Effect of Upper Extremity Impact Strategy on Energy Distribution Between Elbow Joint and Shoulder Joint in Forward Falls

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Abstract

In a forward fall, the majority of the impact energy is absorbed by the elbow joint and shoulder joint. This study examines the effect of the impact strategy on the energy absorption distribution between the two joints during the impact phase of a forward fall. Twenty healthy young male subjects with an average age of 24 years participated in a series of forward fall experiments. The kinematics and kinetics of the upper extremity and the impact forces at the elbow joint and shoulder joint are investigated for three impact strategies, namely elbow dominant, intermediate, and shoulder dominant. The energy absorption ratio and pain score of the elbow dominant group are significantly lower (Energy absorption ratio: p = 0.011, pain score: p = 0.012) than the corresponding values of the intermediate and shoulder dominant groups. The low energy absorption ratio of the elbow dominant group indicates a more uniform distribution of the impact energy between the elbow joint and the shoulder joint. This implies that elbow flexion provides a beneficial damping effect during impact, and therefore reduces the energy absorbed at the shoulder joint. Overall, the results suggest that the elbow dominant impact strategy is optimal for forward falls. The results can aid the development of an effective impact strategy for minimizing the risk of upper extremity injuries due to forward falls.

Keywords: Forward fall, Energy absorption ratio, Elbow joint, Shoulder joint, Upper extremity

1. Introduction

Outstretching the hand to arrest a fall is an instinctive reaction to prevent impact to sensitive parts of the body, such as the head, cervical spine, and hips. As a result, falls onto an outstretched hand are the leading cause of upper extremity injuries. Fall-related elbow injuries include distal radius, humeral neck, and supracondylar fractures [1-4]. Fall-related shoulder injuries include acromio-clavicular joint dislocation, gleno-humeral joint dislocation, proximal humerus fracture, clavicle fracture, scapular fracture, rotator cuff tear, and superior labrum anterior to posterior lesion [5-7].

Chou et al. [8] compared the elbow loads induced in forward falls performed with elbow flexion and elbow full extension models, respectively. The results showed that the elbow valgus-varus shear force was 68% lower (p = 0.002) in the elbow flexion model. In addition, it was shown that elbow flexion not only reduces the intensity of the initial peak impact force, but also delays the occurrence of the second maximum peak. Therefore, the authors suggested that elbow flexion provides an effective damping mechanism in forward falls, and consequently minimizes the risk of upper extremity injuries. The effect of various forearm rotation postures on the elbow load and elbow flexion angle in a forward fall was investigated in another study [9]. It was shown that falls with the forearm internally rotated resulted in both a greater elbow flexion angle (i.e., 40.3°) and a lower valgus-varus shear force (i.e., 4.3% of body weight) than those of falls with the forearm externally rotated. These results show that the energy dissipation resulting from elbow flexion increases with increasing elbow flexion angle. Thus, elbow flexion plays an important role in

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minimizing the risk of impact injury in forward falls. In a previous study on the effect of posture on shoulder joint loading during a fall, Hsu et al. [10] found that the shoulder medio-lateral shear forces in the externally rotated forearm group were 1.61 and 2.94 times higher than those in the non-rotated and internally rotated forearm groups, respectively. The shoulder flexion angles in the externally rotated, non-rotated, and internally rotated forearm groups were 0.6°, 8.0°, and 19.2°, respectively, and the corresponding shoulder abduction angles were 6.1°, 34.1°, and 46.3°, respectively.

Chiu and Robinovitch [11] found that fall heights greater than 0.6 m carry significant risk for wrist fracture and that the shoulder absorbs the majority of impact energy during a fall. Sran et al. [12] proposed that a reasonable goal in arresting a fall with the upper extremities is to absorb sufficient energy to prevent impact to the head and reduce to safe levels the residual energy that must be absorbed through contact by the trunk and pelvis.

In practice, the impact energy associated with a forward fall is absorbed not only by the elbow joint [9], but also by the shoulder joint [10,11]. Therefore, in developing effective impact strategies for forward falls, it is necessary to determine the distribution of the impact energy absorption between the elbow joint and the shoulder joint for various impact strategies. Accordingly, in the present study, a series of simulated forward fall experiments were conducted under three impact strategies, namely elbow dominant, intermediate, and shoulder dominant. The corresponding kinematics and kinetics of the upper extremity, and the impact forces acting on the elbow and shoulder joints were compared to determine which impact strategy yields the most uniform distribution of the impact energy between the elbow joint and the shoulder joint.

2. Methods

2.1 Subjects and experimental protocol

Twenty male subjects volunteered to participate in the study. The subjects were 24.5 ± 1.9 (22–29) years of age, 65.7 ± 6.9 (57–82) kg in weight, and 170 ± 4 (164–176) cm in height. The subjects were all right-hand-dominant and free of any musculoskeletal disorders of the upper extremity. The experimental procedure was approved by the Orthopedic and Rehabilitation Research Center, National Cheng Kung University, Taiwan, and by the National Science Council of Taiwan. Furthermore, each participant signed a consent-release agreement before taking part in the experiments. To determine the effect of the impact strategy on the distribution of the impact energy between the elbow joint and the shoulder joint, the subjects were asked to perform simulated falls using three impact strategies, namely Strategy 1: Elbow Dominant; Strategy 2: Intermediate; and Strategy 3: Shoulder Dominant. Three trials were conducted for each strategy. Note that before performing each fall, the subjects were requested to select one of the impact strategies at random. To utilize the Elbow Dominant Strategy, the subjects were instructed to flex the elbow joint slightly at the moment of impact. For the Shoulder Dominant Strategy, the subjects were instructed to flex the shoulder joint slightly upon impact. Finally, for the Intermediate Strategy, the subjects were instructed to avoid deliberate flexion of either the elbow joint or the shoulder joint. As shown in Fig. 1, the subjects were positioned with their knees in contact with the ground and were then raised with the assistance of a suspension system such that the distance between their outstretched hands and the ground was equal to 5 cm. The subjects were instructed to keep their elbows in full extension with a 45° internal forearm rotation prior to impact and were then dropped such that they fell onto both hands. After each fall, the subject’s sensation of pain was recorded using a questionnaire with a pain index ranging from 0 (no pain) to 10 (extreme pain). To evaluate the relative effectiveness of the three impact strategies, the distribution of the impact energy between the shoulder joint and the elbow joint was quantified following each fall using the following energy absorption ratio:

$$r_s = \frac{\text{Total energy absorbed by shoulder}}{\text{Total energy absorbed by elbow}}$$

None of the subjects had any prior training in effective fall techniques (e.g., as part of baseball, football, or general sports training). As a result, it was impossible to be certain that the chosen impact strategy for each fall was correctly implemented. Thus, once each individual had completed all the simulated falls, the corresponding energy absorption ratios were divided into three groups; the lowest energy absorption ratio ($r_s = 1.476$) was defined as the Elbow Dominant Group (EDG), the middle energy absorption ratio ($r_s = 1.709$) was defined as the Intermediate Group (IG), and the highest energy absorption ratio ($r_s = 1.994$) was defined as the Shoulder Dominant Group (SDG).

![Figure 1](image_url)

Figure 1. (a) Lateral and (b) frontal view of experimental setup. Subjects were placed with knees in contact with the ground and with their forearm internally rotated by 45°. Then, with the help of a suspension system, the subjects were dropped from a height of 5 cm (i.e., the distance between the outstretched hand and the force plate) such that they fell onto both hands.

Prior to the experiments, eleven reflective markers were placed on selected anatomical landmarks of each subject. The landmarks were selected in accordance with rigid body assumptions for the trunk (cervical vertebra 7, thoracic vertebra 4, and acromion), upper arm (acromion process, medial, and lateral epicondyles of the elbow), forearm (medial and lateral epicondyles of the elbow, ulnar styloid process), and hand (radial and ulnar styloid processes, third metacarpal bone). In
addition, a triangular frame carrying three markers was placed on the upper arm in order to minimize the risk of experimental error caused by the movement of skin over the epicondyles during a fall. In analyzing the kinematics of the upper extremity, the center of the shoulder joint was defined as a point located at 90% along the length of a virtual line commencing at the center of the elbow joint and terminating at the acromion marker [13]. The relative joint positions and ground reaction forces during the simulated fall were measured using an ExpertVision motion system (Motion Analysis Corp., Santa Rosa, CA, USA) incorporating six cameras (with a sampling rate of 240 Hz) and a single force plate (with a sampling rate of 1000 Hz) (Type 9281B, Kistler Instrument Corp., Winterthur, Switzerland).

The experiments were physically supervised by the author or one of the coauthors (each with a sports medicine background) in order to avoid the risk of accidental injuries.

2.2 Theorem and governing equations

In analyzing the elbow and shoulder joint forces produced under the three fall strategies, the upper extremity was modeled as a three-joint multi-linkage system comprising of the hand, forearm, and upper arm. The free-body diagrams of the three joints (i.e., the wrist, elbow, and shoulder) are shown in Fig. 2. The governing equations of the joint forces, moments, and energies are derived as follows.

From the free-body diagram of the hand:

\[
\mathbf{F}_{\text{wp}} = m_i \dot{\mathbf{a}}_i - m_i \mathbf{g} - \mathbf{F}_{\text{wp}}
\]

(2)

\[
\mathbf{M}_{\text{wp}} = I_i \dot{\mathbf{a}}_i - \mathbf{M}_{\text{wp}} - (\mathbf{I}_i \times \mathbf{F}_{\text{wp}}) + \dot{\mathbf{v}}_i \times (I_i \cdot \dot{\mathbf{a}}_i)
\]

(3)

\[
\mathbf{E}_\mathbf{p} = \int_{h-p} \mathbf{M}_{\text{wp}} \cdot d\mathbf{\theta}
\]

(4)

From the free-body diagram of the forearm:

\[
\mathbf{F}_{\text{wp}} = -\mathbf{F}_{\text{wp}}
\]

(5)

\[
\mathbf{M}_{\text{wp}} = -\mathbf{M}_{\text{wp}}
\]

(6)

\[
\mathbf{F}_{\text{wp}} = m_i \dot{\mathbf{a}}_i - m_i \mathbf{g} - \mathbf{F}_{\text{wp}}
\]

(7)

\[
\mathbf{M}_{\text{wp}} = I_i \dot{\mathbf{a}}_i - \mathbf{M}_{\text{wp}} - (\mathbf{I}_i \times \mathbf{F}_{\text{wp}}) + \dot{\mathbf{v}}_i \times (I_i \cdot \dot{\mathbf{a}}_i)
\]

(8)

\[
\mathbf{E}_\mathbf{p} = \int_{h-p} \mathbf{M}_{\text{wp}} \cdot d\mathbf{\theta}
\]

(9)

From the free-body diagram of the upper arm:

\[
\mathbf{F}_{\text{wp}} = -\mathbf{F}_{\text{wp}}
\]

(10)

\[
\mathbf{M}_{\text{wp}} = -\mathbf{M}_{\text{wp}}
\]

(11)

\[
\mathbf{F}_{\text{wp}} = m_i \dot{\mathbf{a}}_i - m_i \mathbf{g} - \mathbf{F}_{\text{wp}}
\]

(12)

\[
\mathbf{M}_{\text{wp}} = I_i \dot{\mathbf{a}}_i - \mathbf{M}_{\text{wp}} - (\mathbf{I}_i \times \mathbf{F}_{\text{wp}}) + \dot{\mathbf{v}}_i \times (I_i \cdot \dot{\mathbf{a}}_i)
\]

(13)

\[
\mathbf{E}_\mathbf{p} = \int_{h-p} \mathbf{M}_{\text{wp}} \cdot d\mathbf{\theta}
\]

(14)

Note that in Eqs. (1)-(13), \( h, f, \) and \( u \) represent the hand, forearm, and upper arm, respectively. The remaining notations are defined as follows:

- \( \mathbf{F}_p \)-proximal joint force
- \( m_i \)-segment mass
- \( \mathbf{M}_p \)-proximal joint moment
- \( f \)-mass moment of inertia
- \( \dot{\mathbf{a}} \)-angular acceleration of local segment
- \( \dot{\mathbf{M}} \)-proximal joint moment
- \( \mathbf{V}_p \)-rotation matrix describing relative rotation between local coordinates of proximal segment and global coordinates
- \( \dot{\mathbf{V}}_p \)-rotation matrix describing relative rotation between local coordinates of distal segment and global coordinates
- \( \dot{\theta} \)-angular velocity of local segment
- \( E \)-energy of joint
- \( \theta \)-angular displacement of local segment

The kinematics and kinetics data were obtained directly from the experimental measurements. In accordance with Newton’s third law, the governing equations of the joint forces, moments, and energies were solved under the assumption that the force and moment acting on a given joint are equal and opposite to those acting on the proximal joint.

![Figure 2. Segmental free body diagram of upper extremity (Chou et al. [3]).](image-url)
segment mass and inertial data were estimated via anthropometry [19], and the angular velocity and acceleration were calculated using Euler’s parametric method [14]. The forces acting on the hand were assumed to be equal and opposite to those acting on the force plate. Based on the segmental free-body diagram of the upper extremity shown in Fig. 2, the wrist loading and wrist moment were calculated using an inverse dynamic procedure based on the Newton-Euler equations [14-16]. Finally, the elbow and shoulder joint loadings were calculated directly from the governing equations given in Section 2.2.

Data smoothing was performed using a generalized cross-validation spline smoothing routine with a cutoff frequency of 6 Hz [20]. The joint angles and joint forces at the elbow and shoulder were then calculated as functions of time over the impact period and were used for further analysis [8,16,20,21].

2.4 Data analysis

The one-way analysis of variance with repeated measures was used to test for any statistical differences among the mean values of the energy absorption ratio, pain score, and joint absorption energy under the three impact strategies. The tests were performed using SPSS 13.0 software (SPSS Inc., Chicago, Illinois, USA) with a significance level of 0.05. A post-hoc test was then performed using the Bonferroni method in order to identify any significant differences among the three fall strategies in terms of their impact on the energy absorption ratio, pain score, and joint absorption energy.

3. Results

3.1 Energy absorption ratio (r) and joint energy

After each fall, the shoulder absorption energy and elbow absorption energy were substituted into Eq. (1) to calculate the energy absorption ratio. The energy absorption ratio (r) values acquired for all of the trials were found to vary between 1.26 and 2.20. The range r = 1.26–1.57 corresponded to the EDG and the range r = 1.58–1.81 corresponded to the IG and the range r = 1.82–2.20 corresponded to the SDG. The mean pain scores for the EDG, IG, and SDG were 4.90, 5.95, and 6.85, respectively. As shown in Table 1, the pain score of the EDG is significantly lower than those of the IG and the SDG (p < 0.001).

Table 1. Energy absorption ratio and pain score values for three impact strategies.

<table>
<thead>
<tr>
<th>Strategy 1</th>
<th>Strategy 2</th>
<th>Strategy 3</th>
<th>Post-hoc</th>
</tr>
</thead>
<tbody>
<tr>
<td>EDG mean (std)</td>
<td>IG mean (std)</td>
<td>SDG mean (std)</td>
<td>p*</td>
</tr>
<tr>
<td>Energy absorption ratio r</td>
<td>1.476 (0.091)</td>
<td>1.709 (0.112)</td>
<td>1.994 (0.112)</td>
</tr>
<tr>
<td>Pain score</td>
<td>4.90 (0.995)</td>
<td>5.95 (0.605)</td>
<td>6.85 (0.587)</td>
</tr>
</tbody>
</table>

# p value is significant in one-way analysis of variance
* p < 0.05
T refers to the time at the start of fall
T* refers to the time at the end of fall

Table 2. Joint-absorbed energy values for three impact strategies from T1 to T2

<table>
<thead>
<tr>
<th>Joint</th>
<th>Strategy 1</th>
<th>Strategy 2</th>
<th>Strategy 3</th>
<th>p*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elbow</td>
<td>24.12 (5.99)</td>
<td>20.98 (4.33)</td>
<td>19.55 (3.57)</td>
<td>0.011* EDG &gt; SDG</td>
</tr>
<tr>
<td>Shoulder</td>
<td>35.78 (3.96)</td>
<td>35.99 (5.51)</td>
<td>39.97 (6.40)</td>
<td>0.027* EDG &gt; SDG IG &lt; SDG</td>
</tr>
</tbody>
</table>

# p value is significant in one-way analysis of variance
* p < 0.05

3.2 Results

The forward fall used here is very similar to the up-to-down movement during push-ups. The kinematics, kinetics, and range of motion of the joints of the upper extremity during push-ups have been extensively studied (e.g., see [21] and [10]), and are thus not the focus in the present study.

4. Discussion

Arresting a fall with an outstretched hand is an instinctive protective strategy for avoiding injuries to sensitive parts of the body, such as the head, cervical spine, and hips. Many studies have investigated the effect of elbow flexion on the elbow joint load in forward falls [8]. As for the effect of forearm rotation posture, Chou et al. [9] found that the valgus-varus shear forces in the 45° externally rotated forearm group were 1.4 times greater than those in the non-rotated forearm group, and 2.7 times greater than those in the 45° internally rotated forearm group. The elbow joint remained at almost full extension (3.9°) in the externally rotated forearm group, whereas elbow flexion was observed in the non-rotated forearm (24.6°) and internally rotated forearm (40.3°) groups. However, the problem of quantifying the absorbed energy in forward falls has received little attention. Sran et al. [12] calculated the total
energy absorbed by the upper extremity in forward falls by integrating the area under the hand contact force vs. arm deflection trace. In the present study, the impact energy absorbed at the elbow joint and shoulder joint, respectively, was calculated directly from the experimental data. The relative effectiveness of each fall strategy in distributing the impact energy between the two joints was then evaluated by computing the energy absorption ratio defined in Eq. (1).

As shown in Table 3, the percentages of the total energy absorbed at the elbow and shoulder joints are 40.27% and 59.73%, respectively, in the EDG; 36.83% and 63.17%, respectively, in the IG; and 32.85% and 67.15%, respectively, in the SDG. For all fall strategies, the energy absorbed at the shoulder joint is higher than that absorbed at the elbow joint. In other words, the total amount of impact energy absorbed as a result of shoulder flexion is higher than that absorbed by elbow flexion. This suggests that in addition to the damping mechanism provided by elbow flexion upon impact [8,9], shoulder flexion provides a beneficial damping effect [10].

The results presented in Table 1 show that the mean values of the energy absorption ratio are 1.476, 1.709, and 1.994 and the mean pain scores are 4.90, 5.95, and 6.85 in the EDG, IG, and SDG falls, respectively. The results indicate that the pain score increases with increasing percentage of the total impact energy absorbed at the shoulder joint (i.e., increasing energy absorption ratio). The pain score decreases with increasing percentage of the total impact energy absorbed at the elbow joint (Pearson’s r = 0.441, p = 0.000). Consequently, the elbow dominant impact strategy results in the lowest energy absorption ratio and the lowest pain score. This strategy improves the distribution of the impact energy between the elbow joint and the shoulder joint, and therefore reduces the pain induced by impact.

The results show that in falls from a height of 5 cm, the elbow dominant strategy ($r_g = 1.476$) is the most effective in terms of minimizing the pain experienced upon impact. This finding may be useful for sports coaches and instructors in developing improved fall arrest strategies for minimizing the risk of fall-related injuries.

As in previous related studies [8-10], the subjects in the present study were dropped from a height of just 5 cm in order to minimize the risk of injury during simulated falls. However, in real life, the fall height is likely to be much larger. Thus, the limited fall height is a potential limitation of the present study. In practice, fall height may affect the joint loading of the upper extremity, the ground reaction force, and the joint force during falls onto an outstretched hand [8]. Previous studies have shown that the shoulder joint can absorb more impact energy than the elbow joint due to its larger cross-sectional area [11,22]. It is reasonable to expect that the amount of energy absorbed by the shoulder joint relative to that absorbed by the elbow joint increases with increasing fall height. Nonetheless, it is unclear whether the greater amount of energy absorbed by the shoulder joint leads to a corresponding increase in the pain score. Thus, the correlation between the fall height, the energy absorption ratio, and the pain score in forward falls onto an outstretched hand should be addressed in a future study.

5. Conclusion

This study investigated the effect of three impact strategies on the energy absorption distribution between the elbow joint and the shoulder joint during forward falls onto an outstretched hand. The results show thatPath for all fall strategies tested, the energy absorbed at the shoulder joint is higher than that absorbed at the elbow joint. In other words, the total amount of impact energy absorbed as a result of shoulder flexion is larger than that absorbed as a result of elbow flexion. However, the mean pain score for the shoulder dominant strategy was 1.39 and 1.15 times higher than those of the elbow dominant and intermediate strategies, respectively. The pain score increases with increasing percentage of the total energy absorbed at the shoulder joint. Overall, the results show that in falls from a limited height (5 cm), the elbow dominant strategy is the most effective in terms of reducing the pain induced by impact. This finding is relevant to clinicians and sports coaches and instructors in devising effective fall arrest strategies that could minimize the risk of upper extremity injuries.

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References
