

Original Article

Safety considerations for forward falls

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Abstract

Objective: Forward falls are among the most frequent causes of upper extremity fractures. This study investigated the safety considerations to prevent wrist injuries during bimanual forward falls. **Methods:** A biomechanical model was developed with two separated arms to facilitate investigation of asymmetrical contact and predict the impact force applied to each hand separately. To validate the developed model, a series of fall experiments were conducted in which one hand collided with a hard surface, while the other collided with a soft surface. **Results:** The results show that the impact force applied to each hand is independent of the other. Using these results and our model, the safety aspects of human forward falls were analyzed with a view to preventing injuries. Specifically, we sought to determine the safe range of surface stiffness and damping to ensure that the occurrence of forward falls does not lead to trauma. **Conclusion:** The results of this study can be applied in the design of compliant flooring to ensure the safety of people in environments with potential fall hazards. From a robotics viewpoint, the results are applicable in the design of compliant flooring for shared workplaces, where robots collaborate with people and collisions between humans and robots may cause falls.

Keywords: Bimanual Forward Fall, Biomechanical Model, Compliant Flooring, Impact Force, Wrist Injuries

Introduction

In recent years, fall studies have attracted significant attention from robotics, safety, and biomechanical researchers. Fall-related injuries are the second most common cause of accidental or unintentional injury deaths worldwide¹, which might incur substantial costs and long-term rehabilitation². Such incidents might frequently occur in the elderly due to musculoskeletal frailty³⁻⁵. The risk factor associated with fall incidents⁶ as well as training⁷, interventions⁸, and arresting strategies⁹ have been widely studied. The highest percentage of falls among the elderly is associated with forward walking¹⁰. Elevating, lowering, and skip motion have been reported as balance recovery strategies during walking¹¹; however, unsuccessful recovery motions can cause injuries and fractures. Forward falls are a

major cause of upper extremity injuries such as distal radius fracture, which is the most common type of fracture among young adults¹².

Furthermore, collisions might occur in workplaces where humans and robots collaborate. Upon collision, although the robot might immediately stand still, the imposed force on the human body might result in a forward fall. The time for a falling person to react against a fall¹³ and avoid fall injuries is only 500 ms¹⁴. Although fall breaking strategies might alter the applied impact force, a realistic evaluation of the impact force during a fall and physical interventions to reduce injuries, such as compliant flooring, are important¹⁵. A few previous studies have proposed biomechanical models to describe the mechanism behind a forward fall and predicted the impact force using various models or software¹⁶.

Zhou et al.¹⁷ measured the tripping force (i.e. obstacle-foot impact force), which was applied to the swing foot in a twelve degree-of-freedom model. They simulated the forward fall induced by tripping; however, the human body was modeled with rigid links, and the effect of the soft tissues was not considered. Lo et al.¹⁸ used a seven degree-of-freedom model to investigate the effect of the body configuration on the impact force. Their results showed that not only the upper extremity configuration, but also the lower extremity adjustments affected the impact force significantly. Xu et al.¹⁹ used a five degree-of-freedom model (specifically, a

The authors have no conflict of interest.

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Edited by: G. Lyritis

Accepted 29 October 2019



model without arms) and formulated the forward fall motion. Although the arms were not considered in their model, the results from their study were useful for distinguishing the falls from daily activities.

Moreover, researchers have conducted experiments to reproduce fall motion with less severity to investigate the effective parameters and have proposed biomechanical models inspired by the experimental data. Chiu and Robinovitch²⁰ designed a free-fall experiment to investigate the impact force during falls. The subjects in their experiment were instructed to lock their elbows throughout the landing. Their experimental results indicated that the impact force is governed by a primary high magnitude peak (f_{max1}) that occurs shortly after contact, and a secondary lower peak (f_{max2}). The first peak impact force usually causes a fracture²¹. Moreover, they observed that the ground stiffness reduction attenuated the magnitude of the first peak impact force, but did not affect the secondary maximum impact force²². Chiu and Robinovitch also proposed a two degrees of freedom (DOF) mass-spring-damper model to predict the impact force during a forward fall resulting from a standing/walking position. Subsequently, their results were confirmed by²³.

The results reported by^{20,22,23} stemmed from fall experiments where a single force plate was used to measure the impact force. These studies assumed that the impact occurring on one hand deteriorates (i.e. the impact force decays to zero) and subsequently the other hand strikes the ground. Although bimanual forward fall is the most common strategy²⁴, the model proposed in²⁰ suggested the condition/situation where a single hand arrests a forward fall. This is the only model that predicted the hand impact force during a forward fall from a typical standing/walking position. Later, several research studies employed the above-mentioned results to estimate the risk of bone fractures²⁵⁻²⁷.

The effects of asymmetrical fall conditions including asymmetric loading and body postures, and asymmetrical contact (i.e. one hand colliding with a hard surface while the other hand is in contact with a soft surface) have rarely been studied. Particularly, very few studies have discussed the effect of asymmetric loading²⁸, and to the best of our knowledge, the effect of different types of contact has not been studied previously.

In this paper, we seek to ascertain the impact force acting on each hand during a forward fall with asymmetrical contact. A series of fall experiments were performed using two force plates. The subjects were instructed to use a bimanual fall arrest strategy, which is the most common hazardous fall breaking method among humans. To avoid the previous assumption²⁰ (i.e. where one hand is delayed until the force on the other side decays to zero), a force plate was mounted under each hand to measure the impact force applied to each hand separately. One force plate was covered by a soft surface, while the other was considered as a hard surface. A biomechanical model with two separate arms was developed inspired by the real experiments. This model enabled us to investigate the effect of asymmetrical contact and predict the impact force applied to each hand separately during an actual

forward fall. The effects of surface stiffness and damping on the impact force were investigated, and the safety aspects necessary to prevent fall-related injuries were discussed. Related information was also provided, which could be useful in compliant surface design.

Material and method

Participants

For performing these experiments, we decided to use healthy young subjects to avoid any kind of injuries. Twenty healthy male subjects ranged between 20 and 38 years participated in this study (mean±standard deviation [SD]=28.7±7.5). Their average heights and weights were 172.4±5.8 cm 64.8±11.3 kg, respectively. The subjects filled a questionnaire to ascertain that they do not have any related health problems or special fall arresting skills that may cause deviations in the results of our experiments. The participants did not report any bone diseases that would increase the risk of injuries or may cause any changes in the magnitude of the impact force. None of them suffered from movement disorders or neurological and medical illnesses, and they did not have prior training experience regarding fall prevention, martial arts techniques or Jodo. The subjects have never experienced serious fall accidents and injuries. All participants signed a specific consent form before the experiments and then received a short training. They had an opportunity to do some test experiments before starting the main trials. This study was conducted with the approval of the Institutional Review Board of University. All the experiments were conducted in accordance with the approved guidelines.

Experimental protocol

The experimental area was covered with soft yoga mats for safety reasons; two separated force plates (USO6-H5, Tech Gihan Co., Ltd., Japan) were installed to measure the impact force acting on each hand (Figure 1(a)). A supporting harness was attached to the subject's trunk. The participants were instructed to fix their shanks on a soft pillow and extend their knees to lean forward against the harness. The thigh was set at 30° from the horizontal.

By pulling the supporting harness connected to the trunk, the participants were elevated to reach the proper height from the ground (Figure 1(b)). The subjects were instructed to keep their elbows fully extended (i.e. outstretched hands) and the arm angle was set at 15° to the vertical. Thus, the participant's body posture and the experiment's protocol were similar to those of previous experiments^{20,22,23}. The surface of the force plates was considered hard surfaces (i.e. approximately infinity stiffness), whereas for the fall on a soft surface, a foam padding adhered to the surface of the force plate. We designed the experiment to determine the impact force applied to each hand under the following condition: One hand strikes the soft surface and the other one strikes the hard surface.

The experiments were performed at heights of $h=3$ cm

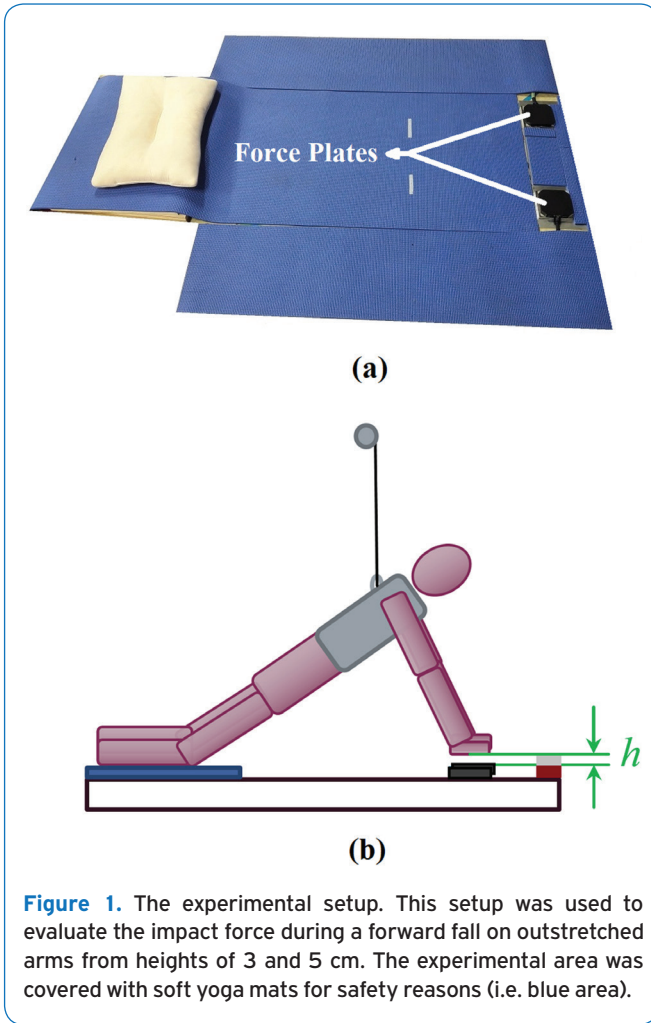


Figure 1. The experimental setup. This setup was used to evaluate the impact force during a forward fall on outstretched arms from heights of 3 and 5 cm. The experimental area was covered with soft yoga mats for safety reasons (i.e. blue area).

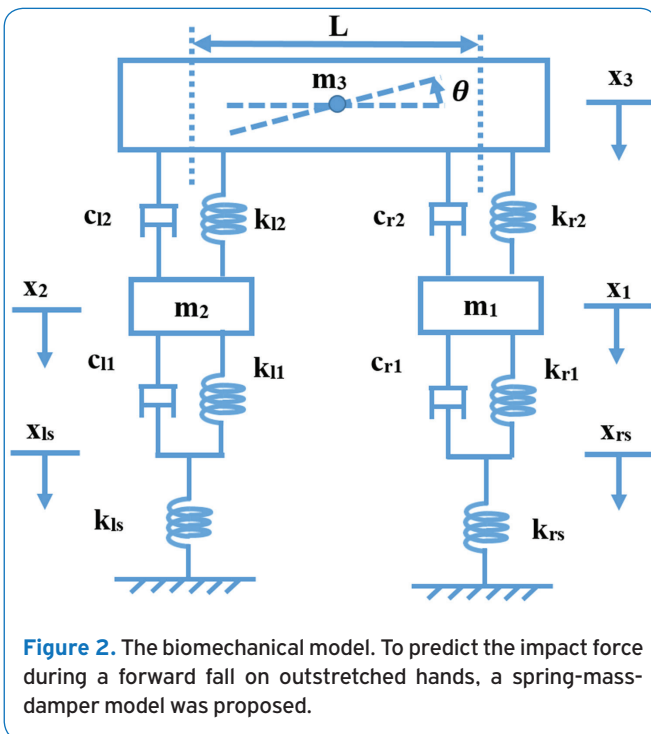


Figure 2. The biomechanical model. To predict the impact force during a forward fall on outstretched hands, a spring-mass-damper model was proposed.

and 5 cm. The height was measured and set as the distance between the palm of the hand and force plates. Each participant completed twelve trials, which involved six trials at $h=3$ cm, 15 minutes of rest, and six experiments at $h=5$ cm.

Biomechanical model

Previously, a simple 2-DOF system was proposed to predict the impact force during forward falls²⁶. The model was proposed based on the assumption that the impact force acting on one hand deteriorates and then the other hand contacts the ground, which means that all the impact force is applied to one hand. The proposed model is not able to distinguish the impact force applied to each hand separately. Specifically, it is not able to evaluate conditions such as the hands striking different types of surfaces or striking the ground at different times.

To overcome the limitations of the above model and predict the impact force during forward falls, a spring-mass-damper model was proposed (see Figure 2). Let us assume that the elbow flexion is negligible during the forward fall. In the figure, m_3 is the torso effective mass and m_1 and m_2 are the effective masses of the right and left upper extremities. k_{r2} and k_{l2} represent the right and left shoulder and torso stiffness. c_{r2} and c_{l2} are the damping of the shoulder and torso. k_{r1} , k_{l1} , c_{r1} , and c_{l1} are the right and left hand, wrist, and palmar tissue stiffness and damping parameters.

The governing equations of motion can be written as:

$$\begin{bmatrix} m_1 & 0 & 0 & 0 & 0 & 0 \\ 0 & m_2 & 0 & 0 & 0 & 0 \\ 0 & 0 & m_3 & 0 & 0 & 0 \\ 0 & 0 & 0 & 1 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \end{bmatrix} \begin{bmatrix} \ddot{x}_1 \\ \ddot{x}_2 \\ \ddot{x}_3 \\ \ddot{\theta} \\ \ddot{x}_{rs} \\ \ddot{x}_{ls} \end{bmatrix} + \begin{bmatrix} -c_{r1}-c_{r2} & 0 & c_{r2} & c_{r2}L/2 & c_{r1} & 0 \\ 0 & -c_{l2}-c_{l1} & c_{l2} & -c_{l2}L/2 & 0 & c_{l1} \\ c_{r2} & c_{l2} & -c_{l2}-c_{r2} & c_{l2}L/2-c_{r2}L/2 & 0 & 0 \\ c_{r2} & -c_{l2} & c_{l2}-c_{r2} & -c_{l2}L/2-c_{r2}L/2 & 0 & 0 \\ -c_{r1} & 0 & 0 & 0 & c_{r1} & 0 \\ -c_{l1} & 0 & 0 & 0 & c_{l1} & 0 \end{bmatrix} \begin{bmatrix} \dot{x}_1 \\ \dot{x}_2 \\ \dot{x}_3 \\ \dot{\theta} \\ \dot{x}_{rs} \\ \dot{x}_{ls} \end{bmatrix} + \begin{bmatrix} -k_{r1}-k_{r2} & 0 & k_{r2} & k_{r2}L/2 & k_{r1} & 0 \\ 0 & -k_{l2}-k_{l1} & k_{l2} & -k_{l2}L/2 & 0 & k_{l1} \\ k_{r2} & k_{l2} & -k_{r2}-k_{r2} & k_{l2}L/2-k_{r2}L/2 & 0 & 0 \\ k_{r2} & -k_{l2} & k_{l2}-k_{r2} & -k_{l2}L/2-k_{r2}L/2 & 0 & 0 \\ -k_{r1} & 0 & 0 & 0 & k_{r1}+k_{rs} & 0 \\ -k_{l1} & 0 & 0 & 0 & k_{l1}+k_{ls} & 0 \end{bmatrix} \begin{bmatrix} x_1 \\ x_2 \\ x_3 \\ \theta \\ x_{rs} \\ x_{ls} \end{bmatrix} = \begin{bmatrix} m_1g \\ m_2g \\ m_3g \\ 0 \\ 0 \\ 0 \end{bmatrix} \quad (1)$$

The equations were solved using the Runge-Kutta 4th order method and MATLAB software (MathWorks, 2015) under the following conditions.

$$\begin{aligned} x_1(0)=0, & & x_2(0)=0, & & x_3(0)=0, \\ \dot{x}_1(0)=v, & & \dot{x}_2(0)=v, & & \dot{x}_3(0)=v, & & v=\sqrt{2gh} \end{aligned} \quad (2)$$

The stiffness and damping parameters were measured beforehand. To obtain similar results but separately for each hand, we used one-half of the values proposed in²⁰. The stiffness of the padding was 160 kN/m, which was measured using the Discovery Hybrid Rheometer (DHR-2-NA, TA Instruments, US).

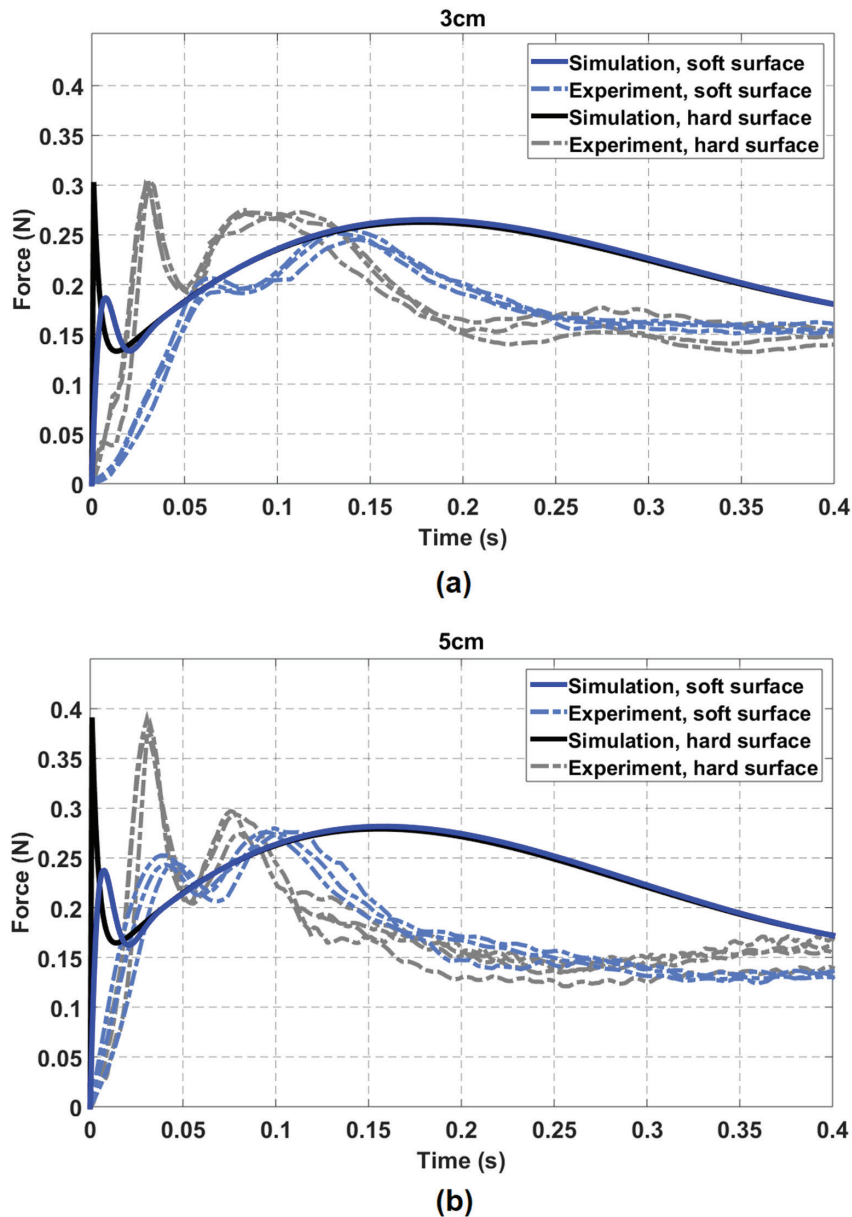


Figure 3. Impact force profile for a forward fall. Simulation and experimental impact forces during forward fall on outstretched arms, where one hand contacts the hard surface and the other contacts the soft surface from heights of (a) 3 cm and (b) 5 cm.

Results

Figure 3 shows the results obtained for one hand contacting the soft surface and the other contacting the hard surface. As in previous studies, two peaks can be recognized. The first peak has a higher frequency and magnitude than the second peak. Increasing the fall height increases the peak impact forces; however, the first peak is amplified more. The first peak for a fall from the 3 cm height on a hard surface is approximately 291 ± 98 N. This force increases to 382 ± 121 N for the 5 cm fall height. In addition, an independent-samples

t-test was conducted to show the effect of the falling height (i.e. 3 cm and 5 cm) on the magnitude of the impact force. The result of statistical analysis (IBM SPSS V.23) confirmed that the magnitude of the impact force increases significantly with increases in the falling height ($p < 0.001$).

Our proposed model can predict the magnitude of the impact force applied to each hand separately; however, as with previously proposed models^{20,22,23}, it cannot correctly estimate the timing of the peak impact forces. Usually, the first peak occurs in a range of 20 to 30 ms after impact; however, the simulation predicts the first peak as occurring immediately after impact. The timing of the second impact

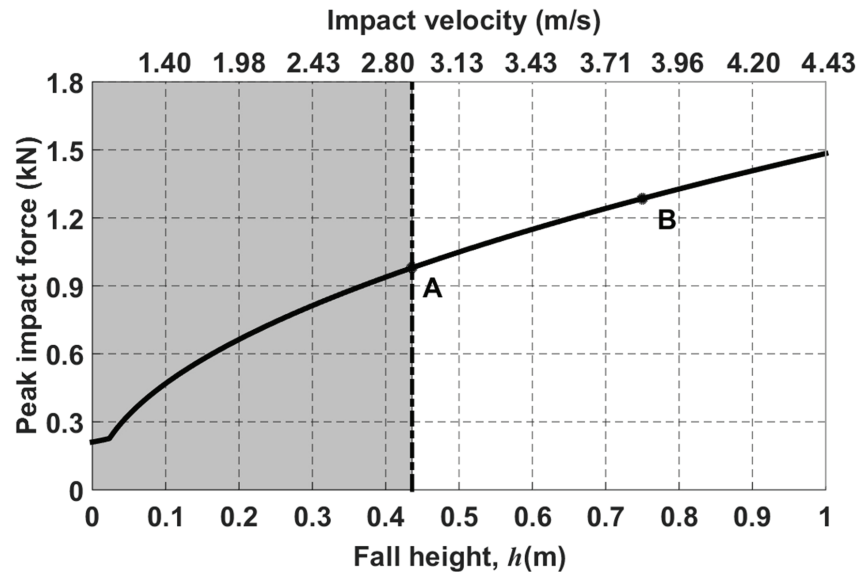


Figure 4. Peak impact force-fall height graph. The peak impact force changes as a function of the fall height; point A shows the border of the safe fall height, and point B indicates the impact force applied to the hand during a forward fall from a standing posture.

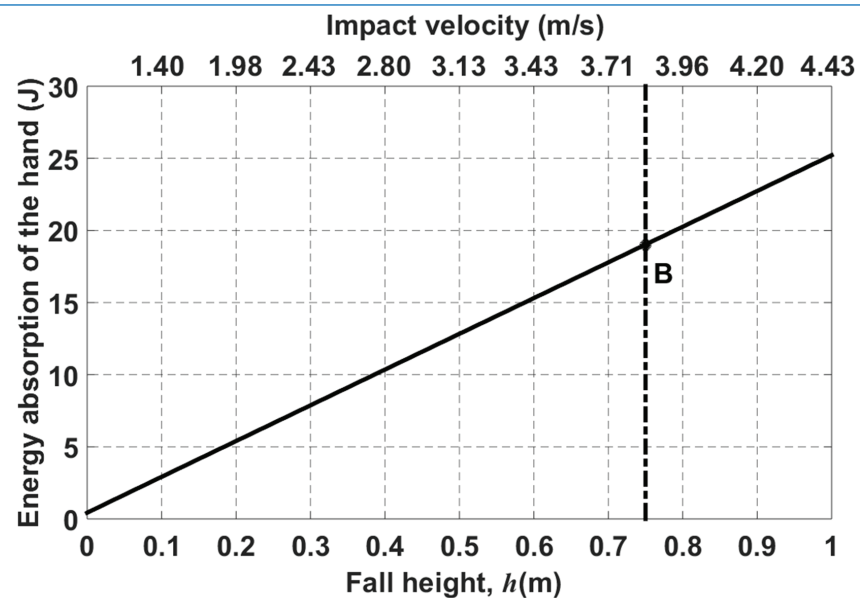


Figure 5. Energy absorption-fall height graph. The energy absorbed by one hand as a function of fall height. The absorbed energy from a walking position is indicated by point B.

force was also over-estimated.

The soft surface affects the first impact force significantly. In contrast, the second peak does not change noticeably. The first peak impact force of falls from 3 and 5 cm heights decreased to 201 ± 81 N and 243 ± 84 N respectively (i.e. approximately 30% and 36% reduction in the first peak impact force compared to the fall on a hard surface for falls from 3 cm ($p < 0.002$) and 5 cm ($p < 0.005$) heights). In

addition, the peaks were delayed and the first impact force occurred in the range of 30 to 60 ms. The statistical analysis also confirmed that the first peak impact force significantly decreased for the heights of 3 cm ($p < 0.001$) and 5 cm ($p < 0.001$) during forward fall on a soft surface compared to a hard surface. A comparison of our results with different fall contact conditions (e.g., both hands contacting a hard surface and both hands contacting a soft surface), reveals that the

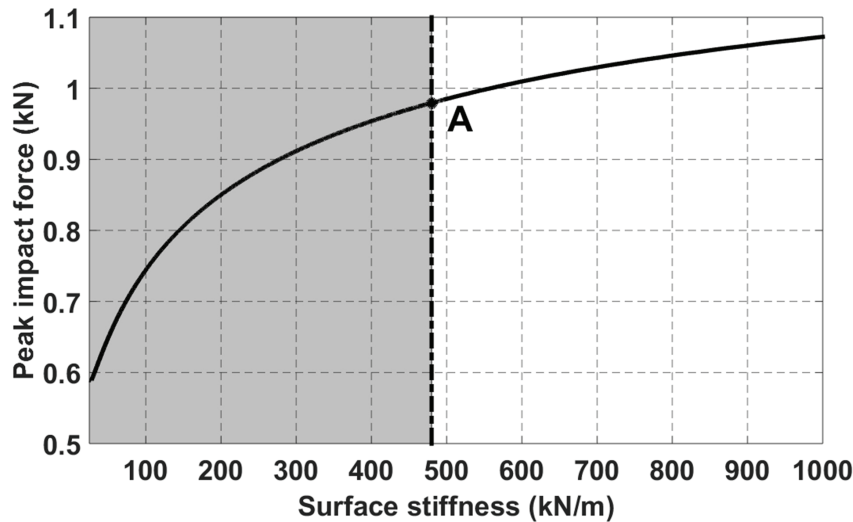


Figure 6. Peak impact force-surface stiffness graph. The effect of surface stiffness on the peak impact force applied to the hand. Point A shows the border of the safe fall height.

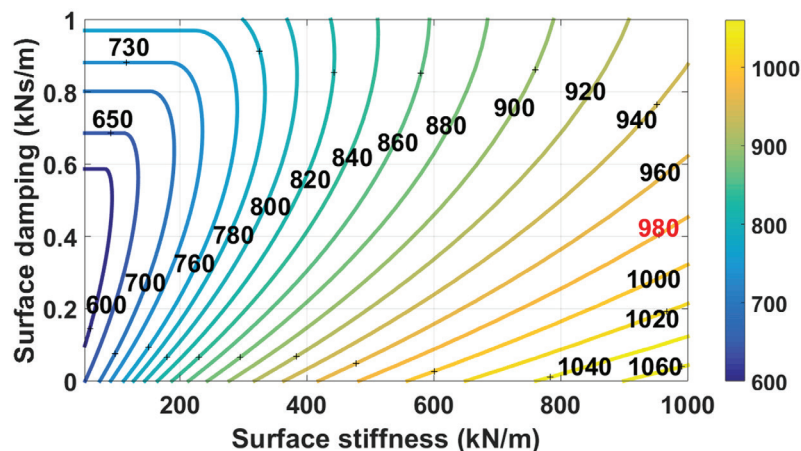


Figure 7. Peak impact force change based on surface stiffness and damping parameters. The effect of a surface with both damping and stiffness elements on the peak impact force during typical forward fall of a human from a height of 75 cm (i.e. standing posture).

magnitude of the peak impact force applied to each hand is independent of the condition of the other hand.

To validate our model, it was used to predict the impact force applied to a hand for forward falls from greater fall heights. Figures 4–7 are associated with the safety issues of forward fall. The presented plots were estimated based on the average characteristics of male and female bodies²⁰. Figure 4 shows how the fall height increment leads to increases in the peak impact force. Several researchers have investigated wrist fractures and reported very different fracture loads. Recent studies²⁹ have shown that the probability of distal radius fracture with maximum impact forces of 551 N, 1053.6 N,

and 1858.2 N are 10%, 25%, and 50%, respectively. A risk order of 10–25% is recommended for design applications. In another study³⁰, a maximum force of 1580 ± 600 N was considered as the fracture force among females. Thus, we specified A as the safe border of the impact force with 980 N in Figure 4. This point reflects a fall height of 0.438 m as the maximum safe falling height. The corresponding impact velocity is 2.93 m/s.

A fall from a standing position is usually considered as a fall from a height approximately one-half of the subject's height^{31,32} or 75 cm²⁰. Point B is associated with the impact force applied to the hand from a walking position with a fall

height of 75 cm. Such a fall posture imposes an impact force on the hand that is equivalent to 1285 N. In cases of higher fall heights, such as 95 cm, the maximum impact force applied to the hand is 1446 N. Consequently, it can be inferred that the risk of fracture for a person falling from a standing/walking position is more than 25% and less than 50%.

The energy absorbed by a hand as a function of fall height is illustrated in Figure 5. The absorbed energy increases with fall height. The energy absorption magnitude can also be considered as another important risk factor for wrist fractures. The absorbed energy for a fall height of 75 cm is approximately 19 J (Point B).

The effect of surface stiffness reduction on impact force attenuation is shown in Figure 6. This graph reflects the fact that the application of compliant flooring reduces the impact force significantly. Point A (indicating 980 N), is responsible for the application of compliant flooring with a stiffness of 480 kN/m.

A compliant surface can be defined not only by stiffness characteristics, but also damping feature. Let us imagine a surface with both stiffness and damping specifications. Figure 7 depicts the contour plot of the peak impact force changes based on the surface stiffness and damping for a typical forward fall from a standing position (i.e. 75 cm fall height). The safe range can be considered as the region under the 980 N line. This line starts from a stiffness of 980 kN/m without any damping characteristics of the surface and ends on the stiffness of 1000 kN/m and 0.45 kNs/m as the damping parameter.

Discussion

The bimanual fall-arresting strategy is the most common strategy among people. In this paper, we proposed a biomechanical model that has two arms to calculate the impact force applied to the hands during forward falls. The model enables us to calculate the applied force to each hand separately, especially when the contacting condition is different, such as striking the hands on two different types of surfaces (i.e. asymmetrical contact type). A set of experiments was designed to validate our model by evaluating the impact force applied to each hand.

Our results show that the impact force acting on the hand consists of two distinct peaks. The frequency and magnitude of the first peak are considerably higher than that of the second. The results show that the compliant surface reduces the first peak value significantly, but does not affect the second peak notably. In addition, the peaks were delayed compared to the fall on the hard surface. A comparison of the results shows that the impact force applied to one hand does not affect that of the other.

In previous studies, only one force plate was used to measure the impact force and the related experiments were conducted under the assumption that the impact force acting on one hand falls to zero and then the other hand contacts the ground. Considering this hypothesis, the impact force

against one hand (e.g., 2.7 kN for a typical fall from standing position²⁰) generates a very high torque that leads to body rotation around the shoulder and the body loses balance. In some of our experiments, one hand struck the ground shortly before the other and the measured impact force was similar to the condition when both hands impacted the ground simultaneously. Even if one hand contacts the ground significantly before the other one, the impact forces applied to the hands show the results similar to the condition where both hands contact the ground at the same time¹⁵. The conclusion stemming from this fact proves that the previous assumption to break a fall using only one hand is not realistic.

We estimated the impact force applied to each hand over the range 1285 N to 1446 N. Based on previous studies, we assumed that 980 N is a safe fracture border. Although the wrist fracture strongly depends on bone strength, the magnitude of the induced impact force significantly affects the possibility of a fracture occurring. Thus, we can assume that the force applied to the hand is strongly associated with a risk of wrist injuries. Thus, based on our results, a forward fall may cause a fracture due to excessive force higher than bone fracture threshold. A soft surface can attenuate the impact force and delay the peak impact forces. Thus, it is desirable to cover the surface with a soft material to prevent injuries in playgrounds and workplaces, where robots collaborate with people any collision between human and robot may result in a fall.

A stiffness component has the capability to receive and store motion energy and then release it over an extended period, consequently resulting in impact force attenuation. A damping element dissipates the energy and causes further impact force reduction. We presented a contour plot that depicts the effect of surface stiffness and damping elements on the first peak impact force. This graph provides information that can be used by researchers to design a soft surface to attenuate the impact force and the severity of injuries. From a design perspective, compliant flooring may have both damping and stiffness characteristics; however, to avoid an increase in the falling risk, the surface should not be excessively soft.

Mounting handrails is an effective means to prevent falls or reduce fall-related injuries. The handrails should be installed at an appropriate height to prevent falls or limit the impact velocity to a safe range. The presented graphs can provide useful information for designers to determine an optimum height for installing handrails.

Conclusion

Realistic assessment of the force acting on the hands during forward falls is necessary to design compliant surfaces to prevent injuries. In this study, we investigated the most common fall-arresting strategy (namely, bimanual fall breaking) under asymmetrical contact and measured the impact force of each hand separately. To address the problem, a series of fall experiments were conducted in

which a force plate was mounted under each hand. Further, to produce a different contact surface type, one of the force plates was covered with a soft padding layer.

The results obtained show that the impact force applied to each hand is independent of the other. The impact force profile consists of two peaks—a high peak followed by a low peak. The experimental and simulation results indicate that the impact force is strongly associated with the fall height (i.e. impact velocity).

A spring–damper–mass model simulating bimanual forward fall arresting, with rigid masses representing the effective mass of body segments, and springs and dampers representing muscles and soft tissues was also developed. The developed model was validated and used to investigate the effect of surface stiffness and damping on the peak impact force during a typical human forward fall. Thus, we were able to determine a range of stiffness and damping values for the surface to prevent forward fall injuries.

The results obtained in this study are useful for designers who wish to produce appropriate compliant flooring to reduce the risk of injuries in places such as playgrounds or nursing homes where fall accidents may occur frequently. From a robotics perspective, this kind of compliant flooring is necessary for human-robot collaboration workplaces, where collisions between humans and robots may result in falls.

Acknowledgement

This work was supported by JSPS KAKENHI Grant Number 26750121 and 30813787, and METI “Strategic international standardization promotion project: International standardization of human tolerance against fall injuries”.

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