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Clinical Biomechanics 16 (2001) 888–894

CLINICAL BIOMECHANICS

www.elsevier.com/locate/clinbiomech

Effect of elbow flexion on upper extremity impact forces during a fall

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Received 22 August 2000; accepted 31 August 2001

Abstract

Objective. The overall objectives are to develop a biomechanical model for a simulated fall with outstretched hand.

Design. Cross-sectional study involving young healthy volunteers in a university research laboratory setting.

Background. Little is known about the factors which influence fracture risk during a fall on outstretched hand.

Methods. A group of 11 male subjects volunteered for this investigation. A set of eight reflective markers was placed bilaterally on selected anatomic landmarks. Subjects were suspended with both elbows extended and wrists dorsiflexed, preparing to impact the ground and force plates from two different fall heights: 3 and 6 cm. Two different postures for the elbows were employed. In the elbow extension experiment, the elbows were extended at all times. In the elbow flexion experiment, the elbows were extended at impact, but then flexed immediately, as though in the initial downward phase of a push-up exercise.

Result. Increasing the fall height significantly increased the upper extremity axial forces by 10% and 5%. No significantly different differences were found in the axial forces applied to the wrist, elbow or shoulder between the elbow flexion and elbow extension trials, but the elbow mediolateral shear force was 68% larger ($P = 0.002$) in the extension trials.

Conclusions. Performing an elbow flexion movement at impact reduces the first peak impact force value and postpones the maximum peak value. Although changing the fall arrest strategy from elbow extension to elbow flexion did not affect the peak impact force on the hand, it did require substantially greater elbow and shoulder muscle strengths.

Relevance

This paper yields insights into how the physical demands of arresting mild falls may relate to upper extremity muscle capacity, joint dislocation and bony fracture.

Keywords: Biomechanics; Forward falls; Impact; Bone fracture

1. Introduction

Accidental slips and falls in daily life, sports activity [1–4] or the occupational environment [5] can result in trauma and functional disability of the upper extremities. As the population of the elderly and athletes increases, the incidence of trauma and arthritis of the upper extremity also increases [6]. These upper extremity injuries include approximately 90% of fractures at the distal radius, humeral neck and supracondylar region of the elbow [7]. In order to better prevent and manage these upper extremity injuries, there is a need to further understand the mechanisms of joint disorder in the shoulder, elbow and wrist joints.

Approximately 75% of all fractures sustained by children occur in the upper extremities and frequently occur during a fall onto an outstretched hand. The majority of these injuries involve the wrist and forearm, but the elbow alone accounts for approximately 10% of all fractures in children [8]. Other clinical studies have also shown that falling on an outstretched hand plays an important role in the upper extremity trauma [9]. It is very important to understand the factors which influence fracture risk during a fall on the outstretched hand.

Robinovitch and Chiu [10,11] studied upper extremity impact force during falls on the outstretched hand. In their study, they measured impact forces during low-

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height (0–5 cm) forward falls onto the outstretched hand, and found that these were governed by an initial high-frequency peak ($F_{\max 1}$) and a subsequent lower-frequency oscillation ($F_{\max 2}$). This phenomenon was well simulated by a two degree-of-freedom lumped-parameter mathematical model. Increases in body mass caused greater increases in the peak magnitude of $F_{\max 2}$ than $F_{\max 1}$. However, increases in fall heights had greater influence on $F_{\max 1}$, which exceeded $F_{\max 2}$ from all but a very low height. Model predictions suggest that fall heights greater than 0.6 m result in significant risk for wrist fracture, since above this height, peak forces surpass the average fracture force of the distal radius. Finally, while the shoulder experiences lower peak force than the wrist, it undergoes considerably greater deflection, and thereby absorbs the majority of impact energy during a fall.

The overall objectives of this study are to develop a biomechanical testing model for a simulated fall situation including two different fall heights and two elbow postures. We investigated the effect of two different fall arrest strategies on upper extremity loading in a fall. This first strategy consisted of flexing the elbow slightly immediately upon impact. The second strategy consisted of maintaining elbow extension at impact. We tested the hypotheses that (a) fall height and (b) fall arrest strategy would not significantly affect the ground reaction impact magnitude or the peak joint moments required to arrest the fall.

2. Methods

2.1. Subject and experimental protocol

Eleven physically healthy male graduate students volunteered for this investigation. They ranged from 20 to 30 yr (mean 26.1, SD 2.6) of age, from 55 to 85 kg (mean 69.3, SD 9.2) in body weight, and from 164 to 181 cm (mean 171.7, SD 5.2) in body height. They had no history of previous upper-extremity injuries or disorders, and all were functionally right-hand dominant.

The ExpertVison motion system (Motion Analysis, Santa Rosa, CA, USA) with six 120 Hz cameras and two 1000 Hz Kistler force-plates (Type 9281B, Kistler Instrument, Winterhur, Switzerland) was used to measure relative joint positions and ground reaction forces.

A set of eight reflective markers was placed on selected anatomic landmarks on the subject. The selected anatomic landmarks were intended to simulate the rigid body assumption for 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 processes), and hand (radial and ulnar styloid processes, third metacarpal bone), as shown in Fig. 1. In addition, a triangular frame with three markers was placed on the upper arm in order to minimize the potential errors due to skin movement of epicondyles during fall experiment. The shoulder joint center was defined by position starting from the elbow joint center, calculated by medial and lateral markers, to nearly 90% of the length from the elbow center to the marker at acromion [12].

In order to analyze the impact forces on bilateral hands and to investigate the stiffness and damping effects on the upper extremity upon impact, subjects were asked to perform two different actions at the moment of impact. The first action was to flex the elbow directly and the second one was to keep the elbow extended. In the flexion experiment, the subjects were asked to spontaneously flex the elbow after the moment of impact. This action was very similar to a flexion motion during a push-up. In the extension experiment, the subjects were asked to keep the elbow at near full extension upon impact. Each action was repeated twice with different fall heights, once at 3 cm and once at 6 cm. Three assumptions were made in this experiment:

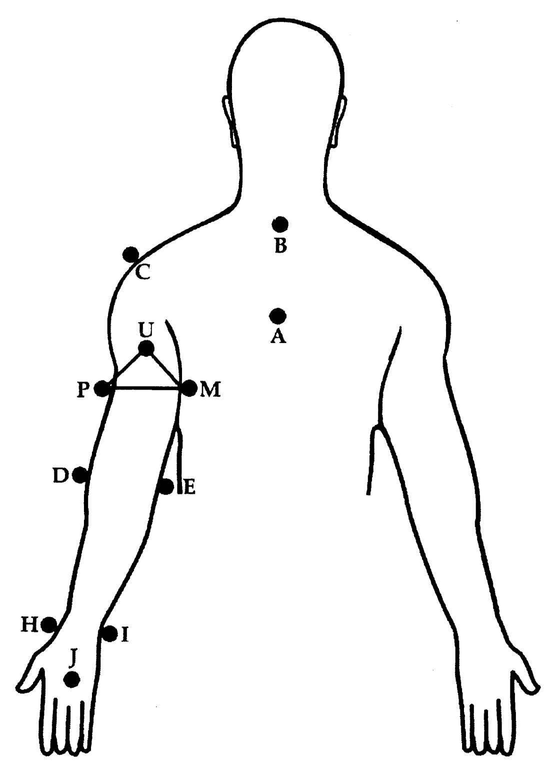

Fig. 1. Markers set up in this experiment. The selected anatomic landmarks were intended to simulate the rigid body assumption for trunk (cervical vertebra 7, thoracic vertebra 4 and acromion), upper arm (acromion process, medial and lateral epicondyles of the elbow), forearm (medial and lateral epicondyles of the elbow, ulnar styloid processes), and hand (radial and ulnar styloid processes, third metacarpla bone).

(1) each upper extremity segment is assumed to be a rigid body with constant density; (2) each joint is assumed to have three degrees of rotational freedom; (3) air friction and rope/pulley friction (of the safety restraint system) are neglected. The testing was approved by Orthopaedic and Rehabilitation Research Center, National Cheng-Kung University and by the National Science Council, Taiwan, for human subject testing and the consent for this fall experiment was signed before performing the testing. Also, it required, at least for the author or one of the co-authors with special sport medicine background to stand by for the potential risk in the experiment.

2.2. Data reduction

Laboratory-developed kinematics and kinetics software were used to calculate the joint angles, resultant forces and moments of the upper extremity. A three-segment model, i.e. hand forearm and upperarm, was employed in the analysis. Each segment was assumed to be a rigid body. Six CCD cameras were used to record the 3-D position of the markers. Three joint angles, hinge angle, rotational angle and horizontal deviation, were calculated using Euler's method with a y–x–z rotational sequence based on the attached markers. A piezoelectric force plate was used to measure vertical and two shear forces as well as the location of the center of pressure on the palm and the moment about the axis normal to the force plate during the fall experiment. Simultaneous measurement of the upper-extremity kinematics was obtained by video recording of the markers. Segment mass and inertia data were estimated by anthropometry [13]. Angular velocity and acceleration were calculated with Euler parameter's method [14]. The force plate loading equals the hand loading with a reversed vector. The wrist loading is then calculated using an inverse dynamic procedure with the Newton–Euler equations [14] [15]. The equations are available from the authors. Then the loading of the joints is determined. A generalized cross-validation spline smoothing (GCVSPL) routine at a cutoff frequency of 6 and 15 Hz was used for marker and force plate data smoothing [16].

Table 1
Mean (SD) time events (in seconds)

Action Fall height	Extend elbow 3 cm	Extend elbow 6 cm	Flex Elbow 3 cm	Flex Elbow 6 cm
T_{12}	0.05 (0.014)	0.05 (0.013)	0.05 (0.012)	0.05 (0.013)
T_{23}	0.08 (0.019)	0.07 (0.023)	0.05 (0.012)	0.06 (0.011)
T_{34}	0.08 (0.021)	0.07 (0.022)	0.04 (0.008)	0.06 (0.009)

Table 2
Mean (SD) measured (GRF) and calculated forces (in % body weight) for each fall height and arrest strategy

	Action			
	Extend elbow	Extend elbow	Flex Elbow	Flex Elbow
Fall height	3 cm	6 cm	3 cm	6 cm
GRF (vertical)	57.2 (8.8)	59.3 (13.2)	57.0 (11.6)	61.1 (11.3)
Calculated joint force				
Wrist				
Axial force	49.4 (12.8)	54.5 (11.8)	52.7 (11.4)	55.5 (10.9)
A/P shear force[a]	12.6 (10.3)	8.2 (3.4)	9.8 (3.9)	11.2 (3.6)
M/L shear force[b]	5.4 (2.2)	7.3 (1.5)	6.5 (3.7)	6.5 (2.4)
Elbow				
Axial force	49.2 (7.3)	50.4 (9.5)	48.6 (10.2)	52.1 (9.5)
A/P shear force [a]	8.3 (2.1)	6.3 (0.3)	4.7 (1.6)	5.5 (0.9)
M/L shear force [b]	6.2 (3.6)	7.2 (5.7)	10.3 (3.0)	10.1 (3.1)
Shoulder				
Axial force	38.9 (8.0)	38.7 (12.1)	38.2 (13.2)	42.6 (11.2)
A/P shear force [a]	24.1 (11.0)	20.6 (12.0)	32.6 (6.5)	33.1 (3.7)
M/L shear force[b]	7.5 (2.5)	8.3 (5.8)	13.1 (5.0)	13.1 (4.7)

[a] Anterial/posterial. Positive values mean anterior force and negative values mean posterior force.
[b] Medial/lateral. Positive values mean lateral force and negative values mean medial force.

2.3. Data analysis

Descriptive statistics were calculated for the ground reaction force, joint forces and joint moments. The ANOVA was used to test the hypotheses, and two-sided paired t-test to test the direction of any fall height or arrest strategy differences. Statistical analysis utilized the computer software SPSS 7.0 for Windows with $P < 0.05$ as statistical significance.

3. Results

3.1. Effect of fall height

3.1.1. Ground reaction force

As shown in Table 2, the maximum impact force was significantly affected by the fall height ($P = 0.006$). As the fall height increased, the impact force likewise increased.

3.2. Joint force

At the moment of impact, the main loading forces of wrist, elbow and shoulder joints were axial forces. The fall height significantly affected the axial forces of wrist and elbow joints ($P = 0.004$ for wrist and $P = 0.023$ for elbow). The axial force of shoulder joint was not significantly affected by the fall height.

For the shoulder joints, both anteroposterior and mediolateral shear forces were significantly affected by the action. Fall height did not significantly affect the shear forces of shoulder, elbow and wrist joints (Table 2).

3.3. Joint moment

As shown in Table 3, the moments in wrist joint were not significantly affected by both fall height and action. The fall height did not significantly affect the moments in both elbow and shoulder joints.

3.4. Effect of arrest strategy

3.4.1. Ground reaction force

For all the volunteers, a high-frequency peak (point 2) and a lower-frequency peak (point 4) as shown in Fig. 2 characterized the time history of the ground reaction forces of the extension experiments. After point 4, the impact force decayed slowly to a nearly constant value. In order to describe the ground reaction forces of the flexion experiments, we defined the time interval from point 1 to point 2 as T_{12}, and the time interval from point 3 to point 4 as T_{34}. For both 3 and 6 cm falls, T_{12} was about 0.04–0.065 s and T_{23} was about 0.05–0.1 s. There was a variation in the moment when the elbow began flexion. Furthermore, the ground reaction data of the flexion experiments were categorized into two types:

(1) *No flexion delay*: the impact force data of this first type were shown in Fig. 3. In this case, the volunteers flexed the elbow in the T_{12} time interval. The first peak (point 2) decreased relative to Fig. 2, and the maximum peak was produced at the time of the second peak (point 4). In order to observe the curve easily, the main curve range was zoomed out. The time interval T_{12} was about 0.04–0.065 s (Table 1).

(2) *Elbow flexion was delayed to the T_{23} time interval*: the impact force data of this second type were shown in

Table 3
Mean (SD) calculated joint moments (Nm) for each fall height and arrest strategy

	Action			
	Extend elbow	Extend elbow	Flex Elbow	Flex Elbow
Fall height	3 cm	6 cm	3 cm	6 cm
Wrist				
Extension[a]	5.4 (2.4)	5.2 (2.5)	6.6 (2.0)	6.4 (2.4)
Abduction[b]	2.4 (2.3)	2.5 (2.0)	3.0 (2.2)	2.9 (1.9)
External rotation[c]	−5.0 (4.3)	−7.0 (3.3)	−7.3 (3.3)	−7.0 (3.6)
Elbow				
Extension[a]	−13.4 (6.4)	−10.7 (3.8)	−21.3 (7.6)	−21.5 (6.1)
Abduction[b]	10.5 (3.2)	8.4 (4.3)	10.5 (5.5)	10.6 (3.8)
External rotation[c]	4.7 (2.6)	4.4 (1.8)	8.0 (3.2)	6.8 (3.5)
Shoulder				
Extension[a]	35.8 (15.2)	32.8 (17.1)	47.7 (12.6)	47.3 (13.4)
Abduction[b]	−4.0 (1.9)	−3.9 (3.7)	−15.8 (7.7)	−12.4 (6.5)
External rotation[c]	4.0 (3.0)	2.5 (1.8)	8.2 (3.5)	7.4 (2.8)

[a] Positive values mean extension moment and negative values mean flexion moment.
[b] Positive values mean abduction moment and negative values mean adduction moment.
[c] Positive values mean external rotation moment and negative values internal rotation moment.

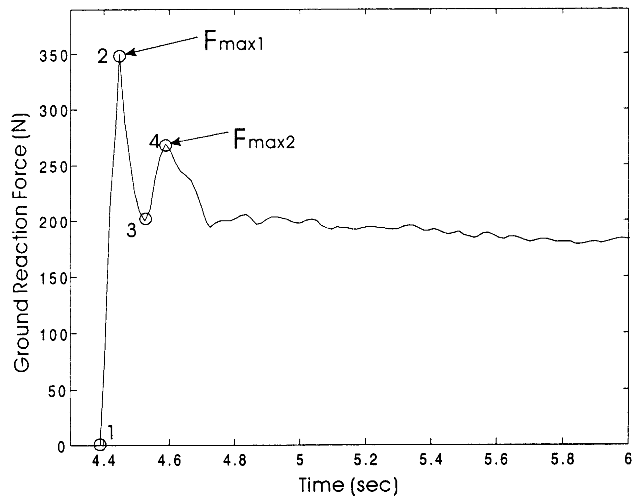

Fig. 2. The ground reaction force data of a male subject (560 N) during extension experiment.

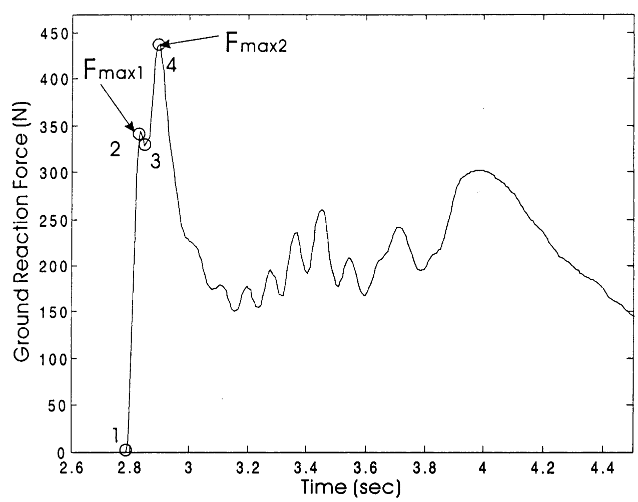

Fig. 3. Example of the first type of ground reaction force time history (for a 700 N male subject falling from 6 cm, and using the flexion strategy).

Fig. 4. The maximum impact force occurred at the first peak (point 2). The time interval T_{12} was about 0.04–0.065 s (Table 1).

The biggest difference between these two patterns was the onset time of the maximum impact force. In the first type, the maximum peak value occurred at the second peak (point 4), while in the second type, the maximum peak value occurred at the first peak (point 2).

The action did not significantly affect the maximum impact force.

3.5. Joint forces

The actions did not significantly affect the axial forces of shoulder, elbow and wrist joints (Table 2). For the

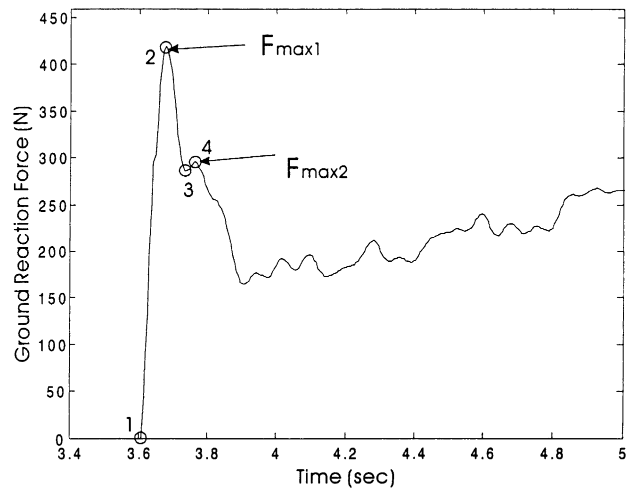

Fig. 4. Example of the second type of ground reaction force time history (for a 680 N male subject falling from 6 cm, and using the flexion strategy).

elbow joints, the arrest strategy significantly affected the anteroposterior force component, and the elbow mediolateral shear force was 68% larger ($P = 0.002$) in the extension trials.

3.6. Joint moments

In the elbow flexion actions, the range of moments in the elbow and the shoulder was much greater than that in the elbow extension actions. The action significantly affected the maximum flexion and pronation moments in elbow joint ($P = 0.027$ for flexion moment and $P = 0.041$ for pronation moment). The joint moments were greater in elbow-flexion action than in elbow-extension action. The maximum flexion moment of elbow joint increased by a factor of 78%, and the maximum pronation moment of the elbow joint increased by a factor of 62%. The action significantly affected the extension, adduction and external rotation moments in the shoulder joint ($P = 0.014$ for extension moment, $P = 0.011$ for all the adduction moment and $P = 0.019$ for the external rotation moment). The moments of shoulder joint in elbow-flexion action were greater than in the elbow-extension action. The maximum extension moment of the shoulder increased by a factor of 39%. The maximum adduction moment of the shoulder joint increased by a factor of 360%, and the maximum lateral rotation moment of shoulder joint increased by a factor of 240%.

4. Discussions

By the year 2030, the number of individuals 65 yr and over will reach 70 million in the US alone; persons 85 yr and older will be fastest growing segment of the population [17]. Injury surveillance data collected in Melbourne also suggest an increasingly important contribution by rollerblading to the pattern of injury seen in young people [1]. As the population of the elderly and athletes increases, the incidence of trauma and arthritis of the upper extremity also increases [6].

In order to better understand the fall mechanism, this experiment was designed to study the kinematics around the wrist, elbow, and shoulder joints during impact. Neglecting the motion of the elbow joint, Robinovitch and Chiu's two-segment model predicted the risk factor for wrist fracture during fall [10]. However, the elbow alone accounted for approximately 10% of all fractures in children. In this study, the effect of the elbow motion, as seen in flexion experiment, on the kinematics of the wrist, elbow, and shoulder during impact was explored.

This is the first study to take the effect of elbow flexion into consideration. The results showed that the action of elbow flexion could decrease the maximal axial force of elbow and delay the time of peak, thus it can provide enough time to adjust and avoid the injury. In comparison with the two spring and mass model proposed by Robinovitch and Chiu, our study hypothesized a three-segment model, taking the flexion of elbow into consideration. From the results of this study, it can be seen that the first reaction impulse, with a time period from hand/ground contact to the first peak ($F_{\max 1}$), decreased and the time of $F_{\max 2}$ was delayed if the elbow flexed upon hand/ground impact. The possibility of upper extremity injury is thereby significantly reduced. This is because the musculoskeletal system can absorb more impulse by delaying the time of the $F_{\max 2}$ and decreasing the value of the $F_{\max 1}$. This could be explained by the additional damping effect caused by the flexion of elbow. It should be noted that without elbow flexion action upon hand/ground impact, our study results are similar to Robinovitch and Chiu's two-spring and mass model.

Possible error effects in this study could be derived from the equipment used in our experiments. Force plate systems measure with reference to the center of pressure. Up to 30 mm error of the calculated center of pressure has been reported, according to the manufactures [18]. Video motion analysis system errors can be referred to the joint center. Six to eleven mm differences are typical when ball markers are used during normal gait [19]. Thus, the various calculations based on force place center of pressure and marker position are subject to large error. In McConville's report [20], the joint moment in low extremity joints changed 13% on the average during normal gait, with an average 10 mm shift in the center of pressure. Joint moment error of up to 11% would result from a 10 mm shift in the center of pressure in our model.

We conclude that elbow flexion at impact results in less axial upper extremity force and delays the maximum ground reaction force. Thus, if possible, such action should be taken when landing from a fall onto the hands. Further studies of falls onto outstretched arms are needed in order to better understand how upper extremity injuries are related to such falls.

Acknowledgement

The authors thank National Science Council on Taiwan for grant support (NSC 89-2320-B-037-032-M08).

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