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IMPACT SEVERITY DURING A BACKWARD FALL DEPENDS ON THE TIMING OF THE "SQUAT" PROTECTIVE RESPONSE DURING DESCENT

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INTRODUCTION

One's risk for injury in the event of a fall depends largely on the position and velocity of the body segments at the moment of impact [1]. Previous studies have shown that impact velocity (and kinetic energy) during a fall can be affected strongly by energy absorption in the lower extremity muscles during descent, as occurs during squatting or sitting [2]. This is achieved through a synergistic pattern of torques and rotations at the ankles, knees, and hips, which we shall refer to as the "squat response." The effectiveness of the squat response in reducing impact severity should depend not only on variables related to strength and flexibility (i.e., attainable magnitudes of joint torque and rotation), but also on how quickly the response is initiated during descent. In the present study, we conducted backward falling experiments to test this hypothesis.

MATERIALS AND METHODS

Seven healthy subjects, ranging in age between 19 and 26 yrs, participated in the study. Each provided written informed consent, and the experiment was approved by the Committee on Human Research at the University of California, San Francisco.

During the trials, subjects stood barefoot on a raised walkway (Figure 1). A 30 cm thick gymnastic mat was located flush with the surface of the walkway directly behind the subject's heels. A six-camera motion analysis system (Qualisys Inc., Glastonbury, CT) having a measurement accuracy of approximately 1.0 mm recorded the 3-dimensional positions of retro-reflective foam markers located at the ankles (lateral malleoli), knees (femoral lateral epicondyles), greater trochanters, junction of L5/S1 vertebrae, shoulders (acromion processes), elbows (lateral humeral epicondyles), wrists (junction between the radius and ulna), and top of the head. Reaction forces on the subject's right foot were measured from a force platform of surface area 0.4m x 0.6m (model 4060H, Bertec Corp., Worthington, OH) mounted flush to the surface of the walkway. Motion data were sampled at 60 Hz, and force data were sampled synchronously at 960 Hz.

Each subject completed three series of falls, which differed in the body configuration at the onset of descent. In series 1, subjects voluntarily initiated their descent from a standing position (lean angle of 0°). In series 2 and 3, subjects were released from a static lean angle of 5° (series 2) or 12° (series 3) by means of an inextensible tether that attached at one end to an electromagnet and at the other end to a chest harness worn by the subject (Figure 1). Five repeated trials were conducted in each series. Before each trial, the subject was instructed to maintain their arms crossed over their chest and their feet stationary during descent, and "to land on the mat as softly as possible."

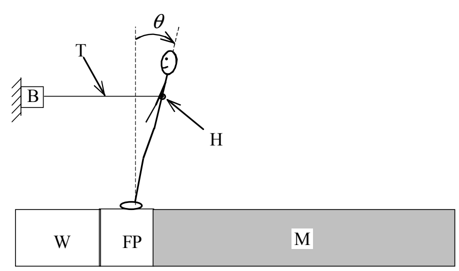

Figure 1. Tether-release falling experiment.

For each trial, custom routines developed in MATLAB (The MathWorks, Natick, MA) were used to filter and differentiate position data, and calculate linear and angular rotations, velocities and accelerations of the various body segments. The kinetic energy of each body segment was calculated based on anthropometric relations obtained by Dempster [3]. An inverse-dynamics routine was used to calculate sagittal-plane torques at the hip, knee, and ankle. Energy-absorbing work at the ankle, knee, and hip during descent (the sum of which we shall refer to with the symbol W_{tot}) was calculated by numerically integrating joint torque-rotation traces. Impact severity was defined by the vertical translational component of the body's total kinetic energy (KEz) and the vertical component of hip velocity (Vz_{hip})

at the instant the pelvis contacted the mat. We used repeated-measures analysis of variance (ANOVA) to determine whether these variables differed between the three series.

RESULTS

Significant differences existed between the series in KE_z ($p < 0.001$), Vz_{hip} ($p < 0.001$), and W_{tot} ($p < 0.001$). When compared to series 1 trials, impact velocities averaged 47% higher in series 2 trials, and 62% higher in series 3 trials (Table 1). Similarly, when compared to series 1 trials, kinetic energies averaged 171% higher in series 2 trials, and 256% higher in series 3 trials. Total joint energy absorption averaged 45% lower in series 2 trials than in series 1 trials, and 59% lower in series 3 trials than in series 1 trials. The latter related to the observation of a gradual reduction between successive series (series 1 > series 2 > series 3) in the magnitude of ankle dorsiflexion and knee and hip flexion rotations, and eccentric knee and hip extensor torques.

Table 1 Average values (± one S.D.) for dependent measures.

Variable	Series 1 (0° lean)	Series 2 (5° lean)	Series 3 (12° lean)
KE_z (J)	66.1 ± 37.3	179.1 ± 52.3	231.6 ± 43.8
Vz_{hip} (m/sec)	1.86 ± 0.47	2.73 ± 0.24	3.02 ± 0.14
W_{tot} (J)	138.9 ± 66.3	76.5 ± 49.0	57.6 ± 40.3

DISCUSSION

Our results suggest that the effectiveness of the squat response in reducing impact severity during a fall depends on the time that it is initiated during descent. Increases in the initial lean angle of the body at the onset of descent led to a reduction in joint energy absorption, and a subsequent increase in impact energy. Based on potential energy considerations alone, one would expect the opposite trend, since the change in potential energy during descent was greatest in series 1. This demonstrates the potentially dominant effect of protective responses on impact severity during a fall.

There are several limitations to this study. We focused on a single fall protective response, and examined its efficacy in reducing impact severity during only backward falls. Furthermore, we simulated different "reaction time" delays in the onset of fall protective responses by varying the initial angle of the body during tether-release experiments. We admit that this is a highly idealized model of a real-life fall. At the same time, we can think of no reason why the basic mechanism we observed would not apply in real life, and regard these results as providing strong evidence for the importance of reaction time (as well as strength and flexibility) in determining one's ability to avoid injury in the event of a fall.

ACKNOWLEDGEMENTS

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