Practical Method for Forensic Testing of Fall Impact Effects on the Human Spine

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The purpose of this study was to develop a safe, inexpensive, and practical method for estimating the force on the human spine during simulated fall landings for forensic investigations. Although the injury tolerances for the spine are known, the mechanics of spinal loading during fall landing are not well-established and thus limit the use of injury tolerances in establishing risk in landing injury cases. Measuring these spinal forces directly using live subjects would be invasive and unsafe; cadaver and animal models are expensive and do not capture the true mechanics of human spinal loading. As an alternative, this study proposed the use of a heavy duty punching bag to simulate the semi-rigid properties of the human body during a fall landing. The heavy bag was instrumented with accelerometers and dropped from various heights to measure impact acceleration. A live test subject instrumented with accelerometers on the foot and low back performed feet-first fall landings at relatively low heights to establish data for comparison. The impact acceleration for the heavy bag and back acceleration of the live subject matched well at corresponding heights. The measured foot accelerations did not correlate directly with those of the bag, possibly due to additional limb acceleration during landing. These preliminary results suggest that a heavy bag could be used to estimate the acceleration experienced by the back during simulated fall landings in forensic investigations.

 DOI 10.1177/1541931213571120

INTRODUCTION

This preliminary study sought to investigate if an instrumented punching bag could be used as a practical method to safely estimate back accelerations during feet-first fall landings. The motivation for this research originated from forensic investigation of an injury case involving a fall landing with disputed circumstances, resulting in compression fracture of the T12 vertebra. In order to conduct a biomechanical investigation of the claim, it was necessary to estimate the accelerations at the back for the various scenarios, to assess the likelihood of injury for each scenario.

Spine Anatomy

Human spinal anatomy is exceedingly complex and presents an enormous challenge for forensic engineers attempting to simulate or reconstruct accidents. The spine is comprised of seven cervical, twelve thoracic and five lumbar vertebrae, as shown in Figure 1, all held in place by an intricate system of tissue, ligaments and muscles.

The spine is an integral component in load bearing and weight transfer through the body. Compressive forces in the spine are transferred from the intervertebral discs to the vertebral endplates and are distributed through the thin trabeculae of cancellous bone within the vertebral bodies (Myers & Wilson, 1997).

Fall Landing Biomechanics

When the feet contact the ground during a feet-first landing, there is a ground reaction force (GRF) applied to the body through the feet. The vertical GRF has been measured and can range from 1.0 to 14.0 times of body weight for normal subjects (Caulfield & Garrett, 2004; Decker et al., 2002; James et al., 2003). The GRF increases with landing height (Zhang et al., 2000) and with the stiffness of the landing surface (Devita & Skelly, 1992). Other factors that influence the GRF include landing posture and technique (Eloranta, 1996; Kovacs et al., 1999).

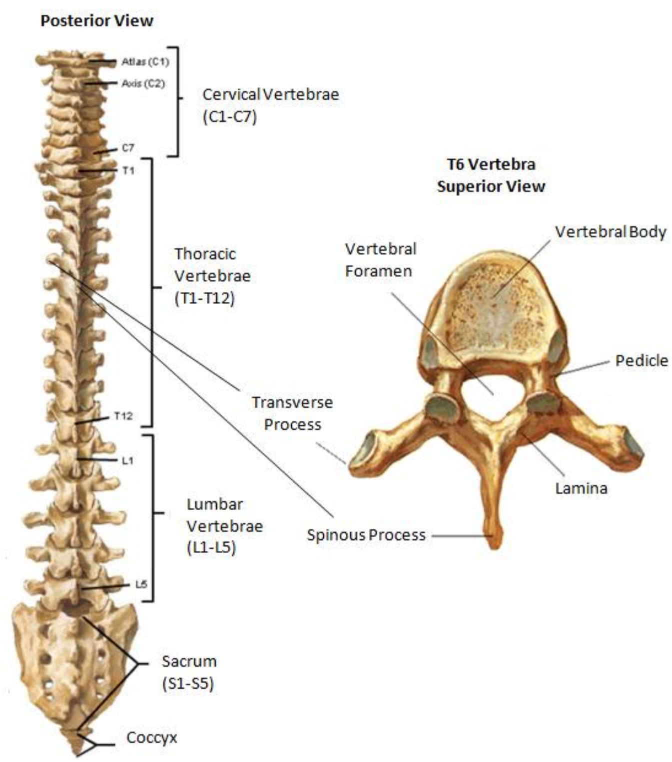

Figure 1: Spinal Anatomy (adapted from Netter, F., Atlas of Human Anatomy, 4th Ed., Saunders, 2006)

This GRF must be absorbed by the body structure. The muscles, especially in the lower limbs, actively contract to oppose the GRF. Ankle, knee and hip flexion occur to dissipate some of the impact energy. The remainder of the impact energy is transmitted to the other tissues and structures in the body, including the spinal vertebrae.

Determining the force applied to the spine during a landing is a complicated process that depends on body weight, external impact loads, and internal muscle forces. Biomechanical studies of feet-first landings after a fall have focused on muscle and joint reaction forces in the lower extremities. To our knowledge, there is no research evaluating the compressive forces in the spine or accelerations experienced near the spine during feet-first fall landings.

Spinal Injury Tolerances

Injury to the spine results when forces on the spine exceed the tissue tolerances of the spinal structures. The ability of the spine to carry load depends on the structural capacity of the vertebrae. Conditions that weaken bone architecture, such as ageing and osteoporosis, limit the ability to absorb impact, and reduce the injury tolerance (Myers & Wilson, 1997). For example, the average compressive strength of a T12 vertebra is approximately 3.5 kN; strong bones may have a fracture tolerance as high as 9 kN, while osteoporotic bones may fracture at loads of only 1 kN (Nachemson, 1966; Edmondston et al., 1994).

Acceleration tolerances of the human spine have been thoroughly studied by the military in the design of aircraft ejection seats (Ewing et al., 1972; Laurell & Nachemson, 1963; Stemper et al., 2011). It is generally accepted that the human spine can withstand accelerations of approximately 20 G at an onset rate of 300 G/s (Laurell & Nachemson, 1963).

Forensic Challenges of Determining Spinal Forces

There are a number of possible techniques for determining the forces in the spine. Direct measurement of spinal compression forces can be studied by inserting a pressure transducer between the vertebrae of living subjects (Ledet et al., 2005, Nachemson & Morris, 1964). Cadaver specimens have been used extensively with mechanical testing to estimate fracture thresholds. Most researchers employ indirect, model-based techniques involving derivation of joint forces from inverse dynamics of measured force plate and motion data of live subjects.

Although these methods may be suitable for biomechanical research studies, their use in forensic investigation of unique fall scenarios is limited. Live human testing of potentially injurious scenarios is not feasible, nor is direct in-vivo measurement of test subject spinal pressures at lower fall heights. Cadaver specimens are too expensive to be feasible for forensic investigation of civil claims and lack the realistic muscular/joint response of a conscious landing. Extrapolations can be attempted using inverse dynamic analysis at low fall heights, but this method requires highly sophisticated instrumentation and software.

Therefore, this preliminary study sought to investigate if a semi-rigid punching bag instrumented with accelerometers could be used as an effective and practical forensic alternative to estimate back acceleration during fall landings. We hypothesized that the semi-rigid properties of a heavy-duty punching bag may be a good representation of the rigidity of the human body during a feet-first fall landing. The applicability of this research is relevant to forensic investigation of landings from jumps, falls, sporting events, trampolines, or other recreational activities.

METHOD

In this pilot study, we compared acceleration data from the back and foot of a live test subject during landings from relatively low fall heights to the acceleration data from falls of the instrumented punching bag.

Test Apparatus

The punching bag used for this study was a professional grade, 68 kg (150 lb) Title Boxing (California, USA) heavy-duty punching bag. An adjustable weight add-on bag was filled with sand and loosely secured on top of the punching bag, centered along the axis of the bag. The total weight of the dropped apparatus was 88.5 kg. The weight of the test subject available for this study was 72.7 kg, which was approximately 18% lighter than our test apparatus. This difference was almost an exact match to the 17% difference in the landing contact area of the test subject's feet versus the landing contact area on the bottom of the punching bag. In this way, the landing pressure at the ground contact was constant for all tests and was eliminated as a variable.

Two three-axis impact accelerometers (X250-2 USB Impact Accelerometer, Gulf Coast Data Concepts, Waveland, USA) were placed on the sides of the bag, approximately 30 cm from the bottom of the bag. The accelerometers had a range of ±250 G, a resolution of 0.0381 G, and a sampling rate of 512 Hz, which is sufficient for a study of this nature. The output of the accelerometers included a time stamp. The accelerometers were located on the punching bag using duct tape and then securely held in place using palette wrap that was wound tightly around the punching bag. Preliminary testing confirmed that the accelerometers were held securely in place and did not permanently shift after a drop test.

The apparatus was dropped from a rope-pulley system (Figure 2) onto a grass-covered lawn from heights of 30 to 120 cm in 30 cm increments, for a total of four drops. The accelerometers were not damaged during the drop tests.

Live Subject Fall Landings

For the test subject fall landings, the same accelerometers were used as above. One accelerometer was attached to the dorsum of the subject's foot and the other one was attached to the subject's mid-lower back near T12. The accelerometers were securely attached using duct tape. The subject performed feet-first standing fall landings from heights of 18, 48 and 76 cm, with 10 to 11 tests conducted at each height. The heights were selected based on convenient yard objects that were available to stand upon. Higher fall heights were not selected for safety reasons. At each height, the subject performed two 'soft' landings (bending the hips, knees and ankles upon landing to lessen the impact), two 'stiff' landings (little to no hip, knee and ankle flexion), and the remainder were self-selected 'normal' landings.

Figure 2: Instrumented punching bag and pulley system apparatus.

To eliminate any error associated with orientation of the accelerometers, the resultant acceleration from each landing was computed. Ground contact was deemed to have occurred when the vertical acceleration increased above the freefall value. The primary impact loading period was determined as the time from ground contact until the maximum acceleration was reached. The 'jerk' (rate of acceleration) was then computed as the rate of change of acceleration during the primary impact loading period.

RESULTS

'Heavy Bag' Drops

The average results from the two impact accelerometers during the heavy bag drops are presented in Table 1. Some data from one of the accelerometers was not recoverable. The impact acceleration values recorded by the two accelerometers were consistent, although the jerk data varied. Both the maximum impact acceleration and jerk increased in a linear fashion with increasing drop heights (Figure 3).

Live Subject Jump Landing

Tables 2 and 3 show the maximum impact acceleration and the jerk for the accelerometers on the test subject. The impact accelerations at the foot were 2.5 to 3.5 times higher than those at the back. In general, stiff landings tended to increase the maximum impact acceleration, while soft landings tended to decrease the maximum impact acceleration. The trend associated with impact acceleration at the foot appears linear with increasing fall height. The trend associated with impact acceleration at the back appears linear with "leveling off" of the acceleration around 20 to 25 G, despite an increase in fall height.

The trend associated with jerk was less definitive. Generally, jerk measured at the foot increased linearly with fall height. Jerk measured at the back during a foot-first fall impact increased with fall height and "leveled off" as described above at around 800 to 1100 G/s. The influence of stiff and soft landings on jerk was highly variable, in particular for the foot data, as indicated by the standard deviation values in Table 2. The jerk data for the back was not as sensitive to the type of landing as the foot data.

Table 1: Max impact acceleration and loading jerk for the heavy bag drops.

	Sensor #1		Sensor #2		Average	
Drop Height cm	**Max Impact** G	**Jerk** G/s	**Max Impact** G	**Jerk** G/s	**Max Impact** G	**Jerk** G/s
30	7.97	375	7.42	321	7.70	348
60	20.06	1288	-	-	20.06	1288
90	33.25	2148	33.22	-	33.24	2148
120	46.77	4780	40.08	1882	43.42	3331

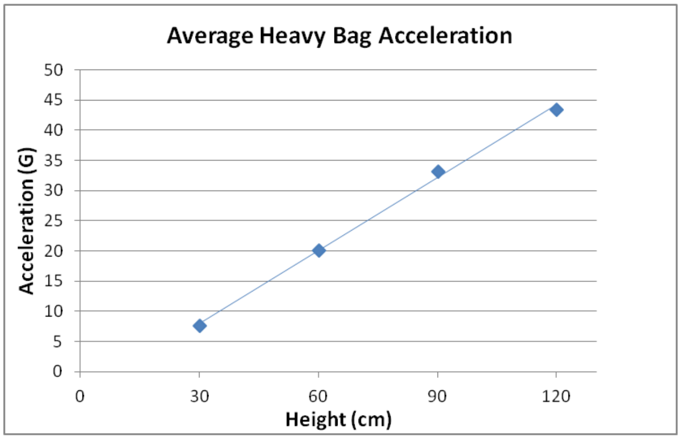

Figure 3: Heavy bag maximum impact acceleration vs. drop height.

Table 2: Max impact acceleration and loading jerk at the foot. Mean ± SD

Height cm	**Trials**	**Max Impact** G	**Jerk** G/s
All landing types			
18	11	28.47 ± 7.38	1727± 959
48	11	48.74 ± 8.60	2545 ± 1217
76	10	76.94 ± 9.26	3854 ± 1505
Self-selected (normal) landings			
18	7	23.82 ± 2.45	1167 ± 448
48	8	47.91± 9.06	2536 ± 1199
76	6	74.41 ± 11.17	3767 ± 1439
Stiff landings			
18	2	38.53 ± 8.65	3121 ± 327
48	2	55.42 ± 4.86	2869 ± 2054
76	2	77.93 ± 4.64	4605 ± 2553
Soft landings			
18	2	34.70 ± 1.58	2290 ± 1079
48	1	41.98	1973
76	2	83.50 ± 0.06	3363 ± 1359

Table 3: Max impact acceleration and loading jerk at the back. Mean ± SD

Height cm	**Trials**	**Max Impact** G	**Loading Jerk** G/s
		All landing types	
18	11	8.81 ± 1.85	421 ± 128
48	11	19.55 ± 4.64	942 ± 245
76	10	21.09 ± 3.94	809 ± 288
		Self-selected (normal) landings	
18	7	8.90 ± 1.22	379 ± 100
48	7	20.48 ± 4.54	1008 ± 237
76	6	20.47 ± 3.03	811 ± 175
		Stiff landings	
18	2	10.90 ± 2.25	545 ± 246
48	2	20.75 ± 6.11	944 ± 332
76	2	25.68 ± 5.55	1128 ± 374
		Soft landings	
18	2	6.42 ± 0.43	442 ± 15
48	2	15.09 ± 2.34	708 ± 108
76	2	18.35 ± 1.96	487 ± 212

Correlation of Acceleration Data

The impact acceleration and jerk measured during the heavy bag drops and live subject landings were plotted together in Figure 4 and 5.

Figure 4 shows that the impact accelerations measured on the punching bag during the drop tests closely matched the accelerations at the exterior of the back. The accelerations recorded at the foot were 2.5 to 3.5 times higher than the back accelerations.

Figure 5 shows that at low drop heights, the jerk data from the punching bag fell within the range measured at the exterior of the back. At higher drop heights, the jerk data from the punching bag was between that of the extrapolated back data and foot data.

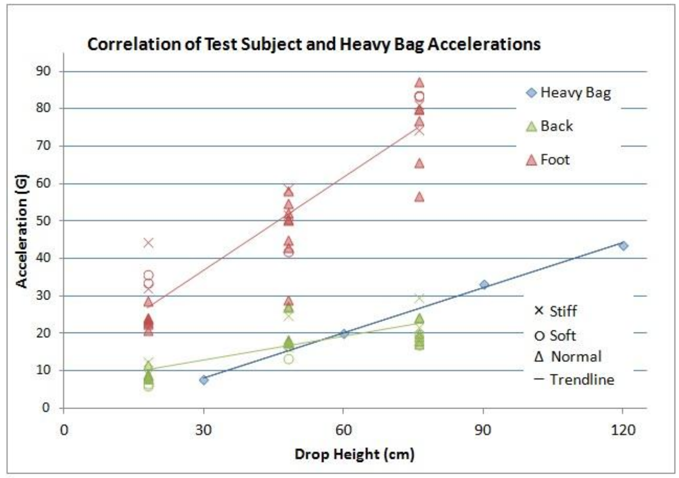

Figure 4: Correlation of test subject and heavy bag acceleration data**.**

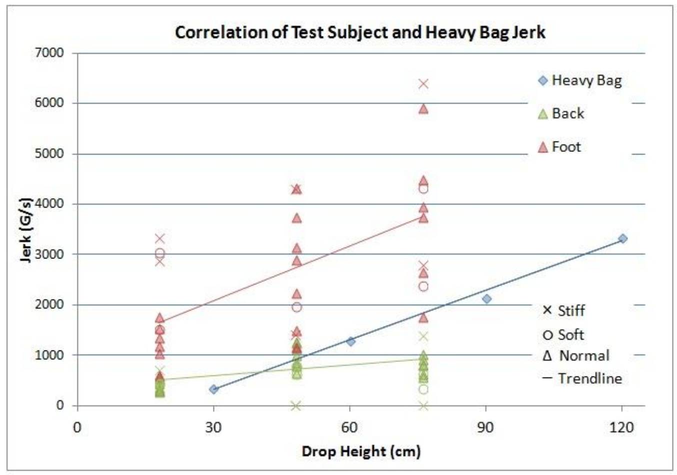

Figure 5: Correlation of test subject and heavy bag jerk data.

DISCUSSION

This study was a preliminary investigation to explore the use of a heavy bag and accelerometers in estimating back acceleration during feet-first fall landings in forensic investigations. We found that the measured heavy bag impact accelerations directly approximated those recorded on the external low back during actual feet-first fall landings.

There were minor differences in the values recorded by each of the heavy bag accelerometers, but no obvious discrepancy that could indicate an error with the equipment or the analysis process. These differences were likely attributed to the bag falling at a slight angle. Since the accelerometers were placed on opposite sides of the bag, an angled fall could cause one side to impact more than the other. For this reason, we chose to use the resultant (direction independent) acceleration in our computations and to average values the two accelerometers in our analysis.

Another potential cause of the measurement discrepancies could be some minor relative motion between the bag and the accelerometer during the drops. Based on our preliminary tests, we believe any error from this source would be minor and would not impact the results of the study.

The data recorded at the test subject's foot during real landings was surprisingly higher than expected. Because the resultant acceleration was used instead of only vertical acceleration, this may also have included components of the lower leg acceleration as the ankle flexed to accommodate landing. This may also explain why there was so much variability observed at the foot between stiff and soft landings, whereas the data at the back was not highly sensitive to the type of landing.

Theoretically, there should be less joint movement in a stiff feet-first landing and more joint movement during a soft feet-first landing. This should result in higher peak accelerations at the foot during stiff landings, as the lower joints are forced to become more rigid. The observed data from this preliminary study matched that expected pattern at all fall heights except one. Even at the back, the observed data matched this expected pattern, although less notably, with stiff landings resulting in higher peak accelerations, and soft landings resulting in lower peak accelerations.

The purpose of having the test subject perform both stiff and soft landings in this study was to develop a preliminary range of acceleration data for feet-first fall incidents. Observation of the test subject confirmed that the 'stiff' and 'soft' landings were performed correctly. Nevertheless, high variability was noted in the data, indicating the substantial variability from one landing to another, even for the same test subject. Additional testing would provide a larger sample size. It is also plausible that muscular control during feet-first fall landings is controlled to some extent by instinct and is not a fully conscious exercise.

Previous experiments and observations in the scientific literature presume the average human spine can tolerate up to 20 G at an onset of 300 G/s. Our data, measured at the exterior of the back, showed that during typical feet-first landings, the back as a whole experienced accelerations of up to 20 to 25 G at an onset of 800 to 1100 G/s without incident. Further research would be required to correlate the acceleration sustained by the back as a whole with the accelerations within the individual vertebrae. This study measured the external acceleration of the back at the location of the T12 vertebra, not the actual inter-disc acceleration at the T12 vertebra. The intra-vertebral accelerations are expected to be lower than our external measurements due to the action of the spinal ligaments and muscles which cushion and absorb the impact, thereby reducing the acceleration on the vertebrae.

Interestingly, we observed a leveling off of the measured external back accelerations around 20 to 25 G and 800 to 1100 G/s. This may imply that the actual acceleration curve for the vertebrae may be more asymptotic than linear, suggesting some ability of the body to limit the impact on the spine below its injury threshold. The question is where does this ability break down, i.e. at what height does the impact become too great and result in fracture?

With that knowledge, the use of a heavy bag to estimate accelerations at the external back could be used as a practical and representative measure of spinal injury risk in feet-first fall landings, or other types of fall landings if the research were extended to consider other fall modes. This preliminary study has shown that the semi-rigid composite structure of a professional-grade heavy-duty punching bag provides a reasonable approximation of the semi-rigid structure of the human body for impact purposes, and can be used for further comparative impact studies.

The jerk measured at the punching bag was too high to be representative of the jerk experienced at the external back, and too low to be representative of the jerk experienced at the feet. Since jerk is dependent on contact time with the ground, a very slight modification in composition of the heavy bag (or in the impacting contact surface of the bag) would likely lead to improved results. Care should be given to avoid adding too much energy-absorbing material on the bottom surface of the bag, as that would have the counter-productive effect of lowering the peak accelerations below the test subject values.

Unfortunately, there is little research on spine biomechanics during fall landings, and no publications noting spine accelerations during feet-first fall landings. The continuation of this line of research would be of great use to the forensic engineer or biomechanist attempting to establish injury risk in fall landing scenarios. This preliminary research was an initial attempt to address that issue by simulating back accelerations with simplified equipment drop tests. This initial study was limited to a small number of landings, a single type of heavy bag, and a single test subject. Future testing should include higher fall heights to validate the linearity of the data trends, the use of additional test subjects, the use of additional heavy bag types, and additional landing modes (e.g. pelvic landings and hip first landings from low fall heights).

REFERENCES

Caulfield, B. & Garrett, M. (2004). Changes in ground reaction force during jump landing in subjects with functional instability of the ankle joint. *Clinical Biomechanics (Bristol, Avon)*, 19(6), 617-21.

Decker, M. J., Torry, M. R., Noonan, T. J., et al. (2002). Landing adaptations after ACL reconstruction. *Medicine and Science in Sports and Exercise*, 34(9), 1408-13.

Devita, P. & Skelly, W. A. (1992). Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Medicine and Science in Sports and Exercise*, 24(1), 108-15.

Edmondston, S.J., Singer, K.P., Day, R.E., et al. (1994). In-vitro relationships between vertebral body density, size, and compressive strength in the elderly thoracolumbar spine. Clinical Biomechanics, 9(3), 180-6.

Eloranta, V. (1996). Effect of postural and load variation on the coordination of the leg muscles in concentric jumping movement. *Electromyography and Clinical Neurophysiology*, 36(1), 59-64.

Ewing, C.L., King, A.I., & Prasad, P. (1972). Structural considerations of the human vertebral column under +Gz impact acceleration. *Journal of Aircraft* , 9(1), 84-90.

James, C. R., Bates, B. T., & Dufek, J. S. (2003). Classification and comparison of biomechanical response strategies for accommodating landing impact. *Journal of Applied Biomechanics*,19(2), 106-118.

Kovacs, I., Tihanyi, J., Devita, et al. (1999). Foot placement modifies kinematics and kinetics during drop jumping. *Medicine and Science in Sports and Exercise*, 31(5), 708-16.

Laurell, L. & Nachemson, A.(1963). Some factors influencing spinal injuries in seat ejected pilots. *Aerospace Medicine*, 34(8), 726-9.

Myers, E.R. & Wilson, S.E. (1997). Biomechanics of osteoporosis and vertebral fracture. *Spine (Phila Pa 1976)*, 22(24S), 25S-31S.

Nachemson, A. (1966). The load on lumbar disks in different positions of the body. *Clinical Orthopaedics and Related Research*, 45, 107-22.

Netter, F. (2006). *Atlas of Human Anatomy*, 4th Ed., Saunders, U.S.A.

Stemper, B.D., Storvik, S.G., Yoganandan, N., et al. (2011). A new PMHS model for lumbar spine injuries during vertical acceleration. *Journal of Biomechanical Engineering*, 133(8), 1-9.

Zhang, S. N., Bates, B. T., & Dufek, J. S. (2000). Contributions of lower extremity joints to energy dissipation during landings. *Medicine and Science in Sports and Exercise*, 32(4), 812-9.