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Analysis of Upper Extremity Motion during Trip-Induced Falls

Saeed Abdolshah, *Member, IEEE*, Yasuhiro Akiyama, *Member, IEEE*, Kento Mitsuoka, Yoji Yamada, *Member, IEEE* and Shogo Okamoto

***Abstract*— Forward fall is one of the most common causes of upper extremity fractures. Significant factors influencing impact force and injuries were widely studied; however, it is necessary to investigate the natural reactions of humans during a forward fall to obtain a realistic evaluation of injuries. The purpose of this study was to analyze the natural motion of the upper extremity during an induced trip. We carried out a tripping experiment using an obstacle colliding with one leg; while recovery step was prevented to produce a forward fall. Results showed that the elbow extension had a slight ascending trend during the forward fall and elbow angle at the moment of hand-ground contact was appropriate to reduce the peak force. Landing on the obstacle-side hand was more likely due to body rotation towards the obstacle-side. To prevent injuries, subjects were connected to a safety harness not to strike the ground with high impact velocity. Thus, the fall motion was simulated using a 12 DOF model to obtain a realistic evaluation of the impact velocity and the related impact force caused by the forward fall was estimated using a sagittal 3-segment model. Results of this study can be useful in human-robot collaboration, where a collision between human and robot may cause a forward fall.**

I. Introduction

Forward falls are one of the most common types of falls especially among older people [1], [2]. Some balance recovery strategies have been reported such as elevating, lowering and skip motion [3]. Unsuccessful balance recovery causes injuries such as wrist fracture, which is one the most prevalent body fractures [4] due to high impact loading during the fall. Conducting controlled human fall experiment in a laboratory can be hazardous and imposes high risks on participants. Thus, some researchers carried out experiments to reproduce a part of fall motion (e.g. simulation of the real condition with less intensity) and provide fall motion data such as impact velocity and force. However, such data may strongly depend on experimental conditions such as fall height, initial velocity and etc.

Tan et al [5] designed a tether-release falling experiment to compare the impact velocity in forward and backward falls. The subjects were released while leaning forward and supported by a tether. They were instructed to land on their knees followed by the hands. The average wrist velocity at the time of landing was 1.33 m/s; however, the maximum velocity was 3.57 m/s.

DeGoede et al. [6] designed a similar experiment. The subjects were released while leaning forward and attached to a tether supporting about 30% of their body weight. Their results showed that the impact velocity was 2.6 ± 0.3 m/sec; and it could be reduced using a different falling arrest strategy.

S. Abdolshah, Y. Akiyama, K. Mitsuoka, Y. Yamada and S. Okamoto are with Department of Mechanical Systems Engineering, Nagoya University, Nagoya, Japan (saeed.abdolshah@mech.nagoya-u.ac.jp)

There are a few studies that proposed a model to calculate the impact velocity and the peak force. Some of them validated their models using the experimental results.

Chiu & Robinovitch [7] proposed a model of damped two degrees of freedom spring-mass system to calculate the impact force. They conducted some experiments to evaluate the impact force during forward fall on outstretched arms and validated their model. One limitation of their study was that the elbow extension was ignored. Later they developed their model with an extra spring as the ground stiffness. They set the impact velocity of 3.83 m/sec for a free-fall of the body from a height of 0.75 m [8]. A similar study without body stiffness characteristics was conducted in [9].

DeGoede et al. [10] proposed a three degrees of freedom model with stiffness and damping elements to calculate the impact velocity of the hand. They simulated the forward fall by an experiment of arresting a moving mass by one arm and verified their model. Later, they used computer simulation to investigate age-related effects on forward fall arresting using a 2-D five-segment model [11]. Usually, the impact force profile has two peaks and the fracture occurs by the first one, which is higher in hand-ground contact [12].

Zhou et al. [13] measured the tripping force (e.g. obstacle–foot impact force), and applied the force to swing foot of a 12 degree-of-freedom model. They did not consider separate arms for their model. Lo et al. [14] used a 7 degree-of-freedom model to simulate forward fall motion by controlling body segments. They reported wrist impact velocity in a range of 2.86 to 3.10 m/s. Xu et al. [15] also used a 5 degree-of-freedom model (e.g. the model without arms) and simulated the fall motion using some motion constraints. Wrist injuries during forward and backward falls were also investigated in snowboarding using a computer simulation [16].

The models and experiments clarified that the impact force is strongly correlated with the impact velocity of the hand and the elbow extension. The previous designed experiments were not coincident with conditions of a real trip-induced fall. To make an appropriate evaluation of the forward fall injuries, it is important to investigate the human natural reaction during a forward fall to obtain elbow extension and impact velocity during the impact phase.

In the current study, an experiment was designed to obtain a more realistic pattern of the trip-induced forward fall. We conducted a tripping experiment using an obstacle colliding with one leg; while preventing recovery step. Subjects' reactions were investigated during the falling motion with special attention to the upper extremity. The experimental data provided a natural fall process pattern. The subjects were connected to a safety harness, which does not affect the human reaction, but it can influence the impact velocity. Thus, to calculate the real forward fall impact velocity, a

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multibody 12 DOF model was used to simulate the fall motion. Finally, the impact force was calculated using a 2-dimensional model with damping and stiffness elements.

II. METHODS

A. Subjects

Six healthy young male volunteers participated in this study. The age of the subjects was between 22 and 24 years old and their average heights and weights were 172.4±3.2 cm and 60.0±8.9 kg respectively. None of the participants had received fall arrest training such as Jodo or martial arts techniques, and they did not have any records of illnesses such as neurological disease, balance disorders or falls. The Institutional Review Board of Nagoya University approved all experiment protocols, and all participants provided a written consent form.

B. Experimental Protocol

Motion capture system (MAC3D System, Motion Analysis Corp.) with 10 cameras and 25 reflective markers were used to measure the kinematics of body segments during the fall motion. For safety reasons, a safety harness was attached to the shoulders; a knee cover and ankle joint supporter were provided to the participants. Two load cells (RSCC-200kg, Unipulse Corp.) measured the applied force on the safety harness and the perturbation on the recovery leg. An actuated linear slider was used to set the rope length regarding the obstacle and tripping location. The obstacle was provided with three axis force sensors (USL-08-H6, Tech Gihan Corp.) to determine the trip perturbation timing. The data were recorded by a frequency of 100 Hz. The experiment setup is shown in Fig. 1.

The participants were dressed in tight sportswear and markers were attached to their clothes. The subjects were instructed to walk regarding the specific rhythm produced by an electric metronome and a visual guide. The visual guide was moved by a three-phase induction motor (TO-K, HITACHI Corp.). The walking speed of 1.5 m/s was set by both guides. The subjects were instructed to walk continuously to adjust their walking speed to the guides.

After the gait adjustment trials, the forward fall experiments started. Tripping was randomly induced to the swing leg (right or left one) using a rigid flat bar. Elevating and lowering are the most common recovery strategies [3]. For a successful elevating strategy, the subject should step forward by placing the swing foot anterior to the body. Due to the shape and height of the obstacle in this experiment, the collision occurred between the obstacle and subject's shank. Consequently, the subject did not have the opportunity to step forward by the tripped leg. To achieve a successful lowering strategy, the recovery leg should move forward; however, to prevent the recovery motion, a rope was connected to the stance leg to constraint the recovery step. Such limitation can be considered as reduced lower extremity range of motion that occurs among older adults and is a kind of functional disability [17]. Thus, subjects could not keep the balance using the lowering strategy and the forward fall occurred.

The position of the obstacle for each trip was decided based on the gait motion of subjects. All participants were asked to wear half-covered glasses to avoid detecting the obstacle. A total number of 40 trial experiments were conducted for each subject, where 30 trials were dummy without any obstacles and in other 10 experiments, the obstacle caused the forward fall. The subjects were permitted to place their arms in their desirable configuration. The significant advantage of such marginal number of experiments without delay is the reduction of learning effects. Thus, the subjects could not get accustomed to the experiments.

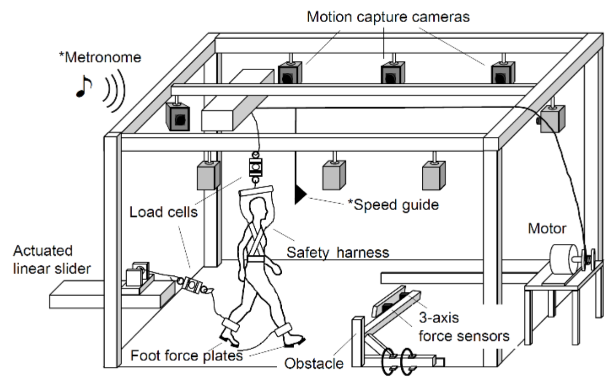

Figure 1. Experimental setup to simulate a trip-induced forward fall. One leg collides with the obstacle, while the other one is attached to a rope to prevent recovery motion. The experiment was equipped with a motion capture system, visual and vocal guide, safety tether and load cells to measure the supporting force and the perturbation on recovery leg.

C. Data Analysis

The whole process of the forward fall in this experiment is divided into three phases:

C. 1. Non-supported phase

The first phase starts when the tripping over the obstacle induces. In this step, the contact between the swing leg and the obstacle occurs at any time during the swing phase and a forward angular momentum of the body, induced by the forward trip causes imbalance. The common recovery strategy is to obtain balance with producing enough backward angular impulse.

The subject makes an effort to balance the body using the other leg, however, the recovery motion is prevented. This phase terminates just before the safety harness supports the body. In other words, during this phase, the force cell on the safety harness does not sense any significant force, which means the body is not supported by the harness. It can be inferred that in non-supported phase, a natural forward fall is in process.

C. 2. Supported phase

The second phase starts when the harness supports the body for safety reasons and finishes just prior to the ground contact. The reaction to the induced trip is not notably affected by the safety harness [18]; however, the harness may support and decelerate the motion. Thus, in this phase, the human reaction during the forward fall is considered to be similar to a real forward fall. Our main focus would be the upper extremity, which is a very important part of the evaluation of impact force and injuries.

The dynamic characteristics of the motion may change because of extra force from the safety harness. Especially the contact velocity of the hand striking the ground is important to calculate the impact force, and we need to regenerate the motion dynamic data.

A 12-DoF model is considered to simulate the second phase motion (Fig. 2). The dynamic equation of the system is generated using Lagrange's equation, which can be expressed as

$$M(q)\ddot{q} + C(q,\dot{q}) + G(q) = \tau, \qquad (1)$$

where M is the inertia matrix; C represents the Coriolis and centrifugal matrix, and G is the gravity matrix and τ represents the joint torques. q denotes the generalized coordinate vectors.

In this step, we regenerate the motion in sagittal plane regarding a free-fall motion using the initial and final conditions of experimental data. The orientation and velocity of the body segments at the end of the non-supported phase are considered as the initial condition; however, the body orientation at the end of the supported phase is the final condition. Some motion constraints were also imposed which were discussed in [14, 15]. The most important result of this step is the contact velocity, which can be used in the next step to calculate the impact force.

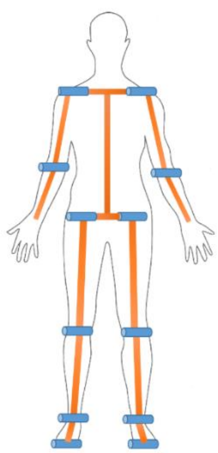

Figure 2. Human model with 12 degrees of freedom to simulate the supported phase in sagittal plane

C. 3. Impact phase

In this phase, hand-ground contact occurs. The collision between heel of the hand with the ground generates a significant instantaneous force. The impact force consists of two local extremes. In case of the hand striking the ground, usually, the first peak is more important and is the main reason of injuries. To calculate the impact force, we used a non-linear model that was proposed in [10]

$$F_{impact} = K_{H-G}|x^3_{Hand}|(1 - B_{H-G}\dot{x}_{Hand}), \qquad (2)$$

where K_{H-G} and B_{H-G} are the stiffness and damping parameters of the hand-ground interface, which are considered as 3,500 kN/m^3, and 6.0 (m/s)$^{-1}$ respectively [10]. x_{Hand}, $\dot{x}_{Hand}$ are the relative deflection and the velocity of the hand into the ground. Initially at the moment of contact, x_{Hand} is zero; however, $\dot{x}_{Hand}$ is the one calculated at the end of the supported phase.

To calculate x_{Hand} and $\dot{x}_{Hand}$, a sagittal 3-segment model [10] is considered as shown in Fig. 3. θ_1(arm angle) is the angle between the horizontal axis and the arm, and θ_2 (forearm angle) is the angle between the forearm and the horizontal axis. Linear shoulder stiffness and damping indices are 3.5 kN/m and 0,12 kNs/m respectively and the rotational elbow stiffness is determined as 2.28 Nm/° [11]. The effective mass can be calculated as [7]

$$M = \frac{F_{static}}{g - m}, \qquad (3)$$

where F_{static} is the force measured by the force plate in static state, when all oscillations damped. The gravitational constant is denoted by g and m is the total mass of hand, arm and forearm. This model was simulated in MSC.visualNastran 4D, 2002.

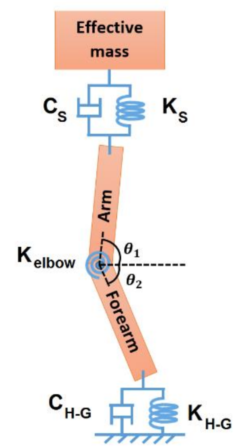

Figure 3. A sagittal 3-segment model of the impact phase

III. Results

A. Supported phase – Body disposition

Fig. 4 shows a typical falling process during the supported phase, where the obstacle struck the left leg and the recovery step of the right leg was prevented. The subject landed first on his hands followed by the knees. Mostly, the participants fell symmetrically without any steps forward or backward; however, in some cases, asymmetrical fall motion was observed. The trips were induced on an average of 32.0 ± 3.7% of gait cycle. The obstacle-side body is marked with circles on the body segments.

The absolute arm angle during the supported phase is illustrated in Fig. 5.a. The initial angle of the obstacle-side arm is higher than the other one. The angles of both arms prior to ground contact are similar and around 75°. At the beginning of the phase, both arms have ascending trends, followed by a descending trend. Arms have the lowest angle at the end of this phase.

Fig. 5.b. shows the absolute forearm angle. Despite the arm, the forearm angle is always ascending. Similar to the arm angle, at the starting point of the supported phase, the recovery-side arm angle has more variation. At the end of the supported phase, the angle of the obstacle-side forearm (82.8°) is higher than the recovery-side forearm (75.7°).

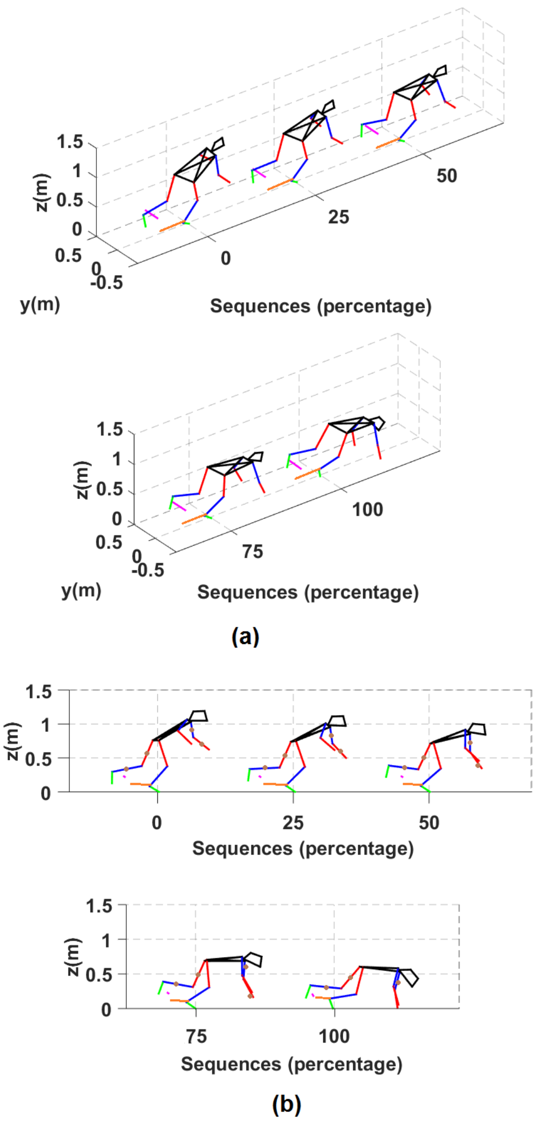

Figure 4. (a) A sample forward fall motion in supported phase, with an obstacle facing the left leg, (b) sagittal plane view of the fall

The elbow extension is shown in Fig. 5.c. The average obstacle-side elbow extension is higher than the other side. The final elbow extension for the obstacle-side is 157.8°; however, it is 150.7° for the recovery-side. The rate of changing elbow extension prior to ground contact is low, which can be beneficial to reduce the peak force [6]. The subjects had not received any training for such elbow extension.

The hand height in the sagittal plane is illustrated in Fig. 6.a. The initial average height is around 0.62 m. The obstacle-side hand has a lower height. Based on this graph, it can be inferred that in average, the obstacle-side hand strikes the ground prior to the other one. Fig. 6.b. shows the center of mass height in the sagittal plane. As expected, the center of mass has a descending trend; and the slope of the trend is less than the hand graph.

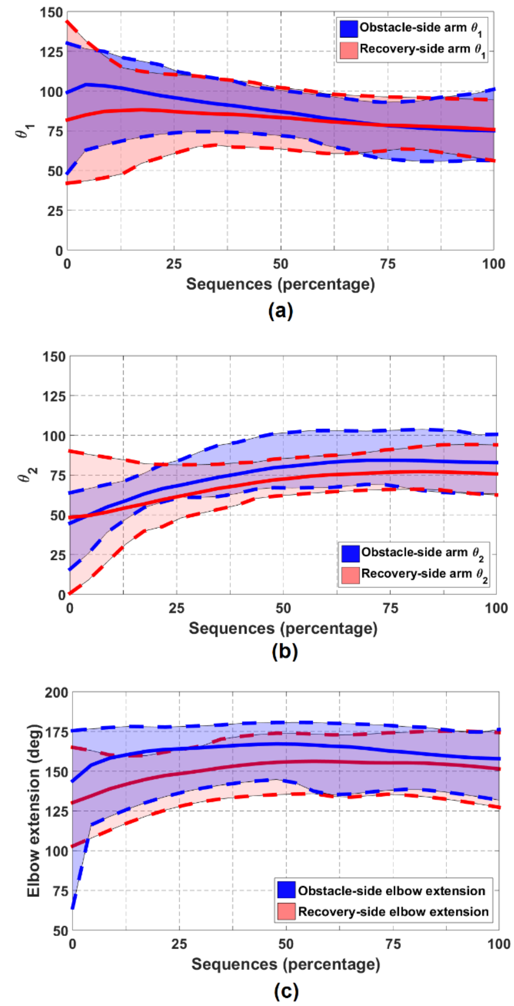

Figure 5. Average, minimum and maximum of (a) arm angle, (b) forearm angle, and (c) elbow extension for obstacle-side arm, and recovery-side arm

B. Supported phase – motion data

The average initial and final condition of body segments were extracted from the experiments. The dynamic model is developed to simulate forward fall process. The most important result of this section is the hand velocity at the end of the supported phase.

Fig. 7.a. shows the regenerated height of the hand during the trip-induced forward fall using the Lagrange's equation. The initial average height is considered about 0.62 m and the related velocity graph is illustrated in Fig. 7.b. The impact velocity is about 3.89 m/s.

C. Impact phase

To calculate the contact force during the forward fall, the model in Fig. 3 was used. The arm and forearm angles were also obtained from the experimental data. The contact force during the impact phase is shown in Figure 8. The first maximum impact force is approximately 2.3 kN. The first peak is related to the high-frequency transient occurring slightly after heel of the hand strike. The second peak occurs after a local minimum and is usually smaller in case of hand collision to the ground.

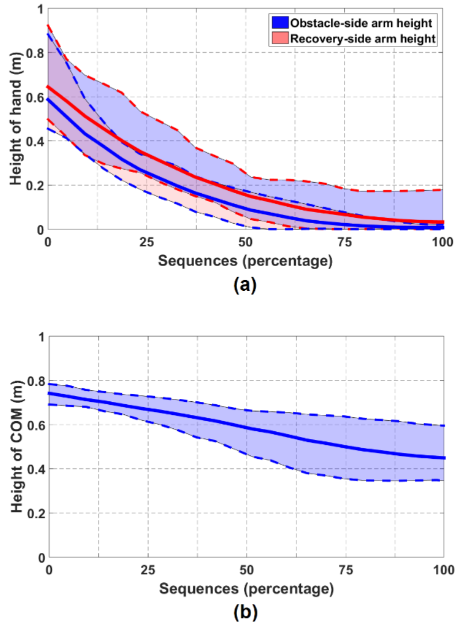

Figure 6. (a) The hand and (b) the COM (center of mass) height from the ground level

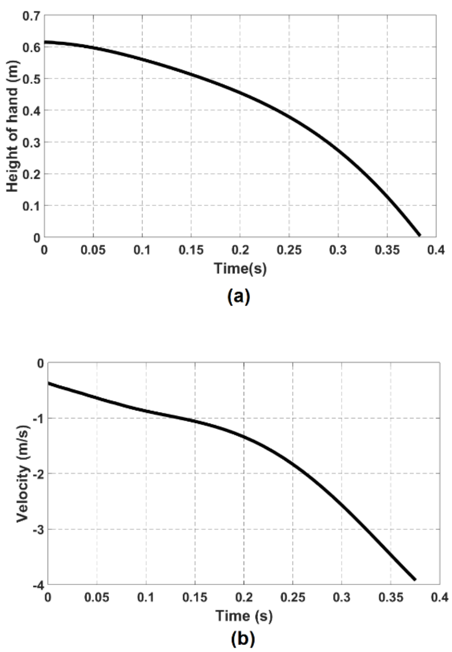

Figure 7. (a) The position and (b) velocity of the hand from the ground level in vertical direction of the sagittal plane

IV. Discussion

This study was designed to investigate the natural reaction against the forward fall. We have limited the scope of our current investigation to upper extremity motion during a trip-induced fall. Due to difficulty of conducting such experiments and the injuries may occur for volunteers, we limited the experiments to six young subjects and did not use elderly. These experiments were enough to obtain a reasonable range of results. The elbow extension was 157.8° for the obstacle-side and 150.7° for the recovery-side. DeGoede et al. [6] reported an average of 168° from the experiments of five subjects. In their study, the subjects were aware of the fall upon the release of the supporting tether. Thus, their prior intention may have influenced the elbow angle. Moreover the subjects had less time to react the forward fall.

Results show that during the falling process, the obstacle-side hand has lower height compared with the recovery-side hand. Therefore, it is more probable that the obstacle-side hand strikes the ground prior to the other hand. The reason is related to the usual attitude of the recovery motion. The subjects rotate the body towards the obstacle-side [19] and the rotation axis is the inclined stance leg. Consequently, the obstacle-side hand has lower height during the falling process, however, our experiments show that the subjects decrease the height difference between hands prior to ground contact. The other reason can be the different elbow angles. In average, the obstacle-side elbow has higher extension compared with the recovery-side elbow as shown in Fig. 5.c.

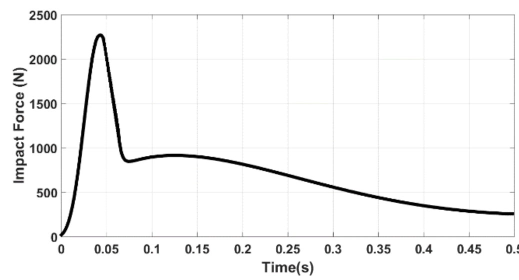

Figure 8. The ground reaction force during the hand-ground contact

The higher elbow extension in obstacle-side may occur as a natural reaction to control the balance. The difference between the elbow extensions may cause diverse impact forces on left and right sides. To the best of the author's knowledge, no previous work discussed the influence of such different elbow angles. All proposed analytical models to calculate the impact force were 2-dimensional, and such difference in elbow angles has not been considered yet.

The average elbow extension and the rate of changing the elbow angle, were both appropriate to reduce the peak force applied to the distal forearm. The subjects chose this configuration of the arm and forearm arbitrarily and without any training. The elbow angle in this study was lower than the natural elbow angle in [6]. One reason can be the type of their experiments, which was a controlled laboratory experiment with a lower falling height. The other reason may be the differences between the types of falls. In their experiments, the subjects did not face any obstacles and the fall occurred by releasing a supporting tether (i.e. No external force applied to the swing leg). To confirm our results, it is necessary to re-conduct the experiment using people with different ages and races. The impact velocity was calculated as 3.89 m/s. In another study [8], it was calculated using the assumption of free-fall motion from 0.75 m height. They evaluated the impact velocity as 3.83 m/s. In this study, the first impact force was calculated as 2.3 kN. This value can be considered as the natural first impact force of forward falling induced by tripping.

V. CONCLUSION

The main purpose of this study was to investigate natural forward fall pattern. Elbow angle and impact velocity play important roles in intensity of impact force and should be taken into consideration to evaluate fall injuries. Most of the fall studies conducted some experiments to reproduce a part of real fall motion; however, we carried out lifelike fall experiments to obtain the human reaction against forward fall without any special instructions. We have constrained the scope of our study to a forward fall arrested with both hands; which is a quite common strategy. Results may not be applicable to the other fall arrest strategies or different types of falls such as sideways fall.

The outcome shows that the arm angle reduces during the fall process; however, the forearm angle increases and the elbow extension has a slight ascending trend. The average elbow extension at the moment of hand-ground contact was about 154.25°; however, the obstacle-side elbow extension was higher (157.8°). The responses to the trip were not eminently influenced by the safety tether. Thus the elbow angle can be considered as the natural elbow angle against forward fall without any practice or instruction. This angle has a significant role in the intensity of the impact force and strangely, in these experiments, the average elbow angle was in a range that reduces the impact force.

Furthermore, results show that the obstacle-side hand has lower height during the falling process. Therefore, the obstacle-side hand may strike the ground prior to the other side. This may occur due to twisting the body towards the obstacle-side with the inclined stance leg as the rotation axis. One more reason can be the higher elbow extension angle in the obstacle-side. The subjects decreased the height difference between hands prior to ground contact.

Due to the application of the safety harness, the forward fall dynamic data could not be used and we regenerated the motion using Lagrange's equation. We used the experimental data of the initial and final conditions of body segments and the motion constraints to calculate the impact velocity in a free-fall motion. The impact velocity was about 3.89 m/s, which is not very different from the previous studies [7]. The related impact force was also estimated regarding the average elbow extension and the impact velocity, which was about 2.3 kN. The results of this study can be useful in human-robot collaboration, where a collision between human and robot may cause a forward fall.

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