Development of a Baseline Model to Investigate the Biomechanics of Fall Incidents in Older Adults Using MADYMO

Numaira Obaid, Fatemeh Khorami, Ahmed Ayoub, Tim Bhatnagar, Stephen Robinovitch, Carolyn J. Sparrey

I. INTRODUCTION

Falls are one of the most common causes of injury in older adults, affecting approximately one in three individuals over the age of 65 each year [1]. The health issues and outcomes attributed to falls not only have adverse impacts on the quality of life of the affected individual but also incur a significant financial burden on the healthcare system. Due to age-related degeneration in their tissues, such as osteoporosis in bones, older adults may be more susceptible to injury after a fall. Co-morbidities in older adults may make invasive medical or surgical interventions more complex or impossible.

Numerous groups have studied fall biomechanics and explored injury prevention strategies. Despite this broad research, fall-related injuries are a significant source of morbidity and mortality in older adults. Several studies have investigated fall events in older adults using observational studies [2]. Quantitative kinematic details, such as the angle or fall velocity can be extracted from fall videos and correlated with an observed injury. Reactions such as stepping and reaching can be important to mitigate the fall impact [2]. In contrast, controlled laboratory studies can quantify fall kinematics and impact forces [3], however, for ethical reasons, these laboratory studies cannot result in injury. The direct measurement of contact forces in real world falls resulting in injury depends on organizing a large study cohort and waiting for fall events, making it logistically prohibitive. Injuries result from forces acting on the body and injury prevention depends on reducing these forces; therefore, we need to accurately quantify forces associated with falls and injury events.

Rigid body dynamics are an effective tool to estimate the external and internal forces of a biomechanical system during a fall event where the kinematics of the event are known. These models are used to reconstruct fall events for accident investigation [4], conduct parametric studies to identify which factors influence fall biomechanics [5], and assess strategies for injury prevention [6-7]. Schulz *et al*. simulated older adult falls from a bed using a Hybrid III anthropometric test dummy [8]. They found that the initial velocity of the body and joint position did not change the contact forces on the head during a fall. Adamec *et al.* reconstructed the real-world fall of a 58-year-old man into a cellar pit and found resultant injuries aligned with force outcomes [4]. Doorly *et al*. investigated head injury in fall, slip and trip cases in seven individuals over the age of 71 and found that the pulse duration was an important factor in fall injuries [9]. Erickson *et al*. used RBD simulations, validated against experimental data collected using dummies, to demonstrate the risk of head injury from wheelchair falls [10]. An important limitation in these models is the passive nature of the simulations, where an initial position is assumed and then a velocity is assigned to the whole dummy when modelling the fall. While the lack of joint reactions may be appropriate for high-speed impacts such as automotive accidents, where the incident occurs within a relatively short time span, fall events typically include joint reactions such as reaching or stepping.

The goal of this work was to investigate the contribution of limb movements to fall biomechanics and impact forces in rigid body simulations of fall events. The specific objectives were to: 1) develop subject-specific fall simulations that mimic fall events observed in the real world [11] and 2) quantify the effect of prescribed initial joint movements that better mimic reactions during a fall on the resulting impact forces and velocities of a fall event. Advancing the biofidelity of rigid body simulations will provide an important additional tool to quantify fall biomechanics and assess injury prevention strategies.

II. METHODS

Rigid body dynamic (RBD) simulations were constructed (Madymo, TASS International, Livonia, MI) to investigate the biomechanics of backward falls in older adults. A female with height of 1.7 m and a weight of 46.5 kg was simulated [11]. A 5th percentile Hybrid III female dummy model was anthropometrically scaled to match the height and weight. Two falls were simulated using the same dummy model to investigate the effect of limb positioning and joint velocities on the resulting impact velocity and impact force of the pelvis and head. The initial positioning and kinematics of the dummy were altered to simulate two real-world fall scenarios [11]

N. Obaid is a postdoctoral fellow in Mechatronics Systems Engineering (MSE), F. Khorami is a doctoral student at MSE, S. Robinovitch is a Professor in the Biomedical Physiology and Kinesiology Department, C. Sparrey (e-mail: csparrey@sfu.ca; tel: 778.782.8938) is an Associate Professor in the School of Mechatronics Systems Engineering, all at Simon Fraser University, Canada.

(Table I). For each fall event, the simulation was run twice: first, without limb motion, and then with defined initial velocities intended to simulate response to falling [11]. In the first scenario (no reaction), the initial velocity conditions were only applied to the center of gravity (CoG) of the dummy; in the second scenario (reaction), initial angular velocities were also assigned to the hip, knee, spine, shoulder and elbow joints. All joint velocities are defined relative to the proximal body segment in the joint (e.g. knee velocity is defined by the motion of the lower leg relative to the upper leg). Initial joint velocities were assigned to be proportional to the initial velocity of the CoG. The lumbar spine was assigned initial angular velocities, W1 (axial rotation) and W2 (flexion/extension) of $0.7W_{1,CoG}$ and $0.7W_{2,CoG}$. Additional angular velocities were defined at both the knees and hip, with a magnitude of $0.6W_{2,CoG}$. A comparison of no reaction and reaction models was used to quantify the effect of joint movement during a fall on the resulting impact forces and velocities. The coordinate system used in this study is shown in Fig. 1.

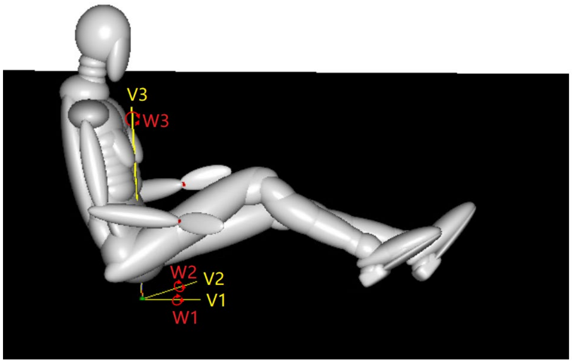

Fig. 1. The model was used to predict the fall trajectory for the falls.

Default joint stiffness values were scaled to mimic the effects of aging and muscle activation while maintaining realistic movement (neck: 1.8, lumbar spine: 1.05, hip: 1.07, and ankle: 1.6) [12]. Flooring was assumed to be linoleum over concrete with a frictional coefficient of 0.6 [13] and a stiffness of 4e6 N/m. Contact between the dummy and the floor was calculated using the assigned tissue characteristics of the dummy model as these properties were orders of magnitude less stiff than the floor [14]. Ground contact forces and velocities were recorded for the pelvis and head; head injury criteria (HIC) was calculated.

TABLE I

DESCRIPTIONS AND INITIAL VELOCITIES APPLIED TO THE INDIVIDUAL IN THE TWO SIMULATIONS ARE PROVIDED BELOW.

Fall	Fall Description
1	The individual experienced a backwards fall from standing height. The initial impact resulted in the hip and the left hand contacting the floor, followed by the head. The legs of the individual were straight during the impact. *Linear Velocities (m/s): V1 = 0.45, V2 = -1.1, V3 = -2.8 | Angular Velocities (rad/s): W1 = -3.0, W2 = -5.2, W3 = 0.6*
2	The individual experienced a backwards fall from a crouched position. The individual used their hands (wrists) to support their fall prior to the hips impacting the floor. The head then impacted the floor. *Linear Velocities (m/s): V1 = 0.5, V2 = 1.0, V3 = -1.5 | Angular Velocities (rad/s): W1 = 1.8, W2 = -2.2, W3 = 0.5*

III. INITIAL FINDINGS

Assigning initial motion to the extremities resulted in fundamental differences in the resulting falls (Fig. 2). For both falls, when initial velocity conditions were applied only to the dummy CoG, the simulations predicted no head contact with the ground (Table II). However, applying velocities to the extremities and spine resulted in head contact in both fall scenarios.

Including extremity movement had opposite effects on pelvic impact velocities when compared with no reaction. Including joint movement resulted in higher pelvic impact forces. In the simulations with active extremity motion, where head contact occurred the head contact force and HIC were higher for Fall 1 than Fall 2. Fall 2 resulted in a higher pelvic contact force (Fig. 3). In Fall 1, the right hand impacted the floor prior to pelvic contact. In Fall 2, the knees of the individual were straight prior to the impact, compared to Fall 1 with bent knees. In addition, a slight turn in the body position results in the pelvis contact occurring only on the left side in Fall 2 instead of a more bilateral contact as in Fall 1.

TABLE II
KINEMATIC AND BIOMECHANICAL DATA RESULTS FROM THE FOUR FALL SCENARIOS SIMULATED IN THIS STUDY.

	Head Impact Results					Pelvic Impact Results			
Fall #	**Time (ms)**	**Velocity (m/s)**	**Vertical Velocity (m/s)**	**Contact Force (N)**	**HIC**	**Time (ms)**	**Velocity (m/s)**	**Vertical Velocity (m/s)**	**Contact Force (N)**
1-no reaction			No head impact		167	57	0.95	-0.30	2795
1-reaction	363	0.4	-0.13	3554	989	58	0.87	-0.25	3097
2-no reaction			No head impact		0.02	38	0.75	-0.58	5872
2-reaction	219	0.7	-0.54	2028	121	38	0.86	-0.63	8631

Fall 1 **Fall 2**

No Reaction **Reaction** **No Reaction** **Reaction**

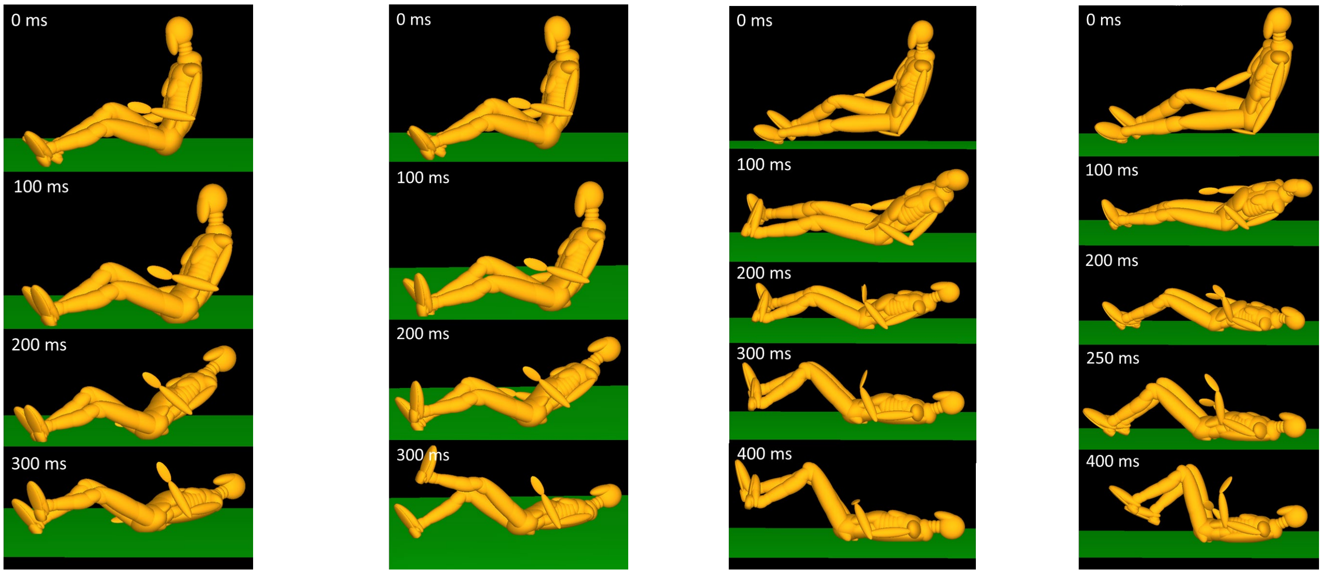

Fig. 2. The model was used to predict the fall trajectory for the falls.

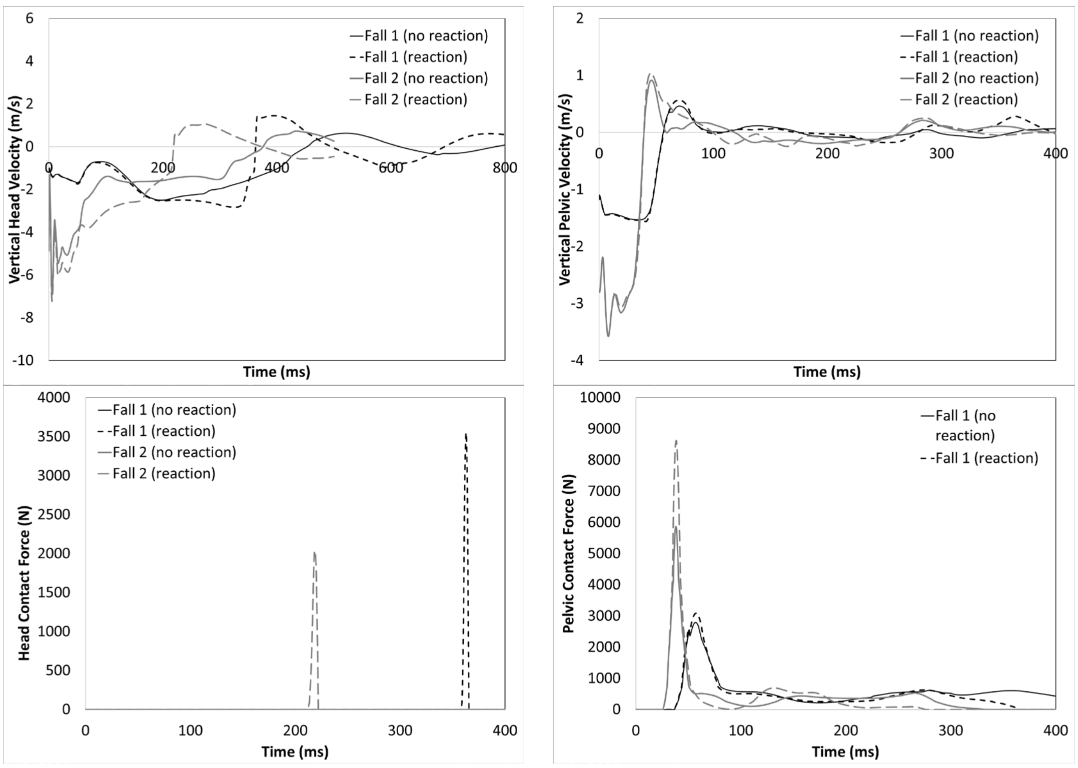

Fig. 3. Contact forces and velocities for the head and pelvis falls with (--) and without (–) joint reactions.

IV. DISCUSSION

To the best of our knowledge, prior rigid body simulations observing the biomechanics of falls [3-4] have not examined the role of joint movement during a fall event. This is particularly important because – unlike automotive accidents where an injury often occurs unexpectedly and rapidly from a static posture, preventing significant reactive movements – fall events often occur during a dynamic activity (walking, transferring, etc) and include active voluntary motions in reaction to the fall, such as reaching, stepping or twisting to avoid impact [2]. In this study, we showed that including extremity movements during a fall affect the fall trajectory, kinematics and impact force. Subtle differences in extremity movement during a fall increased impact forces up to 32% in this study. This could have important implications in resulting injuries and injury prevention strategies.

The simulation predicted that joint movement increased both pelvic and head contact forces in both falls, and therefore resulted in higher head injury criteria. Despite this, the head contact forces predicted from the simulation were substantially lower than the forces typically needed to produce a skull fracture [15]. Injury thresholds; however, are usually lower for older adults. The study also demonstrated that initial position is an important contributor to the biomechanics of a fall, where knee extension and pelvis rotation resulted in higher pelvic impact forces compared to a fall with bent knees and bilateral pelvic impact [16]. The pelvic contact forces for Fall 2 are within the range of forces that result in hip fracture [17].

This exploratory study demonstrated that extremity motion predicted head impact, altered pelvic impact velocity (8-14%) and increased pelvic impact forces by up to 32% in fall simulations. This highlights the importance of replicating the entire body movement when simulating falls. The current model has limitations that will be addressed in future work. First, the findings of the model have not been validated against experimental data. Our future work will focus on validating the results predicted from our simulation with video-recorded falls. Age-related changes in soft tissue properties are not included in the scaled 5thand will affect contact force calculations. Furthermore, the effect of different flooring materials was not explored but is known to affect impact force. Rigid body dynamic models developed from known fall kinematics including extremity motion can predict the impact forces resulting from a fall. Using this approach to study real world falls, particularly those known to cause injury, will provide a better platform for quantifying injury forces, and the efficacy of injury prevention strategies such as compliant flooring, hip protectors, or fall training.

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