physical test using an ATD, in which the initial conditions and impact decelerations were known and contact properties were determined specifically for this single trial. These simulations demonstrate the effects of increasing accuracy of fall kinematics on impact decelerations independent of the head contact characteristics.

MADYMO and LifeMOD are both powerful simulation packages with specific advantages and disadvantages. The default ATD models in MADYMO are constructed of more ellipsoids than the default LifeMOD ATD models, which may replicate the contact surfaces of physical ATDs more accurately, although the default models of both packages can be modified. MADYMO has been used extensively by the automotive industry for crash test simulations and has more established models and better validation to experimental crash test data, but LifeMOD was chosen for this research because its models can be driven by motion capture data and actuated to replicate human protective responses to a fall. Additionally, LifeMOD models can incorporate tendons, ligaments, and actuated muscles that wrap across bones as in the real world.

Although we lack the facilities for cadaver testing, we would expect the results of a falling cadaver to deviate from the ATD results because of the differing mechanical properties. The rubber and metal of the ATD are not identical to the flesh and bone of a human. The joints of the ATD approximate human joints in some situations (e.g., in the head-on automobile collisions for which this ATD was originally designed), but the joints of an actual person can have far more or less resistance to rotation. More importantly, human joints can actively rotate in reaction to or in anticipation of a fall. Reactions such as grasping handholds, attenuating the impact energy of the ground with the hands and arms, protecting vital areas, or others have the potential to drastically change the outcome of a fall. These compensatory reactions are highly variable, subject- and situation-specific, difficult to simulate, and can have a far greater effect on the results than any other simulation parameter. Thus, simulation and testing of passive ATDs or virtual humans should be used only as a worst-case estimate of potential injury as might be expected for a completely immobile person.

The Hybrid III pedestrian ATD used here was originally designed to replicate the responses of living humans and cadavers in sled tests that mimic head-on automobile collisions and subsequently adapted to simulate pedestrian impacts. Since falls may not be sufficiently similar to automobile collisions, this ATD may not provide sufficient biofidelity to accurately assess injury due to falls. Even within the realm of automobile safety, it is acknowledged that the Hybrid III does not provide substantial biofidelity for lateral impacts [23], and other ATDs like the Department of Transportation Side Impact Dummy, EUROSID-1, and EUROSID-2 were developed to fill this gap. Additionally, the Hybrid III headform is stiffer than cadaver skulls, and this difference has been shown to have a significant impact on simulation results [17]. We concede these limitations but assert that the Hybrid III is a well-researched point of comparison and a suitable starting point for this research. Computer simulations hold the potential to better approximate the mechanical properties and reactions to falls of living humans more closely than is possible with ATDs. Additionally, specific individuals and circumstances can be simulated to reconstruct falls that resulted in specific injury or to evaluate the effectiveness of protective measures, such as padded flooring or hip protectors, and of protective responses.

Since accurate fall kinematics and ground contact parameters have been identified as critical components of accurate fall simulations, next steps include using motion capture data, rotational accelerometers, or gyros to improve the accuracy of fall kinematics. This will allow the triaxial accelerometers used here to be oriented within the laboratory coordinate system, allowing both the magnitude and direction of the resultant impact velocity to be determined and both the magnitude and direction of the gravitational acceleration to be correctly accounted for (here it was assumed to be oriented downward). This improved physical test data could then be used to validate improved ground contact parameters and ATD model properties. Such computer simulations could go beyond what is possible in experimental trials and open up new avenues of investigation by bypassing the requirements and limitations of human subjects and ATD testing. Ultimately, protective responses could be incorporated and the models adapted to predict actual injuries in real people from common fall scenarios and how these injuries could best be prevented.

CONCLUSIONS

Estimations of impact deceleration by simple formulas are inadequate to accurately capture the complexity of ground impact due to a fall. Simulations hold the potential to better characterize fall impacts, but accurate fall kinematics are crucial to accurate simulation output, because even subtle changes in fall kinematics can substantially alter the simulation output. Even with accurate fall kinematics, improved contact parameters and thorough validation to experimental data are required before any fall simulation should be used to extrapolate findings beyond what is experimentally practical or possible.

ACKNOWLEDGMENTS

We would like to thank Dr. Gail Powell-Cope for her editorial comments.

This material was based on work supported by the VA Office of Research and Development, Rehabilitation Research and Development Research Enhancement Award Program (grant E2964F), and Patient Safety Center of Inquiry, Tampa, Florida.

The authors have declared that no competing interests exist.

REFERENCES

1. Sjögren H, Björnstig U. Unintentional injuries among elderly people: Incidence, causes, severity, and costs. Accid Anal Prev. 1989;21(3):233–42. [PMID: 2736020]
2. Rivara FP, Grossman DC, Cummings P. Injury prevention. Second of two parts. N Engl J Med. 1997;337(9):613–18. [PMID: 9271485]
3. Campbell AJ, Reinken J, Allan BC, Martinez GS. Falls in old age: A study of frequency and related clinical factors. Age Ageing. 1981;10(4):264–70. [PMID: 7337066]
4. Rubenstein LZ, Josephson KR, Robbins AS. Falls in the nursing home. Ann Intern Med. 1994;121(6):442–51. [PMID: 8053619]
5. Gill TM, Williams CS, Tinetti ME. Environmental hazards and the risk of nonsyncopal falls in the homes of community-living older persons. Med Care. 2000;38(12):1174–83. [PMID: 11186296]
6. Sadigh S, Reimers A, Andersson R, Laflamme L. Falls and fall-related injuries among the elderly: A survey of residential-care facilities in a Swedish municipality. J Community Health. 2004;29(2):129–140. [PMID: 15065732]
7. Sattin RW, Lambert Huber DA, DeVito CA, Rodriguez JG, Ros A, Bacchelli S, Stevens JA, Waxweiler RJ. The incidence of fall injury events among the elderly in a defined population. Am J Epidemiol. 1990;131(6):1028–37. [PMID: 2343855]
8. O'Neill TW, Varlow J, Silman AJ, Reeve J, Reid DM, Todd C, Woolf AD. Age and sex influences on fall characteristics. Ann Rheum Dis. 1994;53(11):773–75. [PMID: 7826141]
9. Nevitt MC, Cummings SR. Type of fall and risk of hip and wrist fractures: The study of osteoporotic fractures. The Study of Osteoporotic Fractures Research Group. J Am Geriatr Soc. 1993;41(11):1226–34. [PMID: 8227898]
10. Vellas BJ, Wayne SJ, Garry PJ, Baumgartner RN. A two-year longitudinal study of falls in 482 community-dwelling elderly adults. J Gerontol A Biol Sci Med Sci. 1998;53(4):M264–74. [PMID: 18314565]
11. Nyberg L, Gustafson Y, Berggren D, Brännström B, Bucht G. Falls leading to femoral neck fractures in lucid older people. J Am Geriatr Soc. 1996;44(2):156–60. [PMID: 8576505]
12. Palvanen M, Kannus P, Parkkari J, Pitkäjärvi T, Pasanen M, Vuori I, Järvinen M. The injury mechanisms of osteoporotic upper extremity fractures among older adults: A controlled study of 287 consecutive patients and their 108 controls. Osteoporos Int. 2000;11(10):822–31. [PMID: 11199185]
13. Hayes WC, Myers ER, Morris JN, Gerhart TN, Yett HS, Lipsitz LA. Impact near the hip dominates fracture risk in elderly nursing home residents who fall. Calcif Tissue Int. 1993;52(3):192–98. [PMID: 8481831]
14. DeGoede KM, Ashton-Miller JA, Schultz AB. Fall-related upper body injuries in the older adult: A review of the biomechanical issues. J Biomech. 2003;36(7):1043–53. [PMID: 12757814]
15. Feldman F, Robinovitch SN. Recalling the mechanics of falls. Young adults can not accurately describe the sites of impact immediately after a fall occurs. In: Proceedings of the 8th Annual Transforming Fall Prevention Practices; 2007 Apr 13–18; Clearwater, FL.
16. Feldman F, Robinovitch SN. Reducing hip fracture risk during sideways falls: Evidence in young adults of the protective effects of impact to the hands and stepping. J Biomech. 2007;40(12):2612–18. [PMID: 17395188]

17. O'Riordain K, Thomas PM, Phillips JP, Gilchrist MD. Reconstruction of real world head injury accidents resulting from falls using multibody dynamics. Clin Biomech (Bristol, Avon). 2003;18(7):590–600. [PMID: 12880706]
18. Lee WE. Forensic engineering case analysis in biomedical engineering. J Natl Acad Forensic Eng. Forthcoming 2008.
19. Hannon P, Knapp K. Forensic biomechanics. Lawyers and Judges Publishing Co, Inc; 2006. The application of biomechanical principals to impacts. p. 57–60.
20. Versace J. A review of the severity index. Document No. 710881. In: Proceedings of the 15th Stapp Car Crash Conference; 1971; New York: SAE, Inc; 1971. p. 771–96.
21. Eppinger RH, Sun E, Kuppa S, Saul R. Development of improved injury criteria for the assessment of advanced automotive restraint systems. II. Washington (DC): National Highway Traffic Safety Administration; 2000.
22. American Society for Testing and Materials (ASTM). ASTM F1292-04 Standard specification for impact attenuation of surfacing materials within the use zone of playground equipment. Philadelphia (PA): ASTM; 1991.
23. Backaitis SH, Mertz HJ. Hybrid III: The first human-like crash test dummy. New York: Society of Automotive Engineers, Inc; 1994.
24. Society of Automotive Engineers, Inc. (SAE). SAE Recommended Practice J211. Instrumentation for Impact Tests. New York: SAE; March 1995.
25. Bowers B, Lloyd JD, Lee WE, Powell-Cope G, Baptiste A. Biomechanical evaluation of injury severity associated with patient falls from bed. J Rehabil Nurs. Forthcoming 2008.

Submitted for publication June 22, 2007. Accepted in revised form May 15, 2008.