Effect of Loading Rate on Injury Patterns During High Rate Vertical Acceleration

Brian D. Stemper, PhD[1], Jamie L. Baisden, MD[1], Narayan Yoganandan, PhD[1], Frank A. Pintar, PhD[1], John DeRosia, PhD[1], Phillip Whitley, PhD[2], Glenn R. Paskoff, MS[3], Barry S. Shender, PhD[3]

Abstract Military occupants can be exposed to more severe environments than civilian. High-rate vertical acceleration occurs under a variety of military activities and spinal injury distribution may be dependent upon acceleration characteristics. This preliminary investigation determined spinal fracture patterns in post-mortem human subject (PMHS) lumbar spines for two simulated environments: catapult phase of aviator ejection and helicopter crash. Vertical accelerations simulating ejections had peak magnitudes of 20-22 G with rates of onset less than 525 G/s. Accelerations simulating helicopter crashes had peak magnitudes of 44-65 G with rates of onset exceeding 1000 G/s in one specimen and exceeding 2000 G/s in two specimens. In this study, two lumbar spines were subjected to simulated ejections and three spines were subjected to simulated helicopter crashes. Results demonstrated fractures primarily affecting vertebral bodies and a majority occurring under axial compression mechanisms. Although fracture types were not different between environments (burst and anterior compression fractures occurred in each), injury location migrated caudally for higher severity accelerations. Whereas compression fractures affected the L1 spinal level for ejection accelerations, fractures were distributed between L1 and L4 levels for helicopter crash accelerations. More severe helicopter crash accelerations (>60 g, >2000 g/s) demonstrated injuries affecting L2-L4 levels. Results from this experimental study are validated by clinical reports of military personnel, wherein caudal injury locations were evident for higher severity accelerations such as helicopter crashes.

Keywords biomechanics, Gz, lumbar spine, military, injury tolerance.

I. INTRODUCTION

Military personnel can be exposed to a variety of unique and extreme loading environments. Military vehicles are different from civilian vehicles and crash or failure scenarios are more severe. Commonly encountered scenarios include aircraft ejection, helicopter crashes, and underbody blast exposure in ground-based vehicles. Each of those loading scenarios has a component of high rate vertical acceleration. For aircraft ejection, vertical accelerations of 18 g's with 250 g/s were established as the upper limit for aviator egress in the Advanced Concept Ejection Seat (ACES). Helicopter crashes can be even more severe, although crash dynamics are considerably more complex with loading dependent upon altitude, airspeed, impact topography, and vehicle orientation at the time of impact. One study that analyzed reports for 298 crashes occurring in four military helicopter models over a six year period indicated that mean vehicle vertical change in velocity was between 4.3 and 16.0 m/s for all crashes and between 3.5 and 8.8 m/s for survivable crashes [1]. Seat pan accelerations measured during a full-scale crash test of a military helicopter with similar vertical velocity (11.6 m/s) were 40 and 33 G's occurring over approximately 15 ms for the pilot and co-pilot [2], respectively, resulting in approximately 2,500 g/s rate of onset.

High-rate vertical loading environments can induce a variety of injuries in military vehicle occupants due to the initial acceleration event or impact of body components. The spine bears a majority of the load during the initial acceleration event and, therefore, spinal fractures are common in all three loading environments described above [3-6]. However, different loading rates in each scenario lead to different injury types and

BD Stemper is an Associate Professor in the Department of Neurosurgery, Medical College of Wisconsin (bstemper@mcw.edu). N Yoganandan and FA Pintar are Professors and JL Baisden is an Associate Professor in the Department of Neurosurgery, Medical College of Wisconsin. J DeRosia is a Research Associate at the Medical College of Wisconsin. P Whitley is President of Criterion Analysis, Inc. BS Shender is a Senior Scientist in the Human Systems Department at the Naval Air Systems Command (NAVAIR). GL Paskoff is a Biomedical Engineer in the Crew Performance Technology and Assessment Department at NAVAIR.

locations, although other factors such as occupant posture may also play a role. A recent review article identified differing injury types and locations for spinal injuries occurring in military environments [7]. That study identified caudally biased injury locations in higher loading rate environments such as helicopter crashes. Specifically, wherein ejection-related injuries were distributed across the thoracic and lumbar spine, with concentrations at the mid-thoracic and thoraco-lumbar regions, helicopter crashes were biased toward the lumbar spine. Sixty-eight percent of vertebral fractures sustained in helicopter crashes occurred at or caudal to T11. In contrast, only 44% of ejection-related vertebral fractures occurred at those levels. This differentiation in injury location between the three scenarios may be attributed to differences in axial loading rates. An understanding of spinal tolerance at these loading rates is required to produce more effective safety enhancements for military vehicles.

Biomechanical tolerance of human tissues can be quantified using experimental fixtures ranging from the whole body post-mortem human subjects (PMHS) mounted on servo-hydraulic sleds to isolated tissues tested with electro-hydraulic test devices [8, 9]. Dynamic tolerance of isolated thoraco-lumbar spines has been investigated using weight-drop [10, 11] and electrohydraulic piston techniques [12, 13]. A majority of those studies focused on quantifying fracture tolerance in short segment models (i.e., 2-3 vertebra segments) at a single loading rate. Although some information regarding rate dependence of thoraco-lumbar vertebrae can be derived from a conglomerate analysis across multiple studies, differences in testing protocols and specimen types add variability to the dataset that may mask the true influence of the rate factor. Therefore, an experimental investigation designed to quantify the rate dependence of thoraco-lumbar vertebrae is warranted.

Our laboratory has recently developed a new experimental model to simulate high-rate axial loading to the spine in military environments [14]. The model produced repeatable kinematics and clinically relevant vertebral fractures, including burst and wedge fractures. The model is applicable for the study of vertebral fracture rate dependence in military environments due to the ability to control the magnitude and shape of the acceleration pulse applied to the base of the lumbar spine. Characteristics of the vertical acceleration versus time pulse, including peak acceleration, rate of acceleration onset, and change in velocity, can be independently controlled. The purpose of this investigation was to employ our novel experimental model to quantify rate dependence of lumbar vertebral fracture tolerance during high-rate axial loading.

II. METHODS

Five lumbar spine specimens (T12-L5) were obtained from PMHS with a mean age of 42.2 ± 13.7 years. All spinal soft tissues remained intact. Three male and two female specimens were used. Each specimen was fixed at cranial and caudal ends using polymethylmethacrylate (PMMA). Fixation was accomplished such that the L2-L3 intervertebral disc was horizontal. Specimens were attached to the lower platform of our drop tower apparatus [14]. Upper and lower horizontal platforms were attached to a vertical track drop tower (Fig. 1) using decoupled carriages attached to the track using guide wheels (Bishop-Wisecarver Corp., Pittsburgh, CA). A six-axis load cell was attached between the base of the specimen and the platform to measure tri-axial forces and moments. A uniaxial accelerometer was attached to the lower platform to measure vertical deceleration of the specimen base. All biomechanical data were antialias filtered prior to digitization, recorded at 10,000 Hz, and digitally lowpass filtered according to Society of Automotive Engineers (SAE) specifications.

The test protocol consisted of adding 32 kg to the upper platform to simulate mass of the torso, upper extremities and head-neck complex of a 50th percentile male occupant. Specimens were pre-flexed using a 5-Nm moment and the contacting cylinder from the upper platform placed in contact with the cranial PMMA fixation to maintain initial specimen positioning. The cylinder was aligned 3.5 cm anterior to the L2-L3 posterior longitudinal ligament in all specimens. Both platforms were then raised to the specific drop height. Once released, gravity accelerated both platforms down the drop tower until the lower platform contacted pulse-shaping foam at the base of the tower. The foam decelerated the lower platform, allowing the upper platform to place an inertial load on the cranial PMMA fixation. Foam characteristics for each deceleration pulse were determined prior to PMHS testing through a series of pulse shaping experiments using an anthropomorphic test surrogate. Acceleration of the lower platform was designed to be within the range of either the catapult phase of aviator ejection from military aircraft or military helicopter crash.

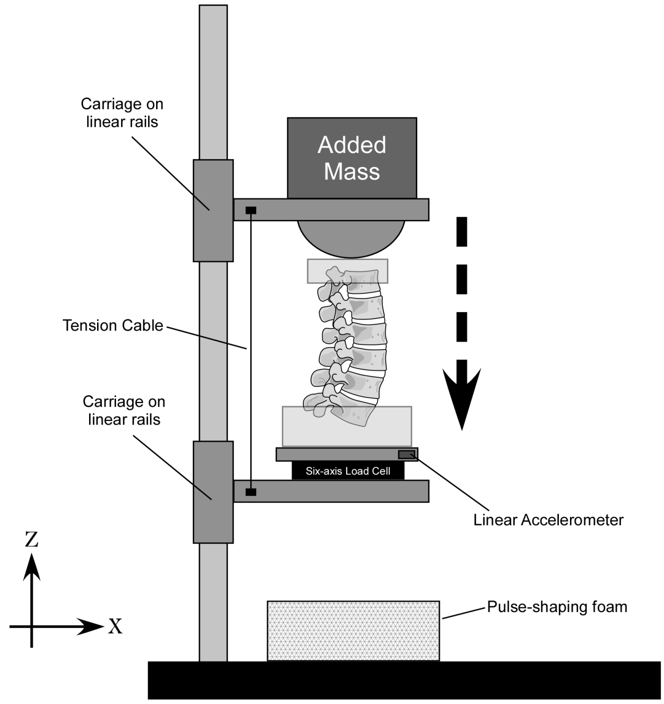

Fig. 1. Experimental testing apparatus to simulate high-rate vertical acceleration of the lumbar spine.

Sagittal x-rays and axial computed tomographic (CT) scans were obtained of each specimen following testing. These images were used to classify injury type and location by a member of our Neurosurgical staff. Each test was classified according to acceleration characteristics of the lower platform. Peak acceleration was determined as the absolute value of the maximum deceleration. Rate of acceleration onset (i.e., jerk) was determined as the peak acceleration divided by the time duration between initiation of deceleration and peak acceleration. Change in velocity was computed as the area under the acceleration-versus-time curve from the initiation of deceleration until the first zero crossing following peak acceleration. Peak force and sagittal bending moment were correlated to injuries.

III. RESULTS

Five PMHS lumbar spine specimens were exposed to high-rate axial loading to simulate either aviator ejection or helicopter crash environments. Characteristics of the helicopter crash acceleration pulse generally demonstrated a more severe environment with higher peak acceleration and rate of acceleration onset (Table I). Due to variability in helicopter crash accelerations, two acceleration pulses were incorporated in this study; a less severe pulse with lower peak acceleration (44.3 g) and rate of onset (1065 g/s) and a more severe pulse (~ 65 g, >2500 g/s).

TABLE I
ACCELERATION CHARACTERISTICS FOR PULSES INCORPORATED IN THIS STUDY.

Type	Peak Acceleration (g)	Rate of Onset (g/s)	Rise Time (ms)
Ejection Catapult 1	20.7	228	91
Ejection Catapult 2	22.3	513	44
Helicopter Crash 1	44.3	1065	42
Helicopter Crash 2	64.9	2638	25
Helicopter Crash 3	65.0	2500	26

Differences in acceleration characteristics between simulated ejection and helicopter crash events were consistent with differences in the loading environment (Table II). Axial forces were directed in compression and bending moments were directed in flexion. Peak axial forces and bending moments increased by 18% and 117%

for a 170% increase in peak acceleration from ejection catapult to the helicopter crash pulses. Peak force and moment occurred at approximately 96 ms for Ejection Catapult 1, 45 ms for Ejection Catapult 2, 40 ms for Helicopter Crash 1, and 25 ms for Helicopter Crashes 2 and 3.

TABLE II
LOADING CHARACTERISTICS FOR PULSES INCORPORATED IN THIS STUDY.

Type	Peak Axial Force (N)	Force Rate of Onset (kN/s)	Peak Moment (Nm)
Ejection Catapult 1	5598	58	98
Ejection Catapult 2	5827	122	166
Helicopter Crash 1	7184	190	386
Helicopter Crash 2	5236	238	179
Helicopter Crash 3	7821	215	293

Each of the five specimens sustained compression-related injuries during dynamic axial loading (Table III). Injuries involved the vertebral body in each specimen and the lamina in one specimen. Compression fracture locations involved the L1 spinal level for both ejection catapult pulses and L1-L4 levels for helicopter crash pulses, indicating possible inferior migration of fracture location for higher severity tests. Each specimen subjected to helicopter crash pulses demonstrated compression-related injuries at spinal levels caudal to L1, with involvement of the L3 and/or L4 vertebral bodies in two of the three specimens. The specimen that did not sustain L3 or L4 fractures (Helicopter Crash 1) was subjected to the least severe of the helicopter crash pulses with 30% lower peak acceleration and a rate of onset less than half of Helicopter Crash pulses 2 and 3.

TABLE III
DETAILS OF FRACTURES SUSTAINED BY PMHS SPECIMENS.

Type	Affected Spinal Level	Description
Ejection Catapult 1	L1	Anterior cortex fracture with endplate fracture
Ejection Catapult 2	L1	Burst fracture with retropulsion into spinal canal
Helicopter Crash 1	L1	Anterior compression
	L2	Anterior compression
Helicopter Crash 2	L3	Vertical cortical fracture with laminar fracture
	L4	Mild compression fracture of the anterior cortex
Helicopter Crash 3	L2	Anterior body fracture including cranial endplate
	L2	Vertical cortical fracture of posterior wall
	L3	Burst fracture with retropulsion into spinal canal

Although compression was the primary mechanism for injuries listed in Table III, multiple fracture types/locations were evident across the sample. The fractures described above were judged to have occurred during the initial loading phase of the vertical acceleration pulse.

IV. DISCUSSION

The purpose of this preliminary investigation was to determine fracture patterns resulting from different high-rate axial loading scenarios encountered in military environments. Two environments in which lumbar spine fractures commonly occur are the catapult phase of aviator ejection and helicopter crashes. This study incorporated a novel experimental model to subject PMHS whole lumbar spines to axial accelerations simulating those environments. Aviator ejection acceleration pulses were less severe in terms of peak acceleration and rate of onset than helicopter crash pulses. Post-test examination of specimens using lateral x-rays and sagittal and axial CT scans demonstrated caudal migration of fracture locations with increasing acceleration pulse severity, although compression was the mechanism for a majority of injuries. Aviator ejection accelerations resulted in fracture patterns affecting primarily the L1 spinal level. However, more severe helicopter crash accelerations resulted in fracture patterns distributed between L1 and L4 spinal levels. The lower severity helicopter crash pulse resulted in injuries at L1 and L2 and the higher severity helicopter crash pulses resulted in injuries between L2 and L4. Although the sample size for this ongoing study was limited, with two specimens subjected to the ejection pulse and three to the helicopter crash pulse, fracture patterns from these five specimens demonstrate that injuries occurred lower in the lumbar spine for more severe loading environments, affecting the cranial end of the thoracic spine for aviator ejections and primarily the middle to lower lumbar spine for helicopter crashes. These distributions are consistent with actual injuries sustained by military personnel in those environments as reported in a recent review of military trauma during axial loading scenarios [7]. That study reported that ejection-related injuries were distributed across the spine from cervical to lumbar regions, although specific injury concentrations occurred at the mid-thoracic spine (T6-T9) and at the thoraco-lumbar junction (T11-L2). However, injuries sustained in helicopter crashes were more focused to the caudal spine, with 61% of injuries occurring at the T12 to L4 levels.

A number of factors may have contributed to changing fracture locations within the lumbar spine. Those factors include biomechanics of the environment (e.g., loading or acceleration metrics) and specimen-related aspects such as the column length, curvature, and bone quality. Consistency in specimen age, preparation, and initial position for each test likely minimized outcome differences due to specimen-related aspects. Biomechanical metrics generally increased from ejection catapult to helicopter crash pulses incorporated in this study (Tables 1 & 2). Notable because of its implication in injury tolerance, peak axial forces (i.e., fracture force) increased from ejection catapult to helicopter crash pulses. Increased fracture force can be attributed to loading rate effects or differing fracture thresholds for caudal vertebrae, although other factors may play a role. With regard to fracture tolerance at different spinal levels, conflicting evidence exists in literature regarding the level dependence for vertebral body ultimate strength within a specific spinal region. It has generally been accepted that fracture tolerance varies from region to region (e.g., thoracic to lumbar spines), but does not vary remarkably from level to level within a specific region [15]. However, some experimental studies quantifying compressive tolerance of lumbar vertebrae under quasi-static loading have demonstrated evidence of increasing compressive strength from L1 to L4 vertebrae [16, 17]. For example, Hansson et al. demonstrated 15% increasing compressive tolerance from L1 to L2 and 47% increase from L1 to L4 [16]. These studies demonstrating spinal level dependence in lumbar spine fracture mechanics were conducted at primarily quasi-static rates. To our knowledge, similar studies have not been performed under dynamic conditions in the lumbar spine. However, Kazarian et al. demonstrated spinal level dependence in the thoracic spine under dynamic conditions [18]. A possible explanation for inferiorly migrating injuries under more severe axial loads could be attributed to increasing fracture tolerance at caudal spinal levels. Depending on the rate of force/acceleration onset, load levels may exceed fracture tolerance of caudal vertebrae more quickly, with the fracture dissipating energy and sparing cranial levels.

Another factor that can modulate fracture type and location is spinal orientation at the time of loading. All specimens in the present study were pre-flexed with a 5-Nm load to attain consistency in initial orientation. However, the present experimental model provides flexibility in terms of spinal orientation at the time of load application through pre-bending (e.g., flexion, extension, lateral bending) or by changing the location of the "torso" load application relative to the cranial aspect of the specimen [14]. In this manner, the effect of different lumbar spine orientations (modeling different vehicle occupant postures) can be investigated. Limited data exist in the literature with regard to this factor. Langrana et al. [19] demonstrated considerably lower tolerance for three-vertebra thoraco-lumbar preparations tested in neutral posture than for those pre-extended at the time of dynamic compressive load application. Likewise, two studies by Yoganandan et al. demonstrated considerably lower tolerance for thoraco-lumbar spinal columns tested under dynamic compression in neutral [13] than pre-flexed positions [20]. Therefore, spinal orientation likely affects dynamic lumbar compressive tolerance and delineation of the effects of this factor remains a focus of this ongoing research.

Although other fractures occurred, some of the fracture types common to multiple specimens are discussed in more detail here. Fracture types can be defined according to the three-column concept, as described by Denis [21, 22], wherein the anterior column consists of the anterior vertebral body, anterior intervertebral disc, and anterior longitudinal ligament, the middle column consists of the posterior vertebral body, the posterior intervertebral disc, and the posterior longitudinal ligament, and the posterior column consists of the posterior ligamentous complex including the pedicles, facet joint, laminae, and spinous processes. Due to the involvement of anterior and posterior columns or the middle column, these injuries can be attributed to axial compression. Vertical fractures affecting the anterior (Helicopter Crash 2) or posterior (Helicopter Crash 3) cortex were sustained at L2 and L3 by specimens subjected to the most severe acceleration pulses, as shown in

Figure 2. Those fractures occurred in the caudal-cranial direction at approximately 1/3 the anterior-posterior body depth from the anterior or posterior cortex and involved the anterior/posterior cortex and cranial endplate, although integrity of the caudal endplate was not affected. One of those specimens also sustained a longitudinal fracture through the lamina and proximal aspect of the spinous process, and also affecting the pars interarticularis (Fig. 2, right).

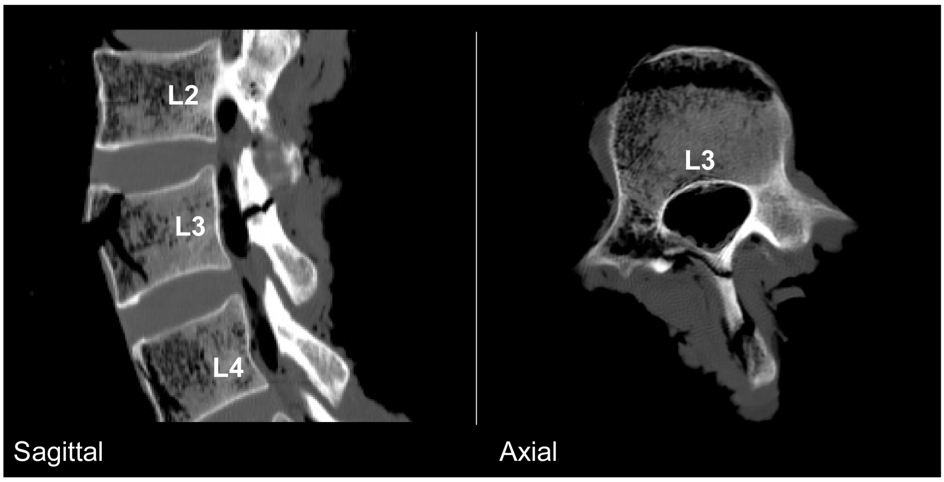

Fig. 2. Fractures sustained at the L3 spinal level during the Helicopter Crash 2 test.

Burst fractures affecting anterior and posterior cortices as well as cranial and caudal endplates were not confined to one loading type (i.e., ejection catapult or helicopter crash) and occurred at different spinal levels. The specimen subjected to the Ejection Catapult 2 accelerations sustained a burst fracture at L1 that included retropulsion of bony fragments into the spinal canal. Likewise, the specimen subjected to the Helicopter Crash 3 acceleration sustained a massive burst fracture at L3 that also included retropulsion of fragments into the canal (Figure 3). Burst fractures occur under pure compression mechanisms due to uniform load applied to anterior and posterior cortices and resulting in fracture of both regions. Some evidence of compression-bending was also evident from post-test CTs. The specimen subjected to the Helicopter Crash 3 acceleration sustained an anterior body compression fracture (i.e., wedge fracture) that included greater compression of the anterior than the posterior cortex. The specimen subjected to the Helicopter Crash 1 acceleration also sustained anterior compression fractures at L1 and L2. The mechanism for this type of fracture is compression-flexion, resulting in higher loading concentration on the anterior cortex.

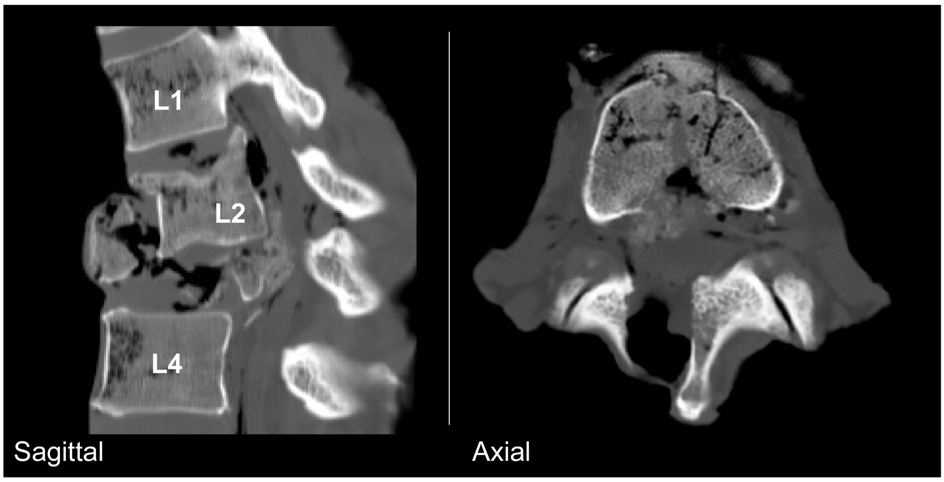

Fig. 3. Fractures sustained at L2 (left & right) and L3 (left) spinal levels during the Helicopter Crash 3 test.

Results from this preliminary experimental study are supported by clinical findings from traumatic military environments. Spinal injuries continue to occur in military environments despite the fact that ejection accelerations and energy absorbing seats are constructed based on design factors for occupant safety.

Acceleration characteristics incorporated in the experiments described in this manuscript were biased toward the magnitudes and onset rates that are historically known to cause spinal injury and, as such, likely represent the higher end of ejection catapult and helicopter crash pulses. Continued testing under the conditions presented in this manuscript will define the role of specific biomechanical metrics on injury outcomes (i.e., fracture type and location) from high rate axial loading. Using the present model, the roles of peak acceleration and rate of onset can be independently investigated. Additionally, occupant orientation at the time of loading can be investigated by controlling initial specimen position and anterior-posterior location of the applied load from the upper platform. Each of these factors likely influence injury type and location following high-rate axial loading in military environments and will be important in the development of advanced safety devices to prevent injury or assess the likelihood of injury under specific loading scenarios. Additionally, acceleration characteristics can be modified to simulate other military or loading scenarios.

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VI. References

[1] Shanahan, D F and M O Shanahan, Kinematics of U.S. Army helicopter crashes: 1979-85, *Aviat Space Environ Med*, 60, 2, 112-21, 1989.
[2] Jackson, K E, E L Fasanella, R L Boitnott, J McEntire, and A Lewis, Occupant responses in a full-scale crash test of the Sikorsky ACAP helicopter, 2002. National Aeronautics and Space Administration (NASA), TM-2002-211733.
[3] Shanahan, D F and M O Shanahan, Injury in U.S. Army helicopter crashes October 1979-September 1985, *J Trauma*, 29, 4, 415-22; discussion 423, 1989.
[4] Ragel, B T, C D Allred, S Brevard, R T Davis, and E H Frank, Fractures of the thoracolumbar spine sustained by soldiers in vehicles attacked by improvised explosive devices, *Spine (Phila Pa 1976)*, 34, 22, 2400-5, 2009.
[5] Poopitaya, S and K Kanchanaroek, Injuries of the thoracolumbar spine from tertiary blast injury in Thai military personnel during conflict in southern Thailand, *J Med Assoc Thai*, 92 Suppl 1, S129-34, 2009.
[6] Helgeson, M D, R A Lehman, Jr., P Cooper, M Frisch, R C Andersen, et al., Retrospective review of lumbosacral dissociations in blast injuries, *Spine (Phila Pa 1976)*, 36, 7, E469-75, 2011.
[7] Stemper, B D, N Yoganandan, G R Paskoff, R J Fijalkowski, S G Storvik, et al., Thoracolumbar spine trauma in military environments, *Minerva Ortopedica E Traumatologica*, 62, 5, 397-412, 2011.
[8] Sances, A, Jr., J B Myklebust, D J Maiman, S J Larson, J F Cusick, et al., The biomechanics of spinal injuries, *Crit Rev Biomed Eng*, 11, 1, 1-76, 1984.
[9] Yoganandan, N, G Ray, F A Pintar, J B Myklebust, and A Sances, Jr., Stiffness and strain energy criteria to evaluate the threshold of injury to an intervertebral joint, *J Biomech*, 22, 2, 135-42, 1989.
[10] Wilcox, R K, T O Boerger, D J Allen, D C Barton, D Limb, et al., A dynamic study of thoracolumbar burst fractures, *J Bone Joint Surg Am*, 85-A, 11, 2184-9, 2003.
[11] Willen, J, S Lindahl, L Irstam, B Aldman, and A Nordwall, The thoracolumbar crush fracture. An experimental study on instant axial dynamic loading: the resulting fracture type and its stability, *Spine (Phila Pa 1976)*, 9, 6, 624-31, 1984.
[12] Shirado, O, K Kaneda, S Tadano, H Ishikawa, P C McAfee, et al., Influence of disc degeneration on mechanism of thoracolumbar burst fractures, *Spine (Phila Pa 1976)*, 17, 3, 286-92, 1992.
[13] Yoganandan, N, S J Larson, F Pintar, D J Maiman, J Reinartz, et al., Biomechanics of lumbar pedicle screw/plate fixation in trauma, *Neurosurgery*, 27, 6, 873-80; discussion 880-1, 1990.
[14] Stemper, B D, S G Storvik, N Yoganandan, J L Baisden, R J Fijalkowski, et al., A new PMHS model for lumbar spine injuries during vertical acceleration, *J Biomech Eng*, 133, 8, 081002, 2011.
[15] Yamada, H and F G Evans, *Strength of biological materials*1970: Williams & Wilkins.
[16] Hansson, T, B Roos, and A Nachemson, The bone mineral content and ultimate compressive strength of lumbar vertebrae, *Spine (Phila Pa 1976)*, 5, 1, 46-55, 1980.
[17] Gozulov, S A, V A Korzhen'iants, V G Skrypnik, and N Sushkov Iu, [Study of the durability of the human vertebrae under compression], *Arkh Anat Gistol Embriol*, 51, 9, 13-8, 1966.

[18] Kazarian, L and G A Graves, Compressive strength of the human vertebral centrum, *Spine (Phila Pa 1976)*, 2, 1, 1-14, 1977.

[19] Langrana, N A, R R Harten, D C Lin, M F Reiter, and C K Lee, Acute thoracolumbar burst fractures: a new view of loading mechanisms, *Spine (Phila Pa 1976)*, 27, 5, 498-508, 2002.

[20] Yoganandan, N, F Pintar, A Sances, Jr., D Maiman, J Mykelbust, et al., Biomechanical investigations of the human thoracolumbar spine, *SAE Technical Papers*, 881331, 1988.

[21] Denis, F, Spinal instability as defined by the three-column spine concept in acute spinal trauma, *Clin Orthop Relat Res*, 189, 65-76, 1984.

[22] Denis, F, The three column spine and its significance in the classification of acute thoracolumbar spinal injuries, *Spine (Phila Pa 1976)*, 8, 8, 817-31, 1983.