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Segmental Dynamics of Forward Fall Arrests: System Identification Approach

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Abstract

Background—Fall-related injuries are multifaceted problems, necessitating thorough biodynamic simulation to identify critical biomechanical factors.

Methods—A 2-degree-of-freedom discrete impact model was constructed through system identification and validation processes using the experimental data to understand dynamic interactions of various biomechanical parameters in bimanual forward fall arrests.

Findings—The bimodal reaction force response from the identified models had small identification errors for the first and second force peaks less than 3.5% and high coherence between the measured and identified model responses ($R^2=0.95$). Model validation with separate experimental data also demonstrated excellent validation accuracy and coherence, less than 7% errors and $R^2=0.87$, respectively. The first force peak was usually greater than the second force peak and strongly correlated with the impact velocity of the upper extremity, while the second force peak was associated with the impact velocity of the body. The impact velocity of the upper extremity relative to the body could be a major risk factor to fall-related injuries as observed from model simulations that a 75% faster arm movement relative to the falling speed of the body alone could double the first force peak from soft landing, thereby readily exceeding the fracture strength of the distal radius.

Interpretation—Considering that the time-critical nature of falling often calls for a fast arm movement, the use of the upper extremity in forward fall arrests is not biomechanically justified unless sufficient reaction time and coordinated protective motion of the upper extremity are available.

Keywords

fall; injury; modeling; biomechanics

INTRODUCTION

Falls occur in all age groups commonly due to unexpected postural disturbances such as tripping, slipping, sudden turning, etc. A forward fall is known to be the most common type of fall and accounts for more than half of the falls among the elderly occur in the forward direction (Nevitt and Cummings, 1993; O'Neill et al., 1994; Vellas et al. 1998). When falling

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occurs, one tries protective motions to regain the balance often by using the upper extremity (UE) or to arrest the falls manually to avoid or reduce risk of injuries to the head or thorax (O'Neill et al., 1994; Hsiao and Robinovitch 1998). Subsequently, the UE becomes a frequent site of fall-related injuries. The distal radius fracture is the most common form of such injuries. Young individuals are also equally susceptible to fall-related injuries due to their frequent engagement in sports activities such as skating and snow boarding (Schieber et al., 1996; Machold et al., 2000). In the analysis of the 2002 National Electronic Injury Surveillance System Equipment data (USCPSC 2003), it is estimated that more than 250,000 in-line skating injuries requiring hospital emergency visits occurred nationwide. Nearly half of them involved the UE, especially the distal radius, and significantly 74% of them were fractures (Schieber et al., 1996; Machold et al., 2000).

The protective motion of the UE during falling is associated with the risk of subsequent fall-related injuries. The forearm kinematics at the time of impact, especially the elbow flexion angle or an outstretched arm, is known to be one of the major risk factors in fall-related injuries (Kim et al., 1997; Chiu and Robinovitch 1998; Chou et al., 2001; DeGoede et al., 2002). Excessive pre-impact reflexive activation of the arms can result in substantially high impact forces on the hand, commonly seen for older adults (Kim and Ashton-Miller, 2003). Thus, it is suggested that soft landing of the hand prior to the impact and coordinated post-impact muscular response would contribute to effective use of the arms in arresting falls, since soft landing would create smaller force peaks and longer timings (DeGoede and Ashton-Miller, 2002; Kim and Ashton-Miller, 2003). Biomechanical importance of the UE in modulating the impact responses necessitates further investigation to understand its efficient roles during fall arrests. One of the major difficulties in the previous *in vivo* and *in vitro* studies has been the limited physical simulation of falling due to subject safety considerations, technical difficulty of measuring rapid impact motion, inability to generate *in vivo* joint kinetics and kinematics, specimen variability, etc. As a complementary alternative, well-validated biodynamic impact models with the aid of sophisticated computer simulation software now readily allow simulation and prediction of potentially injurious impact conditions and thereby yield valuable insight into the mechanical nature of the system without compromising the aforementioned limitations. The simplest yet effective impact model of falling motion is a single degree-of-freedom (DoF) mass-spring-damper model subject to sudden velocity input or impulse force input (Robinovitch et al., 1991; Kim et al., 1997), which can be readily extended to multi-DoF models (van den Kroonenberg et al., 1995; Gruber et al., 1998; Chiu and Robinovitch 1998; DeGoede et al., 2002) or a more sophisticated rigid-body link model (DeGoede and Ashton-Miller, 2003).

Newtonian forward dynamic simulations of falling in most of previous studies require predetermined model parameters. Thus, accurate estimation of model parameters predisposes the validity of the model itself and further simulations of critical impact conditions using the constructed model. Since the mechanical properties of the joints during impact phase of fall arrests are not readily available, they were often estimated from a separate set of quasi-static experiments such as release experiments at one-handed push-up positions (Chiu and Robinovitch 1998) and ballistic pendulum experiments (DeGoede et al., 2002). The model simulation results often showed limited coherence to the measured data due to uncertainty in their model parameters and lack of thorough validation of the model response with separate experimental data, thereby jeopardizing the validity of the further simulations under extrapolated critical impact conditions. In this study, we conducted a system identification study to construct a 2 DoF impact model of bimanual forward fall arrests and to estimate the model parameters from *in vivo* falling experiment data. The constructed model was then used to simulate the response for validation with separate experiment data. Upon validation further simulations under critical impact conditions were conducted to provide improved

understanding on dynamic interaction of the body and the UE in terms of their relative velocity changes on the impact force response.

METHODS

Experiment

Previous experimental data (Kim and Ashton-Miller, 2003) were used for this study. Only brief details are described here. Simulated cable-released falling for bimanual forward fall arrests was done with a total of ten young male subjects. The subject wearing a safety harness stood on an elevated narrow platform (9 cm wide) to ensure a rotational falling motion about the ankle. The subject leaned forward with hands down into the lean control cable at a 10-degree vertical angle. The cable was released after a random time delay (0~5 sec) and the subject arrested the fall by putting the hands on a set of wall-mounted force plates (OR6-5-1, Advanced Mechanical Technology, Inc., Watertown, MA, USA). During descent phase, the subject was asked to fall like a broomstick. The distance between the standing platform and the force plate was controlled at an interval of 20 cm ranging from 40 cm (POS 1) to 100 cm (POS 4) in order to have different initial momentum of the body. The trials were repeated three times at each falling distance. Having dominance in the sagittal plane motion of a forward fall, the subjects had six infrared-emitting diode markers placed over the seven anatomical landmarks on the right-hand side of their bodies. The positions of the markers were sampled at 300 Hz using an optoelectronic motion analysis system (OPTOTRAK 3020, Northern Digital, Inc, Waterloo, Ontario, Canada). The impact forces at each hand were simultaneously recorded at the same sampling frequency with the kinematic data for five seconds for each trial. More details about the experimental setup and data processing can be found in the paper (Kim and Ashton-Miller, 2003).

Impact System Modeling

Despite simplification of a complex 3-D falling motion into a 2-D sagittal plane motion, the subject's body during descent and fall arrest often demonstrates a multiple-link motion (DeGoede and Ashton-Miller, 2003). Further progressive simplification was made as illustrated in Fig. 1 where an inverted pendulum motion about the ankle was converted into a linear motion of a 2 DoF mass-spring-damper system. The spring k_1 and damper c_1 represent stiffness and damping of the shoulder joint, respectively, whereas the spring k_2 and damper c_2 represent lumped stiffness and damping of the UE, respectively. The mass m_2 represents the effective mass of the UE located at its center of mass, while the mass m_1 represents the effective mass of the remaining half-body located at the shoulder. Though the total effective mass could be obtained from the steady state response of the ground reaction force, individual masses were estimated by assuming that the ratio of the masses 1 and 2 was a constant ratio of 8:2, which came from two considerations. First, the broomstick model should have its mass ratio such that $m_1 : m_2 = I_1 : I_2$, where I_1 and I_2 represent the corresponding mass moments of inertia about the ankle, respectively. Based on the standard anthropometric data (Winter, 2005), the mass ratio was found $m_1 : m_2 = 0.84 : 0.16$. Secondly, when the first peak force was considered dynamically coupled only with the mass of the UE, m_2 , and thus m_2 was estimated by dividing the first peak force by the corresponding acceleration of the UE, the averaged value of the estimated m_2 was found to be about 20% of total mass from the later simulations.

The state space formulation for system identification of the dynamic model in Fig. 1, is obtained by introducing two new state variables as (Palm, 2005)

$$\begin{Bmatrix} x_3 \\ x_4 \end{Bmatrix} = \begin{Bmatrix} \dot{x}_1 \\ \dot{x}_2 \end{Bmatrix} \quad (1)$$

Then, the equations of motion described in state space form are

$$\dot{\mathbf{x}} = \mathbf{Ax} + \mathbf{Bu} \quad (2)$$

where the state vector $\mathbf{x} = \{x_1 \ x_2 \ x_3 \ x_4\}^T$ and the input vector $\mathbf{u} = \{m_1 g \ m_2 g \ 0 \ 0\}^T$. The coefficient matrices are given by

$$\mathbf{A} = \begin{bmatrix} \mathbf{O} & \mathbf{I} \\ -\mathbf{KI} & -\mathbf{CI} \end{bmatrix}, \mathbf{B} = \begin{bmatrix} \mathbf{O} & \mathbf{I} \\ \mathbf{MI} & -\mathbf{CI} \end{bmatrix}, \mathbf{MI} = \begin{bmatrix} \frac{1}{m_1} & 0 \\ 0 & \frac{1}{m_2} \end{bmatrix}, \mathbf{KI} = \begin{bmatrix} \frac{k_1}{m_1} & -\frac{k_1}{m_1} \\ -\frac{k_1}{m_2} & \frac{k_1+k_2}{m_2} \end{bmatrix},$$

$$\mathbf{CI} = \begin{bmatrix} \frac{c_1}{m_1} & -\frac{c_1}{m_1} \\ -\frac{c_1}{m_1} & \frac{c_1+c_2}{m_2} \end{bmatrix}, \mathbf{O} = \begin{bmatrix} 0 & 0 \\ 0 & 0 \end{bmatrix}, \mathbf{I} = \begin{bmatrix} 1 & 0 \\ 0 & 1 \end{bmatrix}$$

The initial conditions are

$$\begin{Bmatrix} x_1 & x_2 & x_3 & x_4 \end{Bmatrix} = \begin{Bmatrix} 0 & 0 & v_1^0 & v_2^0 \end{Bmatrix} \quad (3)$$

The initial velocities are taken from the measured motion analysis data at the time of impact. The ground reaction force f to the mass m_2 is given by

$$f = c_2 \dot{x}_2 + k_2 x_2 = \begin{bmatrix} 0 & k_2 & 0 & c_2 \end{bmatrix} \begin{Bmatrix} x_1 \\ x_2 \\ x_3 \\ x_4 \end{Bmatrix} \quad (4)$$

Thus, the output vector is the measured force plate signal, $y(t) = f(t)$, so that it is described as

$$\dot{\mathbf{x}}(t) = \mathbf{Ax}(t) + \mathbf{Bu}(t) \quad (5)$$

$$\mathbf{y}(t) = \mathbf{Cx}(t) + \mathbf{Du}(t) \quad (6)$$

where $\mathbf{C}$ and $\mathbf{D}$ are from Eq. 4 such that

$$\mathbf{C} = \begin{bmatrix} 0 & k_2 & 0 & c_2 \end{bmatrix}, \mathbf{D} = \begin{bmatrix} 0 & 0 & 0 & 0 \end{bmatrix} \quad (7)$$

Equations (5) and (6) can be rewritten as

$$Y(t) = \begin{bmatrix} x(t+1) \\ y(t) \end{bmatrix}, \quad \Theta = \begin{bmatrix} A & B \\ C & D \end{bmatrix}, \quad \Phi(t) = \begin{bmatrix} x(t) \\ u(t) \end{bmatrix}$$

$$Y(t) = \Theta \Phi(t) \quad (8)$$

Model Identification and Validation

There are two different ways in constructing a mathematical model of a system for forward dynamic impact simulation. The deterministic approach uses the laws of physics or separate physiological data to determine the model structure and corresponding model parameters without directly using the relevant experimental data (a white box model approach). On the other hand, the stochastic approach assumes no prior model structure but uses only the experimental data for system identification (a black box model approach). In this study a grey box model, or a partially known model, is constructed by combining both approaches (Fig. 3). The model structure was chosen to be a second order based on the physical observation that the UE and the rest of the body had independent motions in nature. The model parameters were estimated using a single trial data at each falling distance. It was assumed that the subject maintained the time-invariance of the model parameters during trials at the same falling distance and thereby only the impact velocity input would modulate the output force response (Eq. 3). Based on this assumption, the identified model with trial-dependent initial conditions was used for validation with the remaining two trials. In Eq. 8 the linear regressor Θ was estimated using a reaction force profile between the time of the first force peak to the time of the steady state response by the least square method. Thus the current method becomes a prediction-error identification method (Ljung 1987). The estimation errors for the first and second force peak responses were recorded as well as the coefficients of determination for the linear regression between the measured and simulated responses. The procedure was repeated in turn by using each trial for identification and the remaining trials for validation. Out of the three sets of the model parameters one with the best estimation during identification and validation was accepted as the representative model parameters for the subject at the falling distance. The actual implementation of system identification and simulations were made using the *PEM* and *SIM* functions in the MATLAB System Identification Toolbox, respectively (Ljung 2004; MathWorks, Inc, Natick, MA, USA).

Model Simulation

Upon validation, the representative model parameters as a function of vertical falling height were taken from averages of all the subject data at each falling height for separate validation simulations with the conditions found in the literature (DeGoede KM, Ashton-Miller, 2002) and for understanding the segmental dynamics in terms of relative forearm velocity. The effective vertical falling height was estimated from the vertical distance between the shoulder height before the cable release and after the fall arrest at steady state. Then, the corresponding model parameters at the specified falling height were extrapolated from their regression equations. The falling height of 1.0 m and corresponding initial velocities for the natural falling motion were taken from the paper (DeGoede KM, Ashton-Miller, 2002). Similar procedures were taken for simulation of a standing fall at 1.5 m.

Statistical Analysis

Multiple analysis-of-variance (MANOVA) was conducted using SYSTAT statistical analysis software to test the effects of falling distance (POS1 to POS 4) on the model parameters. Correlation analyses and subsequent regression analyses were also conducted to find the associations between the model and response parameters using Microsoft Office Excel 2003. All analyses were based on the level of significance, $\alpha = 0.05$.

RESULTS

Model Identification and Validation

Representative model identification and validation examples are shown in Fig. 3. The measured ground reaction force profiles had two distinct peaks. Overall, the identified models for all the subjects matched well with their timings for each force peak. They had small identification errors for the first (F_1) and second (F_2) force peaks (mean=3.2% (SD, 3.0%), 3.2% (SD, 2.5%), respectively) and high coefficients of determination between the measured and identified force responses ($R^2=0.95 \pm 0.04$). Model validation for simulations with the other trials at the same falling distance had slightly increased errors for the first and second force peaks ($6.9 \pm 5.2\%$, $6.1 \pm 4.2\%$, respectively) and somewhat reduced coefficients of determination ($R^2=0.87 \pm 0.12$).

The estimated model parameters, stiffnesses and damping constants of the shoulder joint and the UE along the falling distance, are summarized in Table 1 and Fig. 4. The falling distance affected most of the model parameters ($P<0.005$ from MANOVA) except the damping constant of the UE, c_2 ($P=0.334$). As the falling distance increased from POS 1 (40 cm from the wall) to POS 4 (100 cm from the wall), the linear momentums of the body and the UE increased in exponential fashion ($R^2=0.79$, 0.59 , respectively from exponential regression analyses). Increased falling distance concomitantly increased the stiffness of the shoulder joint linearly from 3.19 to 7.87 kN/m, whereas the stiffness of the UE decreased (Fig. 4a Table 1). The damping constant of the shoulder joint remained smaller than that of the UE (Fig. 4b). Both damping constants increased along the falling distance but only the damping constant for the shoulder joint reached the statistical significance ($P=0.035$ from MANOVA).

Correlation Analysis

The model parameters demonstrated strong relationship with the response parameters, which can provide useful insight into the mechanics of falling. No strong correlation (correlation coefficient $\rho>0.5$ or $\rho<-0.5$) was noted between the velocities of the body and the UE at impact ($\rho=0.37$), indicating wide variations in strategy using the UE for each individual. Nevertheless, the linear momentum of the body ($L_2=m_2*v_2$) had a strong positive correlation with the linear momentum of the UE ($L_1=m_1*v_1$) ($\rho=0.93$), and both had strong correlation with the falling distance ($\rho=0.63$ and 0.74 , respectively). The first force peak was only strongly correlated with the impact velocity of the UE ($\rho=0.85$), whereas the second force peak was only correlated with the impact velocity of the body ($\rho=0.84$), corroborating the mass ratio assumption. All the force peaks were strongly correlated with the linear momentums of the body and the UE at impact ($\rho<0.72$) (Fig. 5). It implies the principle of linear momentum that the linear momentum change results in subsequent impact force response. Though the two velocities were not correlated to each other, the correlation between the velocity ratio v_2/v_1 and force ratio F_1 was sufficiently strong ($\rho=0.63$), resulting in a power regression with coefficient of determination $R^2=0.53$. This indicated the importance of proper use of arm in terms of relative velocity of the arm to the body for the post-impact response (Figs.6, 7).

Model Simulations

The average relative velocity ratios ($VR=v_2/v_1$) were found to be 1.73, 1.18, 0.96 and 0.99 at each falling distance, respectively. This range of relative velocity ratio was used for subsequent simulations. The estimated vertical falling heights from the motion data were 0.225, 0.247, 0.277 and 0.300 m at the four falling distances, respectively. The model parameters for the falling height of 1.0 m and 1.5 m were estimated from the regressions of the model parameters with the vertical falling height. The model simulations with the corresponding initial velocities from the paper (DeGoede KM, Ashton-Miller, 2002) indicated that as the relative velocity ratio increased from 1.00 to 1.75 at an interval of 0.25, the first force peak substantially increased

from 690 to 1,480 N (Fig. 6a). The simulation results showed good agreement with the results for 'natural falling strategy' at the reported velocity ratio of 1.5, producing errors for the first and second peaks, 9.8%, 12.9%, respectively, relative to the literature data (Fig. 6b).

Further simulation at the falling height of 1.5 m, a standing height fall, was shown in Fig. 7. As the relative velocity ratio increased from 1.00 to 1.75, the first force peak doubled from 1,250 to 2,610 N. The second force peak force also increased, but not substantially, from 610 to 800 N. Thus, it was demonstrated that under the same falling condition just moving the arm fast enough could cause compressive fracture of the distal radius at 2,350 N (lower range for males; Levine, 2002).

DISCUSSION

Biomechanical modeling studies become a meritorious choice for simulation of critical falling conditions to overcome limitations of in vivo and in vitro studies. Despite the apparent advantages of biomechanical modeling studies, construction of biodynamic models for simulation of fall arrests requires accurate physiological model data on body segment anthropometry and joint mechanical properties. Measurement of the physiological model data during impact phase is extremely difficult so that they are often estimated from a separate set of quasi-static experiments. Chiu and Robinovitch (1998) estimated hand and shoulder linear stiffness and damping factors from release experiments at one-handed push-up positions. DeGoede and Ashton-Miller (2002) estimated the hand and shoulder stiffness and damping properties from ballistic pendulum experiments. However, the mechanical properties obtained from the quasi-static experiments do not remain the same during the rapid falling event. Weak adherence of the overall model response with the experimental data and especially poor matching of the peak timings and second force peak were commonly seen in the previous studies. It is essential to estimate the mechanical properties directly from actual experimental data to reflect true dynamics of fall arrest and to increase the credibility of the model simulation. Thus, our study is a progressive attempt from the previous studies (Chiu and Robinovitch, 1998; DeGoede and Ashton-Miller, 2003). Our model, similar to Chiu's (1998), is simple yet effective in modeling the dynamic interaction between the body and the UE as noted in the high coherence of the model response with the experimental data both in identification and validation processes. The current model could take into account the variations in arm velocity and subsequent adaptive changes in mechanical properties of the joints according to input momentum or falling height (Fig. 4). The input velocity dependency of the model parameters, due to active response of the body and individual fall arrest strategy, could jeopardize the validity of simulations based on modal parameter estimation from quasi-static experiments. Thus, simulations under critical falling conditions require extrapolations of the model parameters according to the specified falling condition at impact (DeGoede and Ashton-Miller, 2003). The current system identification approach in conjunction with successive validations could provide an effective paradigm for biodynamic modeling and simulation of fall arrests.

Simplicity is one of the major virtues of our model to investigate the dynamic interaction of the body with the UE. It has been noted that increased complexity of the model could make it more difficult to estimate model parameters in the system identification process and subsequently to validate with the experimental data. With complex models, empirical tuning and unjustified guess of certain model parameters became necessary as noted in the previous studies (Chiu and Robinovitch, 1998; DeGoede and Ashton-Miller, 2003), thereby obscuring uncertainty of the further simulation results using the model. Furthermore, adding nonlinear mechanical elements (DeGoede and Ashton-Miller, 2003), despite their closer resemblance with the physics, add much more complexity in the system identification and validation processes, since it entails multiple solution sets (local minima) during parametric estimation and thus adds further uncertainty. The current model, however, is simple, requiring minimal

human intervention during system identification, validation and simulation processes. A planar model assumption for modeling a bimanual forward fall arrest is justified by the fact that the arm movement in the medio-lateral direction was most noticeable for the elbow marker but it was less than 12 cm, small enough compared with the other movements (Kim and Ashton-Miller, 2003; DeGoede and Ashton-Miller, 2003). The symmetric motion of the UE was assured from high similarity of the separate reaction force profiles from both hands (Kim and Ashton-Miller, 2003). Nevertheless, actual fall arrests may occur in asymmetric 3-D motion so our model needs further improvements to address such aspects. At the cost of simplicity, individual segmental motion of the UE was lumped into the motion of the center of mass of the UE in the model. Thus, more detailed dynamic interactions of the individual joints could not be differentiated from the current results. The other limitation includes that since the model assumed time-invariance of the model parameters it could not reflect the active adjustment during fall arrest such as elbow and shoulder joint torques. Though our 2-D model is a viable choice, it may require a multiple DoF mass-spring-damper model to study complex 3D dynamic interactions of the individual joints during fall arrests. Adding model complexity incrementally with thorough validations would be the future direction of this study.

The bimodality of the reaction force profile is attributable to the shorter time duration of the first force peak than the time required for the UE musculature to react to the afferent sensory signal from the hand for the second force peak. It has noted that the hand-ground viscoelastic contact stiffness for the first force peak could not be effectively modeled with a linear spring. A linear spring model would result in instant rise of the simulated ground reaction force response after physical contact as noted in the Chiu's model (1998). In our model, the rising part of the reaction force profile from the contact to the first force peak was not taken into account in a way that the initial velocity loading (EQ. 3) was applied instantly to the model at the time of the first force peak instead of the physical time of contact (Figs. 3, 6, 7). Thus, the instant first force peak became purely damping driven, explaining the dependency of the first force peak on the impact velocity of the UE as noted in the correlation analysis results and also from the simulation results for various relative velocities. However, the second force peak is attributed to more gradual buildup of force applied to the hand by the body mass (Gardner et al., 1998) so that it became more dependent on the linear momentum of the body, as also noted in the correlation analysis results. Linear regression analyses between the force peaks and the linear momentums (Fig. 5) indicated that the first force peak was higher and increased at a faster rate than the second force peak for the given and increased momentum conditions of falling, respectively. Thus, the former might become a higher risk factor to fall-related injuries (DeGoede and Ashton-Miller, 2003). However, it is still unclear which force peak is a major risk factor to fractures.

Pre-impact arm movement is a significant risk factor to the impact force (Kim and Ashton-Miller, 2003). The relative velocity of the upper extremity significantly contributed to the first force peak as demonstrated above. A fast arm movement ($VR \approx 1.75$ at POS 1) could double the peak impact force compared with the soft landing ($VR \approx 1$ at POS 4). Under time-critical falling conditions such as at the closest falling distance (POS 1), the mean VR reached 1.73 with a smaller variation than that at the farthest falling distance (POS 4). It is expected that the subjects could have sufficient reaction time to prepare for soft landing at POS 4. However, since under time-critical situations at POS 1 limited available reaction time calls for reflexive arm movements at faster speed, which in turn tend to induce outstretched arms (Kim and Ashton-Miller, 2003). Thus, it is not recommended to use the UE in reflexive fashion, especially for the elderly. They have delayed sensory detection of falling due to the normal aging processes associated with visual, vestibular, and neuromuscular changes (Chen et al., 1994; van den Bogert et al., 2002). The delayed detection would result in an effective decrease in available reaction time for soft landing and could cause a faster arm movement. Thus, any fall-related injury intervention using the UE in reflexive fashion may not be effective for them

at all. This may necessitate other types of falling interventions, which include personal protective devices or various falling techniques disciplined in martial arts for Judo or Aikido, or so-called “ukemi”. The ukemi techniques require to use the UE in such a way as to avoid axial loading to the arm and to absorb the energy in a larger surface area for energy shunting. General principles of ukemi techniques are to avoid landing or contacts with the corners of the body such as head, elbow, spine, etc., to roll the body in a small circle in compact manners, to maintain flexible postures so as not to crash or hit the ground, and to use other body parts for counteractive impact force at touchdown such as hand or leg slapping (Bookman, 1994; Waite, 1999). It has been shown that relaxing the body during a fall and the hand slapping during fall arrests may reduce the impact velocity (van den Kroonenberg et al., 1996) and peak impact force (Sabick et al., 1999), respectively. Yet there exist no conclusive biomechanical evidence on the efficacy of ukemi techniques.

In summary it was demonstrated that a 2 DoF model could provide sufficient information on dynamic interaction between the UE and the body during forward fall arrests through system identification, validation, and simulation processes. The fast arm movement alone could be a major risk factor to fall-related injuries. Considering the time-critical nature of fall arrests, the protective motion should be adopted when a faller has sufficient time to manage the fall arrests.

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Figure 1.
Simplification of the forward fall arrest into a 2 DoF biodynamic impact model.

Figure 2. System identification procedure using a gray box model approach for the impact model.

Figure 3.

System identification example of the ground reaction force on the hand for a young subject (19 years old, 181 cm, 73.3 kg). (a) Identified model response with the coefficient of regression between the measure and identified model responses, $R^2 = 0.96$, relative errors for the two force peaks (F₁ and F₂), 0.9 and 7.7%, respectively. (b) Validated model response for another trial at the same falling condition with $R^2 = 0.95$, the relative errors = 6.6 and 5.2%, respectively.

(a)

(b)

Figure 4. Model parameters for (a) the joint stiffnesses and (b) the damping constants of the body and the upper extremity.

Figure 5.
Association between the two force peaks and linear momentums of the body and the upper extremity.

Figure 6. Model simulations with a representative subject at the falling height of 1.0 m (a) under various velocity ratios ($VR = v_2/v_1$) and for comparison with the literature data (velocity ratio $VR = 1.5$; Kim and Ashton-Miller, 2003).

Figure 7. Model simulations with a representative subject at the falling height of 1.5 m under various relative velocity ratios.

Table 1

Estimated model parameters from system identification analysis of a grey box model

Falling Distance	k_1 [kN/m]	k_2 [kN/m]	c_1 [N-m/sec]	c_2 [N-m/sec]
POS 1 (40 cm)	3.19 (2.47)	2.76 (2.31)	68.6 (45.4)	183.5 (33.4)
POS 2 (60 cm)	5.85 (2.96)	1.45 (1.15)	83.9 (54.0)	187.5 (57.1)
POS 3 (80 cm)	7.55 (3.47)	1.00 (0.38)	119.7 (42.4)	206.2 (61.6)
POS 4 (100 cm)	7.87 (3.17)	1.54 (1.84)	112.5 (29.1)	198.7 (64.2)