Prediction of impact force in sideways fall by image-based subject-specific dynamics model

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The impact force applied to the greater trochanter during sideways fall is a critical factor for determining whether or not a hip fracture would occur. However, the impact force is subject-dependent as it is related to the subject’s anthropometric parameters and the kinematic variables in fall. It cannot be accurately predicted by the currently available dynamics models. We developed and validated a method for constructing subject-specific dynamics models to more accurately predict the impact force. The anthropometric parameters required in the model were obtained from the subject’s whole body DXA (dual energy X-ray absorptiometry) image. The subject-specific dynamics models were then validated by protected fall tests using young volunteers. The effects of anthropometric parameters on the impact force were investigated using 90 clinical DXA images obtained from a local osteoporosis clinic center. The impact forces predicted by subject-specific dynamics models had much better agreement with the experimental data, compared with those predicted by the existing empirical functions. The parametric study results indicated that although body weight and height are the dominant parameters affecting the impact force, other parameters such as the hip vertical velocity before impact also have considerable effects. This finding suggests that the existing empirical functions that only consider body weight and height may not be able to accurately predict the impact force. As whole body DXA images are readily available in osteoporosis clinic centers, the proposed method may have potential applications in the clinic to improve the assessment of fall-induced hip fracture risk.

Keywords: sideways fall; impact force; dynamics model; hip fracture; dual energy X-ray absorptiometry (DXA); anthropometric parameters

1. Introduction

Hip fracture has become a common health problem for people older than 50 all over the world, especially in the North America (Carrière \textit{et al.} 2007; Brauer \textit{et al.} 2009; Leslie \textit{et al.} 2009; Dhanwal \textit{et al.} 2011). About 90\% of hip fractures occur as a result of low-trauma fall (Cumming & Klineberg 1994), typically a fall from a standing height that would not cause any severe injury to a healthy individual. Sideways (or lateral) fall has been identified as the most critical condition for older people to develop hip fractures (Nankaku \textit{et al.} 2005; Silva 2007), as there is very little soft tissue covering the hip. Based on theories of material strength and mechanics, whether or not a fall would result in a hip fracture is determined by both the bone strength and the impact force induced in the fall. Finite element analysis (FEA) based on engineering mechanical models (Lotz \textit{et al.} 1991; Testi \textit{et al.} 1999; Keyak \textit{et al.} 2005; Bessho \textit{et al.} 2007; Majumder \textit{et al.} 2007; MacNeil & Boyd 2008; Luo \textit{et al.} 2011; Tsouknidas \textit{et al.} 2012; Luo, Ferdous, & Leslie 2013; Luo, Ferdous, & Ouyang 2013; Naylor \textit{et al.} 2013) has proved to be a promising tool for more accurately assessing hip fracture risk, as it is able to consider all the involved factors based on well-established mechanical theories and mathematical equations. FEA of hip fracture risk usually requires the impact force as an input for assessment. However, fall-induced impact force is both case and subject dependent i.e. the magnitude of the force is jointly determined by the way the fall is triggered, the body configuration before touching the ground (Kroonenberg \textit{et al.} 1996; Groen \textit{et al.} 2008), the ground hardness (Gardner \textit{et al.} 1998), the subject’s body weight, height (Kroonenberg \textit{et al.} 1995), body mass distribution, muscle strength, the thickness of the soft tissue covering the hip (Laing & Robinovitch 2010), etc. Protected fall experiments and dynamics simulations are the two major methods available for studying fall dynamics. Protected fall tests using young volunteers have been very helpful and valuable for validating fall-related dynamics models (Kroonenberg \textit{et al.} 1996; Groen \textit{et al.} 2008). However, fall test is neither ethical nor practical for older people. Computer simulation based on a validated dynamics model seems to be the only feasible way for studying fall dynamics of the elderly.

Nevertheless, only a few dynamics models are available for studying fall dynamics (Kroonenberg \textit{et al.} 1996; Groen \textit{et al.} 2008).
The dynamics models developed by Kroonenberg et al. (1995) represent the first attempt in computer modeling of sideways falls dynamics. Zhou et al. (2002) and Kim and Ashton (2009) studied forward fall during gait. Lo and Ashton-Miller (2008) investigated the effectiveness of pre-impact segmental movements on reducing injury. Although the existing models have been very helpful in understanding fall dynamics, the major limitation is that they are not subject-specific i.e. the anthropometric and dynamics parameters in the models are not obtained from the subject in concern, but either estimated from a database or taken from experimental cases reported in the literature. Therefore, the models have very limited accuracy in predicting the impact force for a specific individual. Another limitation is that joint stiffness has not been appropriately considered. Studies have shown that joint stiffness has a considerable effect on the dynamics of the human body (Frigo et al. 1996; Troy 2004; Silder et al. 2007; Silder et al. 2008; Whittington et al. 2008). However, the dynamics of the human body, including the behavior of the joints, during an accident fall becomes very complicated due to the instinctive and reflexive action. To our knowledge, very little has been known about human body joint stiffness in an accident fall. The dynamics models in the literature (Kroonenberg et al. 1995; Zhou et al. 2002; Lo & Ashton-Miller 2008; Kim & Ashton 2009) considered the joint stiffness in two extreme ways. In one way (Kroonenberg et al. 1995; Zhou et al. 2002), muscle forces and joint moments are ignored. In the other way (Lo & Ashton-Miller 2008), muscle forces at joints are represented by actively controlled actuators. Fall dynamics of older people is something between the above two extreme cases. The effect of joint stiffness and damping parameters at joints on impact force induced in a sideways fall has been preliminarily studied by Luo et al. (2014).

The objective of this study is to improve the prediction of impact force induced in sideways falls by developing subject-specific dynamics models with joint stiffness appropriately considered.

2. Methods and materials

2.1. Governing equations of sideways fall and impact

A sideways fall consists of two stages, the falling followed by the impact. In the falling stage, body motion is affected mainly by the gravitational force. The potential energy associated with the initial height of the body mass center and the body weight is increasingly converted into the kinetic energy in the fall. The impact stage starts from the time instant when the hip or other parts of the body first hits the ground. Although both gravitational force and impact force occur in the impact stage, the impact force is considerably larger than the gravitational force. Therefore, the impact stage is mainly governed by the impact force. In the following, governing equations for the two stages are established.

Fall dynamics is very complicated due to the complexity in the musculoskeletal structure and the large number of degrees of freedom required to represent all the body motions. A simple dynamics model has been considered in this study for the following reasons. First, the objective of the study was to investigate if a subject-specific dynamics model is more accurate than the existing dynamics models. Second, the study conducted by Kroonenberg et al. (1996, 1995) showed that increasing the number of links in the dynamics model did not improve prediction of the impact force. In addition, a simple dynamics model is preferred in a clinical condition. Therefore, the three-links dynamics model developed by Kroonenberg et al. (1995) has been adopted in our study. The model has been extended and improved to make it subject-specific. The model is shown in Figure 1 and it consists of three rigid links, representing the shank segment, the thigh segment, and the trunk segment, respectively. The shank link is hinged to the ground with the assumption of no slippage between the feet and the ground. The assumption can be justified by the fact that the majority of accident falls in older people occurred in nursing homes where the floors are usually carpeted (Carrière 2007). The knee joint has only one degree of freedom representing the knee flexion-extension motion. The hip joint has two degrees of freedom, representing the flexion-extension and the abduction-adduction motion. Axial rotations of the segments are not considered. Joint stiffness and damping effects have been obtained from experiments (Luo et al. 2014). Based on experimental observations described in Kroonenberg

![Figure 1. Three-links dynamics model for simulating a sideways fall. (a) sagittal plane; (b) coronal plane.](image)
et al. (1996), three generalized coordinates, $\alpha$, $\beta$ and $\gamma$, as shown in Figure 1, are required to describe the motions during the sideways fall. $\alpha$ is the angle between the vertical and the projection of the shank segment on the $y$-$z$ plane, $\beta$ is the angle between the shank segment and the line connecting the hinge at the floor and the hip joint, and $\gamma$ is the angle between the vertical and the projection of the trunk segment on the $y$-$z$ plane. To reduce model complexity and solution time, the angle ($\kappa$) between the vertical and the projection of the trunk segment on the $x$-$z$ plane is considered as a constant according to experimental observations (Kroonenberg et al. 1996; Luo et al. 2014).

For the fall stage, the governing equations have been derived from the Lagrange dynamics equations (Meirovitch 2001),

$$ \frac{d}{dt} \left( \frac{\partial T}{\partial q_i} \right) - \frac{\partial T}{\partial q_i} + \frac{\partial V}{\partial q_i} = M_i \quad (i = 1, 2, 3) \quad (1) $$

In the above equations, $q_1 = \alpha$, $q_2 = \beta$, and $q_3 = \gamma$ are the three generalized coordinates; A over-dot represents the first-order time derivative of the generalized coordinate. $T$ and $V$ are the kinetic and potential energy in the dynamics system, respectively. The expressions of the kinetic and potential energy for the three-links dynamics model are provided in Appendix 1. Due to the complexity of the energy expressions, Maple codes have been developed to derive the motion equations. $M_i (i = 1, 2, 3)$ are the joint moments that have been considered as non-conservative forces, and are represented by a torsional spring and a torsional damper working in parallel (Luo et al. 2014), i.e.

$$ M_i = k_i q_i + c_i \dot{q}_i \quad (i = 1, 2, 3) \quad (2) $$

The coefficients $k_i$ and $c_i$ are determined by experiments. Due to the lack of experimental data of joint parameters of older people, we have used the experimental data (see Table 1) obtained from young volunteers (Luo et al. 2014) in this study, by considering only the passive joint stiffness (and damping) and taking the smallest values from the experiments.

For the second stage of the sideways fall, the main interest is the peak impact force acting onto the hip. An impact model has been introduced to predict the peak impact force (Kroonenberg et al. 1995). The impact model is shown in Figure 2 and it consists of a mass ($M$), a spring ($K$), and a damper ($C$).

**Table 1. Joint stiffness and damping factors adopted in the three-link model.**

|        | $\alpha$ (N$\cdot$rad) | $\beta$ (N$\cdot$rad) | $\gamma$ (N$\cdot$s) |
|--------|-------------------------|------------------------|----------------------|
| $k_i$  | 74.0                    | 70.9                   | 183.8                |
| $c_i$  | 33.7                    | 95.4                   | 84.5                 |

![Figure 2. Impact model.](image)

The mass ($M$) is the so-called effective mass and only represents the part of the body that has a contribution to the impact force. The following energy equivalence relation has been used to determine the effective mass ($M$) in this study,

$$ \frac{1}{2} M v^2 = a T^a \quad (3) $$

The above relation is applied to the time instant when the hip first hits the ground. In the equation, $v$ is the vertical velocity at the hip, and it is determined by solving Equation (1). $T^a$ is the part of the kinetic energy that contributes to the impact force; it is associated with the vertical and rotational velocities of the links in the sideways plane i.e. the $y$-$z$ plane in Figure 1. The expression of $T^a$ is provided in Appendix 2. In a sideways fall, usually the other part of the body besides the hip may simultaneously have contact with the ground at the first time, which would reduce the impact force. The coefficient $a$ ($0 < a \leq 1$) in Equation (3) is used to consider multiple contacts in the fall; in this study, $a = 1$ has been considered as it represents the most critical situation leading to hip fracture. The spring ($K$) and the damper ($C$) account for the effect of soft tissue covering the hip.

The governing equation of the impact model is,

$$ M \ddot{x} + C \dot{x} + Kx = 0 \quad (4) $$

The body weight does not appear in the above equation, as the vibration displacement ($x$) is measured from the static equilibrium position. The initial conditions of the impact stage are those at the end of the falling stage, which include the hip vertical velocity ($v_0$) and the distance of the effective mass from the equilibrium position. The analytical solutions of Equation (4) can be obtained, and then, the maximum impact force can be determined by finding the maximum value in the function $F(t) = Kx + C\dot{x}$.

The spring constant ($K$) and the damping factor ($C$) have been determined from the experimental data. A typical experimental time history obtained from the sideways fall test is displayed in Figure 3, which describes the variation of the hip vertical position ($z$) with time ($t$). The damped period ($T_d$) and the oscillation magnitudes, $z_1$ and $z_2$, at the two time instants (1 and 2 in the figure)
bounding the first period can be measured from the curve. The spring constant and the damping factor can then be determined by the following steps (Meirovitch 2001). The damped frequency \( \omega_d \) is calculated as 
\[
\omega_d = \frac{2\pi}{T_d}
\]
where \( T_d \) is the first period measured from the experimental data. The damping coefficient \( \zeta \) is computed by 
\[
\zeta = \ln \left( \frac{\omega}{\omega_d} \right)
\]
then the damping factor \( \xi \) is determined as 
\[
\xi = \frac{\zeta}{\sqrt{1+\zeta^2}}.
\]
The natural frequency \( \omega_n \) is determined as 
\[
\omega_n = \frac{\omega}{\sqrt{1-\zeta^2}}.
\]
where \( \omega \) is the distance of pixel \( j \) from the distal end or from the mass center of the link.

The hip stiffness and damping property determined from the experimental data also include the effect of the protection pad used in the experiment.

### 2.2. Determination of subject-specific anthropometric parameters from whole body DXA image

All anthropometric parameters required for constructing the subject-specific dynamics model in this study have been either directly measured or calculated from a whole body DXA image of the subject. A whole body DXA image can be scanned using a clinical DXA scanner. One sample DXA is displayed in Figure 4(a). The DXA image is segmented to obtain the contour of the body shown in Figure 4(b). As the bones can be clearly identified on the DXA image, lengths of the links representing the shank, the thigh, and the trunk can be directly determined from the image, as shown in Figure 4(c), using the ratio of the actual height measured by a ruler and the skeletal height measured from the figure.

Other anthropometric parameters, such as segment (or link) mass, mass center, and mass moment of inertia, can be calculated from the DXA image. Studies have shown that there exists a linear correlation between the pixel intensity in the DXA image and the tissue mass density (Durkin et al. 2002; Ganley & Powers 2004; Wicke et al. 2008; Nasiri Sarvi & Luo 2013). The correlation coefficient in our study has been determined by

\[
\chi = \frac{W}{\sum p_i}
\]

where \( W \) is the body weight of the subject that can be measured by an electric weight scale, and \( n \) is the total number of pixels enclosed in the body contour. \( p_i \) is the pixel intensity of pixel \( i \). With the above correlation coefficient, the mass of pixel \( i \) is obtained as \( \rho_i = \chi p_i \). Mass, mass center, and mass moment of inertia of the three links can be calculated based on the definitions i.e.

\[
m_k = \sum_{i=1}^{N_k} \rho_i, \quad r_k = \frac{\sum_{i=1}^{N_k} \rho_i r_i}{m_k}, \quad I_k = \sum_{i=1}^{N_k} \rho_i r_i^2
\]

\((k = 1, 2, 3)\)

2.3. Protected and controlled fall tests using young volunteers

Three young male volunteers were recruited in this study under a human body research ethics approval. The subjects were first scanned by a clinic DXA scanner (Lunar Prodigy, GE Healthcare, USA) and their whole body DXA images were obtained. Subject-specific anthropometric and dynamics parameters were obtained from the DXA images using the method described in Section 2.2. The obtained anthropometric parameters are listed in Table 2. For the purpose of comparison, the parameters of the 95th percentile subject in Kroonenberg et al. (1995) are also provided in the table. Coincidentally, the 95th percentile has very similar weight and height as Subject 2. Three subject-specific dynamics models were constructed using these parameters.

The volunteers then participated in protected and controlled fall tests. The objective of the tests was to obtain experimental data to validate the subject-specific dynamics model and the impact model constructed by the method described in Sections 2.1 and 2.2. Kinematics data in the fall stage and the impact force onto the greater trochanter were collected in the tests. Although the volunteers were young, a fall from a standing height and letting the hip directly hit the ground without any protection would probably cause serious injury if not hip fracture. Therefore, a special system was designed for conducting the tests.

The system is shown in Figure 5 and mainly consists of an electromagnetic release switch, a number of nylon slings, a harness, a protection pad, a force plate (AMTI OR6-7MA, A-Tech Instruments Ltd, Canada) under the
A protection pad, and a motion capture system consisting of six infrared cameras (VICON, Vicon Motion Systems Ltd, UK). The protection pad is made of foam and has a pre-calculated thickness so that it is just sufficient to protect the volunteer from injury. Reflective markers were put at the main joints of the subject. The subject was positioned by adjusting the slings so that the distance between the hip and the protection pad is about 30–50 cm. After the subject was released, the time histories of the reflective markers were automatically recorded by the motion capture system. A data sampling rate of 200 frames per second was used in all the tests.

The motion data of the relevant reflective markers were then used to calculate the angles (the generalized

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**Figure 4.** Determination of anthropometric parameters from whole body DXA image. (a) Original clinic DXA image; (b) segmentation of body contour; (c) division of trunk, thigh, and shank links.

**Table 2.** Anthropometric parameters obtained from whole body DXA images.

| Subject # | 1        | 2        | 3        | 95th percentile (Kroonenberg et al. 1995) |
|-----------|----------|----------|----------|------------------------------------------|
| Total mass (kg) | 75.6     | 70.4     | 64.6     | 70.04                                    |
| Height (m)    | 1.73     | 1.72     | 1.74     | 1.72                                     |
| Body mass index (BMI) | 25.73 | 24.34 | 21.14 | 23.67                                    |
| Segment length (m) | Shank | 0.44     | 0.42     | 0.45                                     |
|                | Thigh   | 0.45     | 0.41     | 0.44                                     |
|                | Trunk   | 0.84     | 0.89     | 0.88                                     |
| Segment mass (kg) | Shank | 7.3      | 6.9      | 6.8                                      |
|                | Thigh   | 21.1     | 18.4     | 18.0                                     |
|                | Trunk   | 47.1     | 45.1     | 39.8                                     |
| Segment mass center* (m) | Shank | 0.25     | 0.23     | 0.22                                     |
|                | Thigh   | 0.26     | 0.24     | 0.27                                     |
|                | Trunk   | 0.37     | 0.38     | 0.38                                     |
| Segment mass moment of inertia (kg m²) | Shank | 0.14     | 0.14     | 0.15                                     |
|                | Thigh   | 0.31     | 0.23     | 0.28                                     |
|                | Trunk   | 2.63     | 2.33     | 1.78                                     |

*Measured from the distal end of the link.
coordinates) and the angular velocities of the links. First, the spatial coordinates of the left and the right markers at a time instant were averaged to obtain the positions of the ankle, the knee, and the hip joint. The angles, $\alpha$, $\beta$, and $\gamma$, were determined from the positions of the joints. The expressions for computing the angles from the spatial coordinates of the joints are given in Appendix 3. The initial conditions of the fall tests, which are required in the dynamics simulation of the falls, were retrieved from the experimental data and they are listed in Table 3.

2.4. Parametric study of sideways fall of the elderly

The impact force induced in a sideways fall is affected by a number of anthropometric and kinetic parameters, including the body height, weight, mass distribution, thickness of soft tissue covering the hip, and hip velocity. Therefore, parametric studies have been conducted to find out how the parameters affect the impact force. A total of 90 clinical cases were obtained from the Manitoba Bone Mineral Density Database, a clinical center for osteoporosis screening and monitoring in the province of Manitoba, Canada. For each case, the information acquired for the study includes a whole body DXA image, body weight, and height of the patient. The clinical cases include 50 females and 40 males. Statistical information of the patients is provided in Table 4. For each patient, a subject-specific dynamics model was constructed using the previously described method. The impact force was predicted by the dynamics simulation. Correlations between the impact force and the involved anthropometric/kinematic parameters were then investigated.

3. Results

The predicted hip velocities and impact forces for the three volunteers together with the experiment data are presented in Section 3.1; these results are intended to validate the method of constructing the subject-specific dynamics and impact models. Parametric study results are provided in Section 3.2, which were used to investigate the dominant parameters affecting impact force in the sideways fall, and to estimate impact force using the dominant parameters if a whole body DXA image is not available.

3.1. Validation of the method for constructing subject-specific dynamics models

For the three volunteers, the predicted hip position, hip vertical velocity, and impact force together with the corresponding variables determined from the experimental data are presented in Figures 6–8.

The hip vertical velocities just before impact and the peak impact forces obtained by both prediction and experimentation are listed in Table 5. The falls in the tests were not from a standing height but from a lower height due to injury concerns. The dynamics simulation results in Table 5 were produced by using the initial conditions in Table 3. For the purpose of comparison, results obtained by Kroonenberg et al. (1995) from their dynamics models and experiments are also presented.

Table 3. Initial conditions extracted from experimental data.

| Subject | 1 | 2 | 3 |
|---------|---|---|---|
| $\alpha_0$ (rad) | 1.10 | 1.11 | 0.70 |
| $\beta_0$ (rad) | 0.00 | 0.00 | 0.00 |
| $\gamma_0$ (rad) | 0.66 | 0.70 | 0.84 |
| $\dot{\alpha}_0$ (rad/s) | 0.90 | 0.85 | 0.52 |
| $\dot{\beta}_0$ (rad/s) | 0.00 | 0.00 | 0.00 |
| $\dot{\gamma}_0$ (rad/s) | 0.00 | 0.00 | 0.00 |

Table 4. Statistical information of the 90 osteoporosis patients.

|                | Age (years) | Height (cm) | Weight (kg) | BMI   |
|----------------|-------------|-------------|-------------|-------|
| Range          | 65–88       | 124.2–186.2 | 29.5–133.8  | 14.8–40.6 |
| Average        | 72          | 162.0       | 69.1        | 26.2  |
The 5th and 95th percentile subjects from Kroonenberg et al. (1995) were a short female and a tall male, respectively, both having normal body mass index (BMI).

To simulate sideways falls of the volunteers from their standing height, the initial conditions in Table 3 were changed to standing configuration i.e. \( \alpha_0 = 0 \), \( \beta_0 = 0 \) and \( \gamma_0 = 0 \), and the dynamics simulations were recondutcted. The predicted hip velocities and the peak impact forces are listed in Table 6. The results are comparable to the experimental results of the 95th percentile reported by Kroonenberg et al. (1995), as the subjects have similar body weight and height.
3.2 Parametric studies of sideways fall

The hip stiffness and damping factor for older people are not available in the literature. To estimate the parameters, the hip stiffness and damping factors obtained from the fall tests were correlated to the subjects’ body mass indices by linear regression. The obtained correlations shown in Figure 9, and Equation (7) was used to estimate the hip stiffness ($K$) and damping factor ($C$) required in Equation (4).

$$K = 2.7 \times 10^5 - 7.9 \times 10^3 \eta,$$
$$C = -4.6 \times 10^2 + 47.0 \eta$$

(7)

By single-variable linear regression, correlations between the impact force and the anthropometric parameters i.e. the height, weight, body mass center, BMI, and thickness of soft tissue at the hip were obtained. The relations are presented in Figure 10. The correlation between the impact force and the hip velocity is shown in Figure 11(a), and the correlation between the hip velocity and the body height is provided in Figure 11(b).

By multivariate linear regression, the correlation between the impact force and the anthropometric parameters has been obtained as

![Figure 9. Correlation between (a) hip stiffness and BMI ($\eta$); (b) damping factor and BMI ($\eta$)](image)
Figure 10. Correlation between impact force and (a) body height, (b) weight, (c) body mass center, (d) BMI, (e) thickness of soft tissue at hip, established by dynamics simulation results.
\[ F_{\text{impact}} = -9722.27 + 113.56 \times \text{BMI} + 89.54 \\
\quad \times \text{Height (cm)} - 24.68 \times \text{Weight (kg)} - 86.57 \\
\quad \times (\text{Height of Mass Center, cm}) + 1512.48 \\
\quad \times (\text{Hip Velocity, m/s}) \\
(\text{r} = 0.693, p < 0.0001) \]

(8)

If only weight and height are considered, the correlation becomes,

\[ F_{\text{impact}} = -894.78 + 24.28 \times \text{Height (cm)} + 25.40 \]

\[ \times \text{Weight (kg)} \]

\[ (r = 0.573, p < 0.0001) \]

(9)

The above relations have been established using the data produced by the dynamics simulations rather than directly from experimental results. Some factors, for example, the angular velocities of the links, have not been incorporated in the relations. Therefore, the accuracy of an impact force estimated by the above relations may be limited.

4. Discussions

From the results in Table 2, the anthropometric and dynamics parameters identified from the DXA images of the three volunteers are very close to those of the 95th percentile subject in Kroonenberg et al. (1995), which, by coincidence, has very similar body weight and height as Subject 2 in our study. These results well verified that our method for estimating anthropometric parameters from the DXA image is reliable and accurate. As shown in Figures 6–8, the time histories of hip position, hip velocity, and impact force predicted by the image-based subject-specific dynamics models have good to excellent agreement with the corresponding experimental data. Compared with those results produced by the existing dynamics model (Kroonenberg et al. 1995), the prediction accuracy has been greatly improved. The relative errors in the predicted hip velocity and the peak impact force, measured with respect to the experimental data (Table 5), have been reduced in average from 40.1 and 35.4% to 6.5 and 7.3%, respectively. The results in Table 6 show that the predicted hip velocities and peak impact forces induced in sideways falls from standing height are also very close to the experimental results reported in Kroonenberg et al. (1996, 1995). The method of constructing subject-specific dynamics models proposed in this study was thus well validated by our own experiments and also by the experimental results reported in the previous study (Kroonenberg et al. 1995). How the subject-specific anthropometric parameters affect the impact force can be explored from a number of aspects. By examining the governing equations in Equation (1) and the energy expressions in Appendix 1, there are basically two groups of variables involved in the dynamics of a sideways fall. One group consists of kinematic variables such as the link angles and angular velocities, which are determined by solving the dynamics equations. The other group includes anthropometric parameters, for example, the segment lengths, segment masses, mass centers, and mass moments of inertia. These anthropometric parameters are not only affected by the subject’s overall physiological attributes such as body weight and height, but also by the distribution of the body mass. Two subjects having the same weight and height but different body shape, for example, longer trunk versus longer lower extremities, or pear-shaped versus upside-down pear-shaped, may have very different segmental anthropometric and dynamics parameters. The differences will affect the resulting kinematics in the fall and thus also the impact force. By the impact model shown in Figure 2, the impact force is explicitly related.
tial energy associated with the large body weight, which is evidenced by the small correlation coefficient \( r = -0.1612 \) and the large \( p \)-value \( (p = 0.15) \) in the correlation shown in Figure 10(c). As can be seen from Figure 10(c) and (d), in addition to body height and weight, body shape also has considerable effect on the impact force. Although body shape cannot be accurately described only by body mass center and BMI, they indeed provide some information of the body shape. The BMI indicates if a subject is thin or fat. The location of body mass center roughly describes mass distribution, for example, a pear-shaped subject has a lower mass center compared to an upside-down pear-shaped subject.

Along with the anthropometric parameters, kinematic variables also affect the impact force. The effect of hip velocity on the impact force is shown in Figure 11(a). As discussed before, other kinematic variables, for example, the rotations of the links, implicitly take effect on the impact force via the effective mass. That explains why the correlation coefficient \( r = 0.693 \), \( p < 0.0001 \) in Equation (8) is still far below unit, as not all the influential factors have been included in the correlation. According to the definition of effective mass in Equation (3) and the energy expression in Appendix 2, the effective mass may not be a constant even for the same subject. Any change in the involved kinematic variables e.g. in the initial conditions, would alter the effective mass and thus the impact force. Nevertheless, it is not practical to consider all the involved factors in the correlation, as the expression would be very complex. Furthermore, the anthropometric parameters and the kinematic variables are dependent to each other, for example, Figure 11(b) shows the hip velocity is correlated to the body height. Therefore, the reliable way to obtain the impact force is by a dynamics simulation using a subject-specific model. The required inputs for the simulation include a whole body DXA image and the subject’s height and weight, which are readily available from a clinical center. If the DXA image is not available, the empirical function in Equation (9) can be used to estimate the impact force, but the accuracy is lower.

Although the proposed image-based subject-specific dynamics modeling method is able to greatly improve the accuracy of the predicted impact force, there are still a couple of factors affecting the accuracy. Axial rotations of the subject may have occurred in the tests but are not considered in the dynamics model. The two lower extremities may also have different motions in the tests, which may have not been well described by a single link. Accident fall in real life is obviously much more complicated and cannot be fully described by the simplified dynamics model developed in this study. To predict the impact force in a real-life fall, a more advanced whole
body dynamics model should be developed with the consideration of complicated body configurations and kinematic motions in a real-life fall.

The subject-specific dynamics modeling method developed in this study and the DXA-based proximal femur finite element model proposed in (Luo et al. 2011; Luo, Ferdous, & Leslie 2013) can be integrated as a clinical tool for assessing hip fracture risk. The impact force predicted by the dynamics model can be used as the input to the DXA-based femur finite element model in calculating the fracture risk index, which is a ratio representing the applied force to the bone strength (Luo et al. 2011; Luo, Ferdous, & Leslie 2013).

5. Conclusions

Subject-specific dynamics modeling is able to more accurately predict impact force induced in a sideways fall. The impact force is affected by a number of anthropometric parameters and kinematic variables. Body height and weight have a dominant effect on the impact force. However, the kinematic variables and the other anthropometric parameters also have non-trivial effects on the impact force. Due to the dependence of the impact force on various kinematic variables, it is difficult to establish an empirical correlation to accurately estimate the impact force. The reliable way for predicting the impact force is to conduct a subject-specific dynamics simulation. Considering that whole body DXA images are readily available in osteoporosis clinic centers, the proposed subject-specific dynamics modeling method can be integrated with other relevant techniques into a clinical tool for assessing hip fracture risk.

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Appendix 1

Symbols

\( m_i \) Mass of Link \( i \)
\( l_i \) Length of Link \( i \)
\( l_i' \) Location of mass center of Link \( i \) from the distal end
\( I_i \) Mass moment of inertia of Link \( i \)

Kinetic energy:

\[
T = \frac{1}{2} m_1 \left[ \left( -l_1' \cos \beta \cos \beta \right)^2 + \left( -l_1' \cos \beta \sin \beta - l_1' \cos \beta \sin \beta \right)^2 \right]
+ \frac{1}{2} l_1 \left[ (\dot{x} \cos \beta)^2 + (\dot{y} \cos \beta)^2 \right]
+ \frac{1}{2} m_2 \left[ -l_1 \cos \beta \cos \beta + l_2' \cos \dot{\alpha} \cos \delta \right]^2
+ \left( -l_1 \cos \beta \sin \beta - l_2' \cos \dot{\alpha} \sin \delta - l_2' \cos \dot{\delta} \sin \beta \right)^2
+ \frac{1}{2} m_3 \left[ \left( -l_1 \cos \beta \cos \beta + l_2 \cos \dot{\alpha} \cos \delta - l_2' \cos \dot{\gamma} \cos \kappa \right)^2 + \left( -l_1 \cos \beta \dot{\alpha} \cos \kappa - l_2 \cos \dot{\delta} \cos \alpha - l_2' \cos \dot{\kappa} \cos \gamma \right)^2
+ \left( -l_1 \cos \beta \dot{\alpha} \sin \beta - l_2 \cos \dot{\alpha} \sin \alpha - l_2 \cos \dot{\delta} \sin \beta - l_2 \cos \dot{\kappa} \sin \alpha - l_2' \cos \dot{\gamma} \sin \kappa - l_2' \cos \dot{\kappa} \sin \gamma \right)^2 \right]
\]

Potential energy:

\[
V = m_1 g l_1 \cos \beta \cos \alpha
+ m_2 \left( l_1 \cos \beta \cos \alpha + l_2' \cos \delta \cos \alpha \right)
+ m_3 \left( l_1 \cos \beta \cos \alpha + l_2 \cos \delta \cos \alpha + l_2' \cos \kappa \cos \gamma \right)
\]

In the above expressions, angle \( \delta \) is not an independent variable. By using the sine law and the geometrical relation shown in Figure 1, the angle and its first-order derivative with respect to time can be expressed by other variables as

\[
\delta = \sin^{-1} \left( \frac{l_1 \sin \beta}{l_2} \right)
\]

and

\[
\dot{\delta} = \frac{l_2 (\sin \delta)^2 \cos \beta}{l_1 (\sin \beta)^2 \cos \delta} \dot{\beta}
\]

Appendix 2

\[
T' = \frac{1}{2} m_1 \left[ \left( -l_1' \cos \beta \sin \beta - l_1' \cos \beta \sin \beta \right)^2 + \frac{1}{2} l_1 (\dot{x} \cos \beta)^2 \right]
+ \frac{1}{2} m_2 \left[ \left( -l_1 \cos \beta \sin \beta - l_1' \cos \alpha \dot{\alpha} \sin \delta \right)^2
+ \left( -l_1 \cos \beta \sin \beta - l_2' \cos \dot{\alpha} \sin \delta \right)^2 + \frac{1}{2} l_2 (\dot{x} \cos \delta)^2 \right]
+ \frac{1}{2} m_3 \left[ \left( -l_1 \cos \beta \sin \beta - l_1 \cos \beta \sin \alpha \right)^2
+ \left( -l_2 \cos \dot{\alpha} \sin \delta - l_2 \cos \dot{\delta} \sin \beta \right)^2 + \frac{1}{2} l_2 (\dot{x} \cos \delta)^2 \right]
+ \left( -l_2' \cos \dot{\alpha} \sin \beta - l_2' \cos \dot{\alpha} \sin \alpha - l_2 \cos \dot{\alpha} \sin \beta - l_2' \cos \dot{\delta} \sin \beta - l_2' \cos \dot{\delta} \sin \beta - l_2' \cos \dot{\kappa} \sin \gamma \right)^2 + \frac{1}{2} l_3 (\dot{\gamma} \cos \kappa)^2
\]

Appendix 3

Symbols

\( x_0, y_0, z_0 \) Spatial coordinates of the marker at the forehead
\( x_H, y_H, z_H \) Average spatial coordinates of the two markers at the hips
\( x_K, y_K, z_K \) Average spatial coordinates of the two markers at the knees
\( x_A, y_A, z_A \) Average spatial coordinates of the two markers at the ankles

The angles (generalized coordinates) are calculated from the spatial coordinates of the markers via the following expressions:

\[
\alpha = \cos^{-1} \left( \frac{|z_K - z_A|}{\sqrt{(y_K - y_A)^2 + (z_K - z_A)^2}} \right)
\]

\[
\beta = \cos^{-1} \left( \frac{|z_K - z_A|}{\sqrt{(y_K - y_A)^2 + (z_K - z_A)^2}} \right)
\]

\[
\gamma = \cos^{-1} \left( \frac{|z_O - z_H|}{\sqrt{(y_O - y_H)^2 + (z_O - z_H)^2}} \right)
\]