

Assessment of Compressive Thoracolumbar Injury Potential and Influence of Seat Cushions on Vertical Impact Loading of a Seated Occupant

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ABSTRACT

Thoracolumbar vertebral fractures are most commonly due to compressive loading modes and associated with falls from height. Two injury metrics are generally referenced for assessing the potential for compressive thoracolumbar injury; the Dynamic Response Index (DRI) and the compressive load measured between the pelvis and lumbar spine using the Code of Federal Regulations (CFR) title 49 part 572 subpart B anthropomorphic test device (ATD).

This study utilizes an ATD to investigate the injury mitigation potential of a variety of seat cushions during vertical impact in an unrestrained seated posture. ATD responses and DRI are reported for 65 vertical impacts with and without cushions from heights between 4 and 80 inches. The cushions investigated reduced ATD peak pelvic acceleration $63 \pm 11\%$ and compressive lumbar load $42 \pm 9\%$ on average. Results indicate a drop height of approximately 19 inches onto a rigid surface will generate ATD lumbar loads of 1500 lbs, whereas drop heights of 42 ± 10 inches were required to produce equivalent lumbar loads with a cushion present. Conversely, the presence of a cushion had little or no influence on the resulting DRI. A 40 inch drop onto a rigid surface resulted in a DRI of 20.3 and drops of 40 inches onto a cushion resulted in DRI of 20.5 ± 1.3 . ATD lumbar load is a superior tool for qualitative comparative purposes. Neither metric has been correlated to human tolerance during unrestrained vertical impacts. Further research is required to better define their relationships to injury potential.

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INTRODUCTION

Traumatic spinal cord injuries, though infrequent, are a significant cost to society. Since 2010, falls from height and motor vehicle crashes account for the vast majority of spinal cord injury. Additionally, at least 76% of documented spinal cord injuries occur with documented vertebral fracture or dislocation [1]. Thoracolumbar vertebral fractures are most commonly due to compressive loading modes and associated with falls from height followed by transportation related injury, sports related injury, and blunt impact [2,3,4]. A sample of the diverse environments where thoracolumbar fractures have been reported includes motor vehicle crashes, snowboarding, wave/wake jumping activities using personal watercraft craft (PWC), high speed water entry, and blast exposure due to improvised explosive device (IED) detonation [5,6,7,8,9,10,11].

For decades, researchers have investigated the human response under caudocephalad (headward) accelerations and the corresponding potential for thoracolumbar injury. The majority of research to date has focused on aviation injury prevention, specifically ejector seat and helicopter/small plane crash environments. A number of factors that influence injury potential and human tolerance levels have been identified by prior research, including: acceleration pulse amplitude, duration, and rise-time; occupant posture; restraint system use and

design; and seat (cushion) characteristics [12,13,14]. The vertical impact scenario of the unrestrained occupant is of interest in a number of recreational motorsports environments including: all-terrain vehicles, snowmobiles, PWC, and recreational boating. Unlike the aviation ejector seat and crash environments, restraint systems are not feasible as injury prevention and mitigation tools. Therefore, potential thoracolumbar injury mitigation tools are limited to cushions and, in some cases, suspension systems.

To date, there is very little information available in the literature which quantifies the potential influence of seat cushions on thoracolumbar injury risk. Prior investigations in the ejector seat and aviation crash environments have reported the presence of a seat cushion may actually increase the probability of injury [15]. With a rigid seat pan (no cushion) a well-restrained occupant pelvis will accelerate with the underlying seat. However, with a cushion present, the occupant acceleration can be delayed, allowing a relative velocity to build between the seat and the pelvis until the cushion "bottoms out," effectively generating larger acceleration and onset rates for the occupant. Hearon and Brinkley found that "rate-dependent" open-cell polyurethane foams could be used in place of traditional seat foams and result in comparable performance to a rigid seat pan [14]. In an impact environment, intuition suggests that a cushion would have a beneficial effect for the occupant acting to reduce the potential for

thoracolumbar compressive injury. In a vertical impact, a cushion could be expected to reduce peak acceleration levels and lengthen the impulse duration relative to impact on a rigid surface. It remains unclear if, and to what extent, the presence of a cushion translates into a quantifiable reduction in the potential for thoracolumbar injury during a vertical seated impact.

Dynamic Response Index (DRI)

Human thoracolumbar tolerance in vertical loading environments has primarily been characterized by seat kinematics in whole-body experimentation and by failure kinetics during vertebral segment level experimentation. Much of the early work investigating thoracolumbar injury tolerance to caudocephalad (headward) acceleration was related to aviation and ejector seats. One of the first acceleration tolerance curves was developed by Eiband [12] for simple trapezoidal pulses. Eiband reports a threshold of 20g for ejector seat pulses greater than 5 msec. in duration based on static compressive testing of cadaver vertebrae by Ruff [16]. Operational data found that human subjects were generally able to withstand 40 msec. duration pulses at 16 g with lap / shoulder belts and face curtains which assisted in maintaining an erect posture during ejections.

In the 1960s, basic amplitude / duration tolerance curves were expanded with the development of the DRI model for the human response to vertical acceleration. DRI is a simple base-excitation spring-mass-damper model intended to represent the response of the spine and upper body to vertical acceleration. The foundation of the DRI model is discussed in more detailed in Payne and Stech [17]. The governing equation for the DRI model is defined by Eq. 1, where δ is the compression of the system (i.e. spine analog), ζ is the damping coefficient, and ω_n is the natural frequency. These model parameters were derived from human vibration and mechanical impedance data reported by Coermann [18]. The model parameters (damping and natural frequency) are reported to vary with age based on work performed by Perey [19] as well as with peak DRI value. The most commonly reported DRI parameters are $\omega_n = 52.9$ rad/s (8.42 Hz) and $\zeta = 0.224$, which were originally chosen for a 27.9 year old 160 pound male at a maximum DRI of 20.

$$\ddot{z}(t) = \ddot{\delta} + 2\zeta\omega_n\dot{\delta} + \omega_n^2\delta \quad (1)$$

Eq. 1 is solved for a given acceleration pulse, $\ddot{z}(t)$, to determine the maximum compression of the DRI (spine) model, δ_{max} . A DRI value for a specific acceleration pulse is typically reported which is related to δ_{max} via Eq. 2.

$$DRI = \frac{\delta_{max}\omega_n^2}{g} \quad (2)$$

The DRI model was initially used to evaluate the pulse from the seat pan during aviation ejection events. While the DRI model parameters were originally developed based on human vibration tests, subsequent research has reported both seat pan and ATD pelvis acceleration pulses as suitable inputs for the DRI model [20]. DRI has been considered an improvement over the early duration / amplitude relationships for human tolerance, as DRI can be applied to acceleration pulses of virtually any shape, amplitude, or duration [21]. Brinkley and Shaffer report linearly correlated DRI with post-mortem human subject (PMHS) experimentation and operational data. A DRI value of 21.3 was found to correspond to a 50% probability of spinal injury based on PMHS vertebral strength testing. A slightly higher DRI value, 22.5, was found to produce the same probability of injury based on operational data for young male aviators [22].

While considered successful in reducing the incidence of thoracolumbar injury due to ejector seats, DRI is not without limitations. There are a number of parameters that the DRI model does not take into account that are known to influence human injury tolerance. A study of aircraft seats by Laananen [23] states that injury mechanisms and spinal deformation “cannot be reliably predicted from seat pan acceleration, as internal body response can vary significantly for various occupant types with similar inputs from the seat [23].” In the ejector seat scenario, DRI cannot account for the influences of a seat cushion, the restraint systems (belts and face curtains), posture of the individual, alignment of the pulse with the individual, and off-axis loads.

Similar limitations are expected with the application of the DRI model to the vertical impact of an unrestrained seated subject. DRI-based injury risk and tolerance levels, developed based on ejector seat data, should not be expected to apply directly and equivalently to an unrestrained seated impact. Careful consideration should be given when applying DRI to loading environments / conditions different than it was originally developed for and the relationship to injury potential should not be accepted as universally applicable.

Hybrid II ATD Lumbar Compressive Load

Another thoracolumbar injury metric was developed in the 1980s focused on occupant protection and seat design in the helicopter and light-aircraft crash environments. A lumbar compressive load criterion was developed for the part 572 subpart B (Hybrid II) ATD [24] and adapted by the Federal Aviation Administration (FAA) in 1988 [25]. The load criterion was developed for assessing the crashworthiness of aviation crew seats and restraint systems under a prescribed representative crash pulse. To satisfy the performance criterion, the peak compressive force measured by the lumbar load cell must not exceed 1500 pounds.

Chandler [21] outlines the development of the 1500 pound lumbar load threshold chosen by the General Aviation Safety Panel (GASP), a working group formed in 1983. Generally, the threshold was based

on an investigation of energy absorbing seat structures with a rigid bucket (seat pan) and measurements made at the Civil Aeromedical Institute (CAMI). The CAMI tests enabled computation of DRI from seat pan acceleration and correlation with the measured ATD lumbar loads. The selected lumbar load of 1500 pounds was chosen by GASP as it corresponds to a DRI of about 19, based on the CAMI test results. Military operational data suggests that a DRI of 19 translates to an anticipated spinal injury rate of 9%. Since the ATD lumbar force criterion was derived directly from its relationship to DRI, a lumbar load threshold of 1500 lbs. should be expected to exhibit many of the same limitations as the DRI model.

The GASP recommendations used the term “pelvic force” to describe the load limitation to avoid comparison to the load capability of the human spine. Chandler describes that “[s]uch a comparison would be meaningless because the biofidelity of the dummy is not equivalent to the human in the spinal direction.” The authors are unaware of any validation or direct correlation of the ATD lumbar force criterion with human response or injury tolerance.

Both DRI and ATD lumbar load are uniaxial metrics and only address injury potential in the vertical direction. PMHS experimentation demonstrates that sagittal plane moments coupled with compressive loading have significant influence on the type of vertebral injury and load at which it occurs [13]. While it is beyond the scope of this manuscript, it is worth noting that flexion (i.e. anteriorly oriented compressive loads) may play a significant role in human tolerance during an unrestrained vertical impact.

Available research on the response and injury tolerance of a human subject during a vertical unrestrained seated impact is limited. The experimental investigations presented here report the response of the part 572 subpart B ATD to a vertical impact in a seated posture with and without a cushion. Several boat seat cushions and two varieties of aviation foam were selected to evaluate their influence on ATD response and injury mitigation potential. At the present time, no injury assessment reference values exist that are directly applicable to the vertical impact of an unrestrained occupant in a seated posture. While neither of the existing and commonly used thoracolumbar injury tolerance metrics is correlated to injury potential in this loading scenario, they are utilized to assess the influence of a cushion relative to an impact on a rigid structure. Differences in the two injury metrics are observed and their limitations and efficacy for assessment of the injury mitigation potential of seat cushions are discussed.

EXPERIMENTAL METHODS

The reported results were collected during two phases of experiments conducted to investigate the influence of a cushion on compressive spinal injuries during vertical impact.

The “Hybrid II” ATD (49 CFR part 572 subpart B) was utilized in both experiments and was positioned in a seated posture with torso upright, upper legs horizontal, lower legs vertical, and hands resting

in the ATD lap. The ATD was dropped in this orientation onto a generic mock-up of a boat seat frame. A climbing harness was attached to the ATD pelvis and the ATD was suspended above the seat frame using a four point tether system, two attached to the front and back of the climbing harness and two secured to the ATD knee joints. The tethers were secured to an electromagnet which functioned as a release mechanism to drop the ATD.

The ATD included comparable instrumentation during both phases of testing. A three-axis (anterior-posterior shear force, axial force, sagittal plane moment) lumbar load cell (Phase I: Denton, Model 1891 and Phase II: Humanetics, Model 1842) was integrated between the ATD pelvis and the lumbar spine. Two $\pm 2000g$ accelerometers (Model 7264C) were mounted in the standard ATD pelvis and chest locations to measure the vertical response. Sensor data were collected at 10 kHz with a data acquisition system (PicoDAS16) utilizing an analog anti-aliasing filter. The data channels were digitally post-processed with phase-less low-pass filters according to specifications of SAE J211. The lumbar forces and moments were filtered at channel frequency class 600 (CFC-600) and the accelerometer data were filtered at CFC-1000. The ATD response was visually recorded using a high speed camera (300 frame/sec) looking laterally from the left side of the ATD. The high speed video and data acquisition were triggered synchronously upon release of the ATD via the switch for the drop mechanism.

The first phase of experiments was conducted to investigate the influence of a single boat cushion taken from a 1988 center-console style, 26 foot off-shore fishing vessel. The cushion came from a dual-purpose area in the bow of the vessel intended for use as a casting platform for fishing and as a passenger seating area while underway. The foam cushion was approximately 2.5 inches thick, and given the age of the seat cushion it may have deteriorated from its original condition. During Phase I, the horizontal surface of the seat-frame was rigidly constructed and positioned 17.5 inches above the floor representing the bow area where the cushion was located (see [Figure 1](#)). The seat frame height was such that the ATD buttocks contacted the seat frame very close to the same time that the ATD feet contacted the floor during the drop tests. During Phase I, a total of 15 tests were conducted where the ATD was dropped from heights ranging from 4 to 80 inches. Drop heights reported for each individual test were measured from the lowest point of the ATD buttocks to the top of the cushion or seat frame if no cushion was present. In addition to the internal ATD instrumentation, a three-axis “pancake”-style load cell (Denton, Model 2177A) was integrated into the seat frame at the ATD impact site to measure the forces experienced between the ATD and seat-frame in three dimensions.

The cushion and foam materials used in Phase II of experimentation are outlined in [Table 1](#). Six cushions from the bow seating areas of recreational boats and two types of aviation foam were selected as test specimens. The seat frame was constructed with lumber and the seat height was 22.5 inches above the floor, 5 inches greater than the frame used during Phase I (see [Figure 1](#)). The seat frame height was increased to prevent the ATD feet from contacting the floor,

effectively directing the entirety of the ATD energy into the cushion and / or seat frame. During Phase II, a total of 50 tests were conducted with drop heights of 5, 10, 20, 30, and 40 inches.

Table 1. Index and brief descriptions of boat cushion and aviation foam specimens evaluated.

Test Specimen	Cushion/Foam Thickness	Description
Cushion 1	3.5"	2007 19" outboard deck-boat
Cushion 2	2.0"	1998 21' inboard ski-boat
Cushion 3	3.5-4.0"	1995 19' I/O bowrider
Cushion 4	3.0"	2003 20' I/O bowrider
Cushion 5	2.5"	2003 23' inboard ski-boat
Cushion 6	3.0"	2011 inboard wakeboard boat
Foam 1	3.0"	DAX-26 (Skandia)
Foam 2	3.0"	DAX-55 (Skandia)



Figure 1. Photographs of ATDs and seat frames in the pre-drop orientation (from 20 inches) from Phase I (left) and Phase II (right).

A high frequency ringing, at approximately 2 kHz, was observed in either one or both of the ATD pelvis and chest accelerometer signals during some of the phase II tests. This high-frequency ringing is believed to be an experimental artifact and not associated with the physical impact response of the ATD. SAE J211 CFC-1000 did not effectively filter the ringing so the accelerometer data was re-processed with a low-pass filter (4-pole phase-less Butterworth) with a -3 dB cutoff frequency of 495 Hz. Data from tests where the ringing phenomenon was not observed were used to compare the CFC-1000 and 495 Hz filters to confirm that the lower cut-off frequency had a negligible effect on the reported peak accelerations. Figure 2 shows the pelvis and chest acceleration traces filtered at CFC-1000 and with the 495 Hz low-pass filter for the highest drop (80 inches) onto a rigid frame. For this very short duration impulse,

the 495 Hz filter reduced the peak pelvis and chest accelerations by approximately 0.4% and 2.2% respectively. Throughout this manuscript, the results follow the sign convention of prior research where compressive lumbar load and head-ward acceleration are reported as positive, with the exception of Figure 3 where the data is presented in the standardized ATD coordinates outlined in SAE J211.

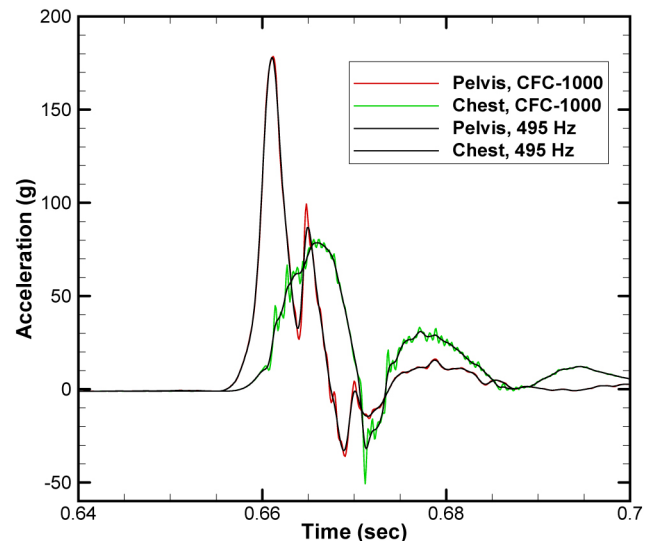


Figure 2. Sample ATD pelvis and chest accelerations with CFC-1000 and the 495 Hz low-pass filter for an 80 inch drop onto a rigid frame.

RESULTS

Time-traces of the sensor data are provided in Figure 3 from a sample drop test (Phase II 40 inch drop with Cushion 1) in SAE J211 coordinates. It should be noted that sensors were zeroed while the ATD was suspended in a stationary pre-drop condition, such that during free-fall the lumbar load cell was unloaded to an output of approximately -55 lbs. and the accelerometer outputs were -1 g. The peak lumbar shear force (X) and sagittal plane moment (Y) were found to scale reasonably well with lumbar axial force (Z) across test conditions and given that they are not explicitly included in the injury criteria evaluated, they will not be discussed in any detail though the remainder of this manuscript. The impulse durations varied between sensor location and with the test configuration. The longest duration pulses were on the order of 100 msec. for lower drops (5 inches) with cushions present and the shortest duration pulses were on the order of 10-20 msec. for higher drops (>40 inches) on the rigid frame.

For demonstrative purposes, a series of still frames from the high-speed video are provided in Figure 4 corresponding to the data shown in Figure 3. Qualitatively, the gross motions of the ATD recorded by the high-speed video were similar with and without a cushion present. The sagittal plane moment experienced at the ATD lumbar load cell was initially negative (extension) during the impulse followed by a positive moment as the upper torso of the ATD flexed forward. This general trend was observed throughout all test conditions.