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A Review of Computational Spinal Injury Biomechanics Research and Recommendations for Future Efforts

by Reuben H. Kraft and Samantha L. Wozniak

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14. ABSTRACT
Developing an understanding of the high rate injury biomechanics associated with blast-induced loading to the spine, lower extremities, central nervous system, and peripheral nervous system is critical to optimize protective equipment for Soldiers. Computational methods and high performance computing are an essential part of the toolset needed to develop the understanding of spinal injury caused by high strain rate loading. These methods provide a wealth of information that otherwise may not be available. While the research area of computational spinal injury biomechanics has received much attention in the past with regard to low strain rate degenerate effects, there are significant opportunities to enhance the state of knowledge of spinal injury in the area of high strain rate military loading. Through this literature review and careful contemplation, a list of six recommendations for future research directions has been developed.

15. SUBJECT TERMS
spinal injury, computational modeling, finite element analysis, explosion, blast

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Preface

Since this is a joint publication between a government employee and a government contractor, a break down of the report is offered to show that it is in the best interest of the government to have a combined government/contractor report. This review article was designed by Reuben Kraft to offer a precise overview of a subtopic within the large research field of spinal injury biomechanics, namely computational spinal injury biomechanics. Samantha Wozniak, a Bowhead, Inc. contractor, significantly contributed to the authorship as she gathered references and wrote summaries for targeted research topics dictated by Kraft. Kraft and Wozniak worked closely to shape this report each making many modifications of their own. If asked to divide the report into ”who did what,” it would be extremely difficult but would look close to the following:

- Introduction: Kraft and Wozniak
- Anatomy Section: Wozniak
- Injury Section: Kraft and Wozniak
- Vertebra Models Section: Wozniak
- Intervertebral Disc Models Section: Wozniak
- Segment Models Section: Wozniak
- System Models Section: Kraft
- Failure Models Section: Kraft
- Uncertainty Quantification Section: Kraft
- Conclusion and Recommendations Section: Kraft

1. Introduction

The incidence of spine injuries among American military personnel is currently estimated to be at its highest in comparison to the last 60 years (*1*). While injuries to the neck and spine only account for 5% of the total injuries seen in the current conflicts, these injuries are commonly the most debilitating and costly for the warfighter (*2*). Spinal injuries that result in spinal cord impingement are of particular concern because they may result in paralysis, loss of bowel and bladder function, inability to regulate automatic functions of the body, and many other complications.

The classifications of spinal injuries are related to the abnormal forces acting on the central axial spinal pillar and the accompanying exaggerated movements that cause spinal instability. The magnitude and type of injuries produced by an explosion are a result of many factors including the type and amount of explosive material, delivery method, the surrounding environment, and the distance between the victim and the blast. Injuries induced by explosions can be divided into four classes: primary, secondary, tertiary, and quaternary blast injury. Primary blast injury is a result of the sudden increase in air pressure after the blast while secondary blast injury is due to projectiles from the blast. Tertiary blast injury is due to the thrust of the victim against stationary objects or wind disruption, and quaternary blast injury is a result of fire and heat generated by the explosion, toxic effects, or post-incident risks (*3, 4*).

A study published in 2010 characterized combat-related spinal injuries sustained by a Brigade Combat Team (BCT), the basic deployable unit in the U.S. Army (*1*). The BCT involved in this study participated in *The Iraq War Troop Surge of 2007*, a major operation of the Iraq conflict. Twenty-nine of the 4,122 Soldiers deployed for 15 months sustained combat spine injuries (*1*). Twenty-three (74%) of these injuries were from improvised explosive devices. Four spine injuries were due to vehicular accidents and a gunshot wound led to one spinal injury. Specific spinal injuries sustained by these service members include lumbar burst fractures, compression fractures, nondisplaced type III fracture of the odontoid, traverse process fractures, and a flexion-distraction injury. Seven Soldiers were considered polytraumas, including one who experienced neurologic compromise due to a burst fracture.

The purpose of this report is to review recent literature on computational spinal injury biomechanics modeling for high strain rate loading, identify gaps in the research, and formulate recommendations for future efforts. This review does not aim to be an exhaustive list of papers that use finite computational techniques to analyze the spine, but rather focuses on current state-of-the-art techniques with regard to high rate injury and failure modeling. Models of the spine range from the sub-micron scale of vertebral bone to models of the whole spine and surrounding structures. This review focuses on models of individual spinal components and small segments of the spine at the macro scale. First, in sections 1.1 and 1.2, we present the spinal anatomy and a biomechanical description of spinal injuries sustained by Soldiers during combat, including the length and time scales associated with blast-induced spinal injury. Next, in section 2, a brief history of spine modeling is discussed, followed by a more in depth review of vertebra, intervertebral disc, segment, and full spine finite element models. In addition, section 2 includes a discussion of computational approaches for modeling spinal injury mechanisms, such as fracture and various techniques for handling uncertainty quantification. Finally, in section 3 we conclude with a summary of the state-of-the-art finite element model techniques and suggest recommendations for future research.

1.1 Anatomy of the Spine

The spine has many functions including to protect the spinal cord and associated nerves, allow for movement, support the body frame in an upright position, allow for flexibility, provide a structural foundation for the shoulder and pelvic girdles, prevent and absorb shock, and provide a structural base for rib attachments which protect the heart and lungs. The spinal column is an intricate structure, comprised of vertebrae and cartilage discs, known as intervertebral discs, stacked alternatively on top of each other. The column starts at the base of the skull and continues to the pelvis. As seen in figure 1a, the spine is divided into five regions: the cervical spine, the thoracic spine, the lumbar spine, the sacrum, the coccyx. Each individual vertebra is named by referring to the first letter of the region (e.g., "L" for lumbar), and, starting with the most superior vertebra in that region, numbered consecutively until the most inferior vertebra in the region has been named.

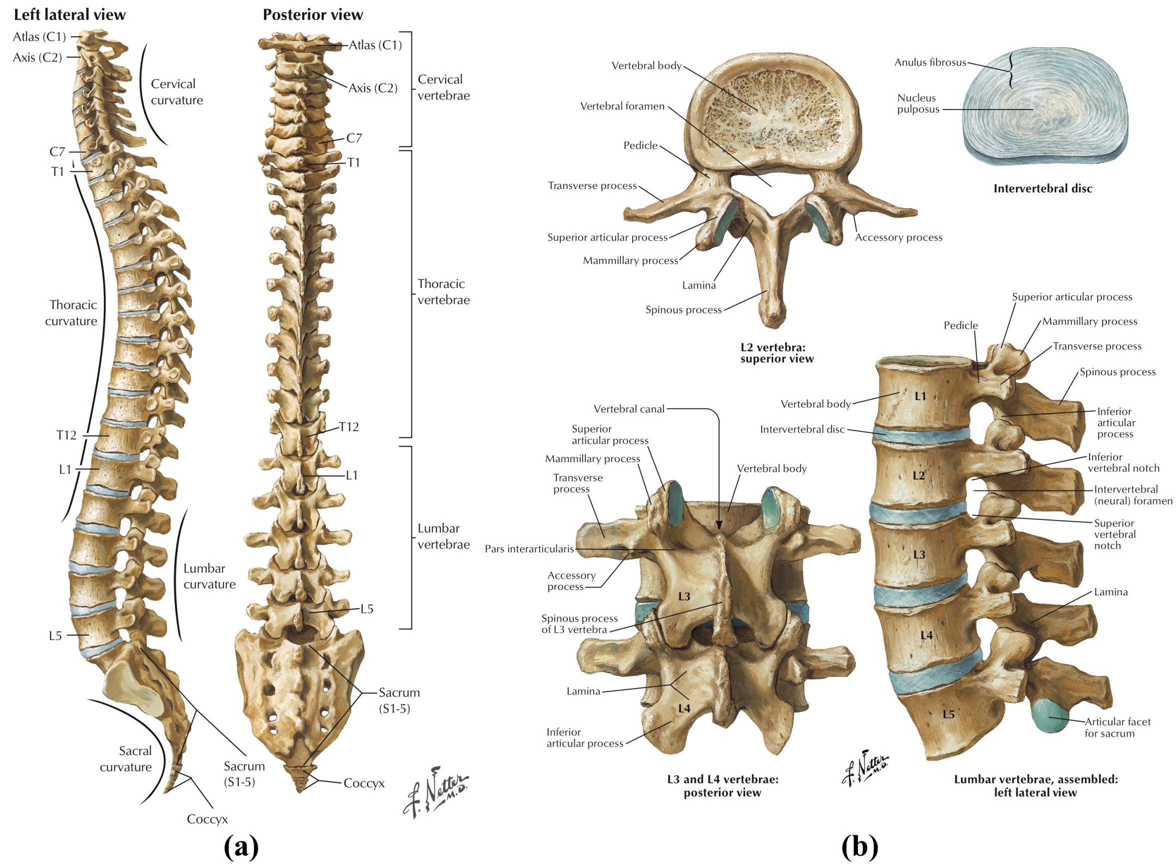


Figure 1. Bony anatomy of the (a) spinal column and (b) vertebra. Netter illustration from www.netterimages.com.

Vertebrae differ in size and shape between regions and also within a region. For example, the 12 vertebrae in the thoracic spine (commonly referred to as T_1-T_{12}) each have two thoracic facet joint sets– one set on the top and one set on the bottom– which are responsible for providing the middle back with the flexibility and strength it needs. A typical vertebra consists of two main parts: an anterior vertebral body and posterior vertebral arch. As seen in figure 1b, the lumbar vertebral arch is formed by a pair of pedicles, and laminae and supports seven processes: four articular, two transverse, and one spinous (note that T_1-T_{12} have six articular processes). The principal biomechanical function of the vertebral body is to support the daily loads of the spine. Accordingly, the vertebral bodies graduate in size from the cervical to the lumbar region. The microstructure of the vertebral body is comprised of highly porous trabecular bone and a thin, dense cortical shell. The trabecular bone supports the majority of the vertical compressive loads while the outer cortical shell helps resist torsion and shear forces (*5*). A thin layer of semi-porous subchondral bone forms a boundary layer between the cancellous (synonymous with trabecular bone) core of the vertebral body and the intervertebral disc, known as the vertebral end-plate. The end-plates of a healthy disc prevent the highly hydrated nucleus from bulging into the adjacent vertebral bone, while simultaneously absorbing hydrostatic pressure that results from mechanical loading of the spine (*6*). In many respects the structure is similar to that of articular cartilage, and while the end-plate cartilage is not connected directly to the bone of the vertebral bodies, it does have direct connections with the disc through the lamellae of the inner (medial) annulus fibrosus (*7*). In the adult disc the end-plates typically are less than 1 mm thick, and although this varies considerably across the width of any disc, there is a tendency for the end-plate to be thinnest in the central region adjacent to the nucleus pulposus (*8*).

Intervertebral discs lie between adjacent vertebrae and form a cartilaginous joint that allows for motion between the vertebrae. These structures help absorb compression forces, support bending loads and resist rotation, tension, and shear (*9*). Discs are complex structures made up of an outer ring of fibrous cartilage (type I and II), also known as the annulus fibrosus, and an inner gelatinous core termed the nucleus pulposus. The nucleus pulposus contains randomly organized collagen fibers and radially organized elastin fibers (*10*). The annulus fibrosis is comprised of alternating layers of type I (outer annulus) and type II (inner annulus) collagen at 30° angles to the end-plate. This arrangement leads to an anisotropic and inhomogeneous material. In addition, the orientation of the collagen fibers makes the structure nonlinear (*11*).

Another load-bearing junction between the vertebrae is the zygapophysial joint (zygapophyseal, or facet joint). This joint is formed between the superior articular process of one vertebrae and the inferior articular process of the vertebrae directly above it as schematically depicted in figure 1b. In each spinal segment there are two facet joints, which, in combination with the disc, allows for motion including flexion, extension, rotation, and lateral bending. These joints also may restrict movement depending on their orientation; they limit flexion in the thoracic spine and rotation in the lumbar spine (*12*). Similar to other joints in the body, each facet joint is surrounded by a capsule of connective tissue and covered by articular cartilage. Within the joint capsule, the joint is lined by a membrane known as the synovium. Synovial fluid surrounds the joint and is compromised of hyaluronic acid, lubricin, proteinases, and collagenases. The lubricin found in the synovial fluid allows the cartilage to move with very little friction. The coefficient of friction for synovial joints in humans has been estimated to be 0.01 for static friction and 0.003-0.0057 for kinetic friction (*13, 14*).

Excessive movement of the spinal components is limited by the spinal ligaments. Ligaments, in combination with muscles and tendons, help preserve the natural articulation of the spine and stabilize the components. There are two primary ligament systems in the spine: the intersegmental and intrasegmental systems (*15*). The intrasegmental system holds individual vertebrae together and is made up of the ligamentum flavum, interspinous ligament, intertransverse ligament, and facet capsule. The intersegmental system holds many vertebrae together and includes the anterior longitudinal ligament, posterior longitudinal ligament, and supraspinous ligament.

Within the vertebral canal is the spinal cord, which is protected by three tissue layers, known as meninges. Beginning as a continuation of cranial dura mater, the outer protective membrane, spinal dura mater, extends to the level of the second sacral vertebra (*16*). Between the dura mater and the vertebrae is a space, known as the epidural space, filled with adipose tissue and a network of blood vessels. The middle layer, or arachnoid mater, is separated from the inner layer (pia mater) by the subarachnoid space which contains cerebrospinal fluid.

1.2 Osseous, Soft Tissue, and Neurological Injury

It is important to understand the length and time scales associated with blast-induced spinal injury. In vehicle mine blasts, tertiary blast effects tend to be the most common injury mechanism (*4*). Within 0.5 ms of the initiation of the explosion, a shock wave hits the bottom plate of the vehicle and causes a large peak in pressure. This pressure results in local acceleration and subsequent deformations of the plate, which apply significant axial loads to the Soldier and may result in spinal, pelvic, and lower limb fractures (*17*). In 2007, the North Atlantic Treaty Organization (NATO) published a technical report, *Test Methodology for Protection of Vehicle Occupants against Anti-Vehicular Landmine Effects*, which provided information about the process of vehicular mine blasts (*17*). As described in the report, as the blast wave reflects under the vehicle, a pressure force acts on the bottom of the vehicle and causes it to lift off the ground. The acceleration of the vehicle and the collisions that follow are a significant source of injury, especially if the Soldier is not properly restrained. As seen in figure 2 (reprinted from the NATO technical report), the body is subjected to both the local effects (caused by shock and high rate deformation of the vehicle) and global effects (vehicle motion over time) of the vehicle mine detonation process. Drop-down effects and subsequent events, such as rollover, are not considered due to a high variability in boundary conditions.

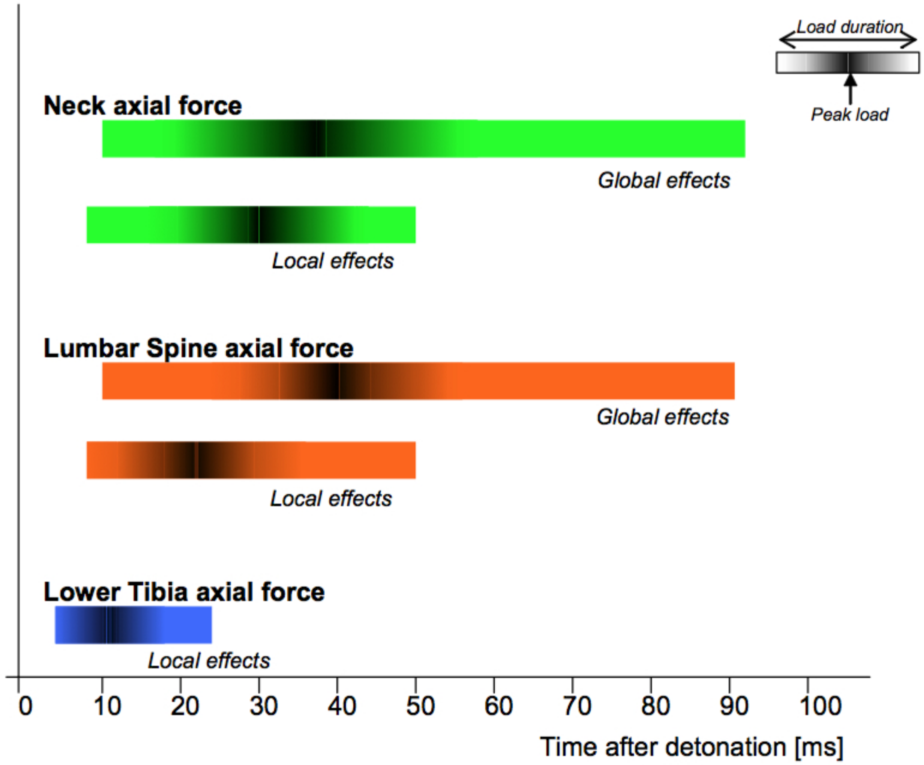


Figure 2. Time scales associated with the loading process in the human body inside a vehicle due to mine detonation (*17*). Figure courtesy of the NATO Research and Technology Organization/Human Factors and Medicine Panel.

Spinal injury in Soldiers is often caused by high accelerative loading and may result from explosions and vehicular accidents. Mechanisms of spinal injury include axial deformation (compression or tension); torsion or axial rotation; segmental translation (shear); and simultaneous or successive combination of the previous three mechanisms (*18*). Principal applied loading mechanisms that may lead to spinal injury are shown in figure 3. The thoracolumbar junction (TLJ), T_{12}-L_2, is a common site of traumatic injuries, accounting for 30–60% of all spinal injuries (*19*). However, in current military operations, for spinal injury cases involving underbelly blast to a vehicle, the common site of burst fractures is mid-lumbar. This is unusual for spinal fractures which tend to happen at C_7-T_1 and T_{12}-L_1. This difference may be due to rate of loading, posture, and armor load on the Solider (*12, 20*).

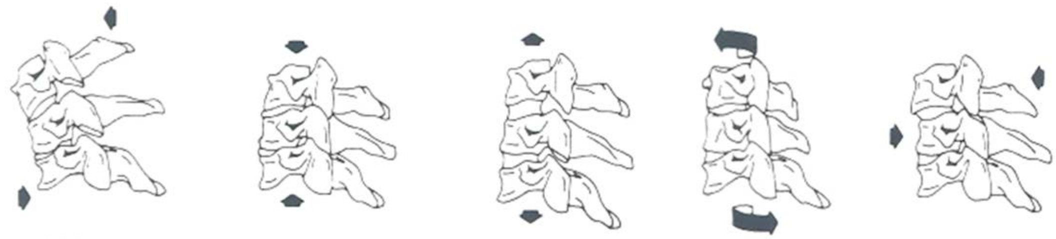


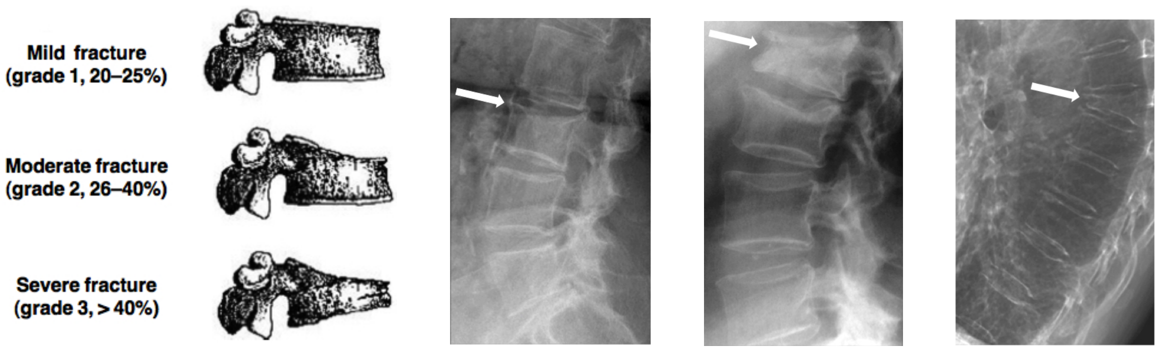
Figure 3. Principal applied loading mechanisms that may lead to spinal injury (*21*). Reprinted with permission from Springer Science + Business Media.

Trauma to the spine may result in vertebral fractures, degenerative disc disease, spinal cord contusions, lesions of connecting ligaments, and other disabilities. Some common disabilities include protrusions of cartilage of the intervertebral disc through the vertebral body endplate into the adjacent vertebra (Schmol's nodes); degenerative osteoarthritis of joints between the spinal vertebrae and neural foraminae (spondylosis, narrowing of space between vertebrae); and anterior displacement of vertebra in relation to the vertebrae below (spondylolisthesis) (*22*). Among adults, the most common injuries are vertebral column fractures and fracture dislocations (*23*).

Aside from visible injuries to the spinal cord, such as fractures and crushing, Soldiers can also experience damage to the nerve roots or myelinated fiber tracts that are responsible for carrying signals to and from the brain. Injuries to the neck (C_1-C_7) tend to be more life-threatening since they may affect the phrenic nerve, a nerve originating from the cervical spine that is responsible for the movement of the diaphragm (*17*). There is also increasing evidence that the central nervous system may be damaged as a direct effect of primary blast (*24, 25*). Systemic acute gas embolism, a result of pulmonary disruption from the blast wave, may result in occluded blood vessels in the brain or spinal cord (*24*).

The spine is capable of withstanding large amounts of pressure (spinal injuries typically do not develop below accelerations of 20–25 G) and the construction of the neural arch in the vertebrae provides significant resistance to forward shear (*26, 27*). However, as the lateral trunk velocity increases, the lateral shear forces increase proportionally and more intensely than other spinal forces (*27*). Compression forces also increase as velocity increases, and thus when the combination of lateral shear and compression forces are too great for the vertebra to withstand, fractures may occur in the pedicle, lamina, transverse, or spinous process; facet; or vertebral body (*17*). The severity of these fractures will depend on the structure stability of the fracture, i.e., whether or not the bony and ligamentous integrity is compromised such that free movement can occur. Even with a stable fracture, the spine is still capable of carrying and distributing a person's weight. Unstable fractures involve an abnormal motion of the fracture site, which could potentially impinge the spinal cord or its vascular supply, resulting in pain or loss of neurological function. Neurologic deficit is present in 10–30% of all spinal column injuries (*23, 28*). Typically, the degree of associated neurological injury is correlated with the extent of the vertebral column fracture and any accompanying dislocation or subluxation.

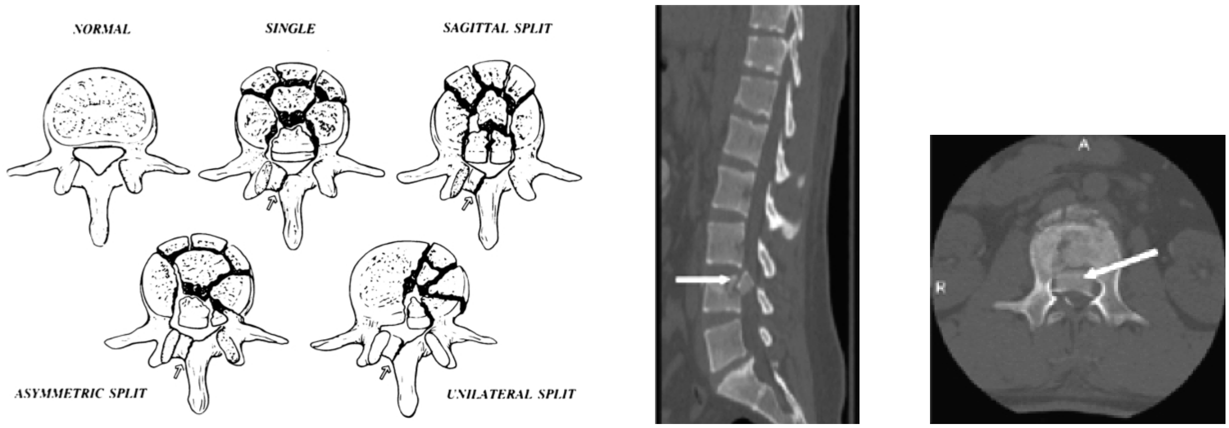
Axial compressive loads commonly cause compression, burst, or chance fractures. A wedge compression fracture occurs when there is a combination of a flexion bending moment and compressive loading, crushing the anterior aspect of the vertebra and forming a wedge shape (figure 4). Three subtypes of wedge fractures can be defined based on the height reductions in the anterior, middle, and posterior dimensions of the vertebra (*29*). The anterior column includes the anterior longitudinal ligaments and anterior one third of the vertebral body. The middle column consists of the posterior longitudinal ligament and posterior two thirds of the vertebral body. The posterior column is comprised of the facets, spinous processes, pedicles, and posterior ligamentous complex (*19, 30*). In a grade 1 fracture, only the anterior column is crushed (stable fracture). In a grade 2 fracture, the anterior column is crushed and the posterior column ligaments fail (potentially unstable fracture). The loss of anterior vertebral body height is typically $<50\%$. Figures 4b and 4c, show a grade 1 and 2 fracture, respectively, of the L_3 vertebra. In a grade 3 fracture, all three columns fail, leading to an unstable fracture. The loss of anterior vertebral body height is usually $>50\%$. A lateral radiograph of a grade 3 wedge fracture of the T_7 vertebra can be seen in figure 4d.



(a) (b) (c) (d)

Figure 4. (a) Drawings show the grading of wedge vertebral fractures. The approximate degree of height reduction determines the assignment of grade to the fractured vertebra (reproduced with permission of Mary Ann Liebert, Inc.). (b) Lateral radiograph of lumbar spine shows a mild wedge fracture (grade 1) of L_3 vertebra. (c) Lateral radiograph of lumbar spine shows moderate wedge fracture (grade 2) of L_3 vertebra and moderate crush fracture (grade 2) of L_2 vertebra. (d) Lateral radiograph of thoracic spine shows severe wedge fracture (grade 3) of T_7 vertebra (*31*). Figures b through d reprinted with permission from the American Journal of Roentgenology.

Burst fractures are typically associated with high energy trauma and are most commonly found between T_5 and T_8 and the TLJ (*29*). However, as mentioned earlier, in current military operations, for spinal injury cases involving underbelly blast to a vehicle, the common site of burst fractures is mid-lumbar. They are characterized by the crushing of the anterior and middle aspects of the vertebral body (figure 5). The state of the posterior elements determines whether the fracture is stable or unstable. Displacement of posterior elements and/or vertebral body or facet dislocation or subluxation define an unstable burst fracture (*29*). Sagittal and axial computed tomography (CT) scans of the thoracolumbar spine, seen in figures 5b and 5c, demonstrates an L_4 burst fracture with retropulsion of bone into the spinal cord. Displacement of bone fragments may lead to life-threatening injuries if they penetrate into the spinal canal and cause loss of spinal cord function (*17*). While burst fractures only account for 15% of spinal injuries, patients with burst fractures can reach up to 50–60% neurological deficit from their pre-injured state (*19*).



(a) (b) (c)

Figure 5. (a) Patterns of the retropulsed fragment. (b) Sagittal and (c) axial CT scans of the thoracolumbar spine demonstrates an L_4 burst fracture with retropulsion of bone into the spinal canal (as shown by arrows) (*32, 33*). Reprinted with permission from the American Journal of Roentgenology and Neurosurgical Focus.

While experimental studies have identified sites of failure initiation in burst loading within the literature, there is a lack of consensus to which region fails first. Several authors believe that the initial fracture occurs in the vertebral body, which leads to expulsion of bone and fluid into the spinal canal. This results in reduced support of the endplates, continuing their exposure to compression and eventually forcing them to crack (*34*). Others report that an increase in pressure within the nucleus pulposus causes the endplates to bulge and crack (*19*). Hongo et al. found high compressive and tensile strains at the base of the pedicle of T_{10}, L_1, and L_4, indicating that the base of the pedicle is the site of burst fracture initiation (*35*).

Flexion-distraction thoracolumbar spine fractures, also known as Chance fractures, commonly occur at the thoracolumbar junction and tend to be the result of violent forward flexion (bending), which causes a distraction injury (pulling apart) to the posterior elements of the vertebra. Chance fractures occur when the anterior portion of the vertebral body is minimally compressed and there is a transverse fracture through the posterior elements of the vertebra and posterior portion of the vertebral body (*2*). Several subtypes of flexion-distraction thoracolumbar spine fractures exist, including a classic Chance fracture, fulcrum fracture, and pure soft-tissue flexion-distraction injury as seen in figures 6a, 6b, and 6c, respectively. The pathophysiology of each fracture depends on the axis of flexion.

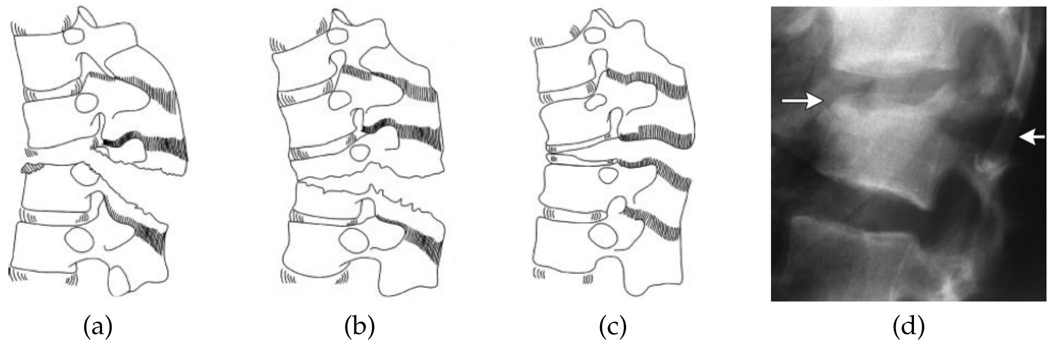


Figure 6. Variations of Chance-type flexion-distraction spinal injuries include (a) classic Chance fracture, (b) fulcrum fracture, and (c) pure soft-tissue flexion-distraction injury. (d) Lateral radiograph shows a classic Chance fracture that extends horizontally through the pedicle (right arrow) and into the vertebral body. There is an anterosuperior wedge-shaped compression fracture (left arrow) in the same vertebral body (*36*). Reprinted with permission granted from The Radiological Society of North America and author.

2. Computational Spinal Injury Biomechanics

Biomechanical models of the human body have the aptitude to support advancements in many fields including medicine, biology, and engineering. Over the past 60 years, a substantial effort has been dedicated to modeling the human body. An assortment of models that vary in fidelity have been developed and range from having one degree of freedom to models with millions of degrees of freedom. The spine is a complex component of the human skeleton, consisting of many intricate components with diverse material properties. Modeling of the spine provides opportunities to gain more understanding and a detailed description of temporal and spatial deformation processes, including failure of the intervertebral discs and vertebrae.

In general, research groups started with relatively simple models of individual components of the spine and eventually developed models that represented larger sections of the spine and surrounding anatomy. Latham (1957) is considered the first to have developed a mathematical model of the spine; focused on studying pilot ejection, he used a one-degree-of-freedom model consisting of two masses, representing the body and ejection seat, interconnected by a spring to represent the response of the spine to vertical acceleration (*37–40*). In Payne's (1961) subsequent model, the head and upper torso were modeled as a rigid body and the spine was represented by a spring-dashpot in a parallel system (*38*). Liu and Murray (1966) contributed to the modeling approach by proposing a mathematical model that consisted of a uniform rod and a rigid end-mass, representing the spinal column and head, respectively (*41*).

It wasn't until the late 1960s that segmentation of vertebrae and intervertebral discs were incorporated into models. Toth (1966) developed an eight-degree-of-freedom model in which vertebrae were modeled as rigid masses interconnected by springs and dampers, simulating the action of discs (*42*). A mathematical model developed by Orne and Liu (1971) took into account the curvature of the spine as well as inertial properties of the vertebral bodies and axial shear and bending stiffness were introduced (*43*). The interaction between the spine and the torso, however, were not considered. Prasad and King (1974) further developed this model to include the interactions of the articular facets (*44*).

Up until this point, models did not take the rib cage or viscera into account. Belytschko et al. (1976) developed the first three-dimensional representation of the entire spine and several other models with increasing complexity (*45*). These finite element models consisted of the head, vertebrae (lumbar, thoracic, and cervical), ribs from T1 to T10, intervertebral discs, articular facets, ligaments, muscle, connective tissue, upper and lower viscera, and the pelvis. Skeletal components were modeled as rigid masses, intervertebral discs as beam elements, ligaments as spring elements, and articular facet interactions as hydrodynamic elements (*38, 40, 46*).

Finite element modeling continues to be popular in spinal research. When constructing a finite element model, three primary areas need to be addressed: geometric representation including segmentation of spinal components; constitutive models and corresponding material properties of the components; and boundary conditions used to apply loading and constraints. More recently, model components are developed with geometries that match that of a specific in vitro specimen or that represent an average vertebra. It is important to define the geometry of the structure as accurately as possible because it greatly influences deformation (*47–49*). Generic models are developed from anatomical measurements while specific geometries can be formed from CT and magnetic resonance imaging (MRI) scans. Software packages convert data obtained from these images into a format that can be interpreted by solid modeling software and then input into finite element software. Wilcox et al. (2007) found that finite element results demonstrated a greater agreement with previous experimental results when the models were morphologically accurate as opposed to generic models (*47*). It should be noted, however, that the accuracy of image-based models is affected by the original resolution of the medical image (*50*).

Constitutive models and their parameters are also a critical aspect of developing a finite element model. Determining the properties of hard and soft tissues, and fluids, as well as discovering their complex interactions and changes over time, are a new set of challenges currently facing researchers. To date, a wide range of material values have been used in finite element models of the spine due to the large assortment of measurement techniques, testing strain rates, and sample preparation (table 1). The verification of mesh and boundary conditions and the sensitivity of results to material properties, however, has been limited (*47*) .

Table 1. Compilation of various constitutive models and parameters found in recent computational models of the spine.[1]

Anatomic Component		Material Model	Material Properties	References
Vertebra	Trabecular bone	Plastic with kinematic yield, Poroelastic	E = 0.1-2.2 GPa v = 0.2 σ_Y = 42-110 MPa	(*34*, *51–56*)
	Cortical bone	Plastic with kinematic yield, Elastic	E = 5-22 GPa v = 0.3 σ_Y = 132 MPa	(*34*, *52–55*)
	Posterior elements	Elastic	E = 1000-3500 MPa v = 0.25	(*52–55*, *57*)
	Cartilaginous endplates	Poroelastic	E = 20-25 MPa v = 0.1-0.4	(*34*, *53*, *54*)
	Facet cartilage	Elastic	E = 11 MPa v = 0.4	(*53*)
Intervertebral disc	Nucleus pulposus	Poroelastic, Hyperelastic	K = 1.667 GPa E = 0.15-4 MPa v = 0.4999	(*34*, *52–55*, *57*, *58*)
	Annulus fibrosis	Orthotropic elastic, Hyperelastic, Poroelastic	E = 2-450 MPa v = 0.3-0.45	(*51*, *53–55*, *58*, *59*)
Ligaments	Anterior longitudinal ligament	Hyperelastic	σ_Y = 22.5 MPa ε_Y = 0.2 E = 7.8-20 MPa v = 0.3	(*54*, *60*, *61*)
	Posterior longitudinal ligament	Piecewise linear, Hyperelastic	σ_Y = 16-20 MPa ε_Y = 0.3-0.45 E = 10-20 MPa v = 0.3	(*34*, *52*, *54*, *60*, *61*)
	Supraspinous ligament	Hyperelastic, Nonlinear elastic	E = 3.4-15 MPa v = 0.3	(*54*, *55*, *61*)
Spinal cord		Elastic	E = 0.26-1.3 MPa v = 0.35-0.49	(*34*, *62*)
Dura mater		Anisotropic elastic	$E_{rr} = E_{\theta\theta}$ = 142 MPa E_{zz} = 0.7 MPa	(*34*)

[1]Not all parameters are listed for each constitutive model. Furthermore, not all spinal components are listed. E is the Young's Modulus, ν is Poisson's ratio, σ_Y is the yield strength, ϵ_Y is the yield strain, and K is the bulk modulus. The symbols rr, $\theta\theta$, and zz denote directions in the polar coordinate system.

2.1 Models of the Vertebra

The vertebral body have several features that should be considered in the development of its simulation. The vertebral body is comprised mainly of porous cancellous bone, with denser cortical bone at the periphery (*11*). Several groups have shown that varying material properties of the cancellous and cortical bone that make up the vertebral body, as well as changes to the material properties of the cartilaginous endplates, do not result in a significant change in biomechanical response (*11, 51*). However, these studies did not consider high-rate failure where fracture may be sensitive to inhomogeneous material properties. Thus, it may be important to consider in the future. Another important feature of vertebrae are their complex posterior geometry, which presents a challenge when meshing that area. Some researchers have simplified the geometry by developing models that exclude posterior elements and focus solely on the vertebral body (*47*, *58*, *63–66*).

Computational models of isolated single vertebrae allow researchers to investigate the natural properties of the vertebra, without having to take into consideration soft tissue or ligament interactions. The principal clinical application for detailed modeling of a single vertebra is to provide insight into the loading regime within the spine. Current models are able to accurately predict compressive strength and are typically validated through direct validation by comparing model predictions with stiffness and strength values measured in vitro (see the appendix for a review on computational model verification, sensitivity and validation) (*54*). Single vertebrae models have also been developed to study the effects of metastatically compromised vertebral segments (*58, 67*) as well as simulate the presence of bone cement after a vertebroplasty procedure (*65*, *66*, *68*, *69*).

In the future, researchers would like to use these finite element models to predict fracture risk in vivo (*54*). The ability to non-invasively assess vertebral strength is critical for the diagnosis and management of osteoporosis. The current clinical standard to predict vertebral strength is to perform a bone mineral density (BMD) assessment through the use of dual energy x-ray absorptiometry (DXA) (*70*). Finite element models derived from quantitative computed tomography (QCT) scans, however, have been shown to better predict the vertebral bone strength than BMD techniques due to their ability to assess geometry and include heterogeneous mechanical properties of the bone (*70–72*). Several groups have loaded vertebral specimens at strain rates up to 0.5% until failure and then analyzed finite element models under matching conditions. The models were able to predict yield loads, strain values, and stiffness values, which can be used to predict fracture sites. Model predictions were found to correlate with experimentally measured values and fracture sites (*71–74*).

2.2 Models of the Intervertebral Disc

Similar to vertebral simulations, one major reason for developing finite element models of the disc is to study the diseased state. Unlike vertebral modeling, which tends to focus on predicting fracture risk, intervertebral disc research has focused on understanding the process of degeneration and its biomechanical effects on the tissue (*54*). Some research groups have also investigated treatments for degenerative disc disease including replacement materials for the nucleus pulposus (*75*) and total disc replacement implants (*76*).

Generally the geometry of intervertebral disc finite element meshes are simplified. Many researchers, for instance, assume the structure to be axisymmetric (*11, 75, 77–81*) or exhibit symmetry in the sagittal (*80*) or transverse planes (*77*). In addition, cranial and caudal surfaces of the disc are often assumed to have a flat surface although some groups are incorporating more realistic curvatures into their models (*81*). Recently, increasingly complexity has been incorporated into material models of the intervertebral disc tissue, including ways to represent fluid content and flow, osmotic forces, and anisotropy (*77, 78*).

Validation of intervertebral disc model predictions often presents as a challenge due to the difficulties in taking in vivo measurements and preserving a realistic tissue environment in vitro. Single studies are often used for indirect validation, however, there tend to be large standard deviations in experimental data. It is often the case that reasonable agreement can be found but not quantified (*54*).

2.3 Segment Models

The motion segment, or functional spinal unit, is comprised of two adjacent vertebra, the interconnecting intervertebral disc, facet joint, and associated spinal ligaments. Since the spine may be considered a structure composed of several motion segments connected in series, motion segments are often used to study the biomechanical behavior of the spine. The functional spinal unit, along with larger spinal segments, can be used to simulate different spinal conditions and assist in the design and analysis of new spinal instrumentation, including pedicle screws, Harrington rods, and interbody cages.

In addition to modeling the vertebral and disc components, these models incorporate the facet joints and spinal ligaments and therefore present additional challenges beyond those seen in the previous two sections due to the limited experimental data available for characterization and the need to incorporate contact.

Due to limited experimental data on the behavior of the facet joint, researchers have developed a number of approaches to model facet interactions. Some groups have incorporated material properties through the use of gap elements that adjust the force transferred across the joint based on the gap size (*55, 82*), while others included and assigned properties to a cartilage layer (*53*). The articular facets surfaces are generally assumed to be frictionless (*53, 57, 83*) or a low coefficient of friction has been applied (*84*). In general, researchers have depended on indirect validation to provide evidence of their model accuracy. A few authors, however, have directly compared their model's results with tests on cadaveric specimens (*51, 52, 55, 59, 85*). All these studies, however, compare a generic model with experiments as opposed to a direct subject-specific segment model validation (*54*).

Advanced models are now beginning to incorporate poroelastic characteristics to simulate the fluid flow between the intervertebral disc and cancellous core. This complex interaction is believed to play a major role in absorbing impact energy and may be partially responsible for a perceived rate-dependency of spinal component response. In a poroelastic model, the disc is modeled as having two distinct phases: a fluid phase and a permeable solid phase. Simon and Wu were among the first to include poroelastic behavior in a finite element segment model (*86*). From short- and long-term creep analyses, they found that failure is likely to be initiated in the endplates or in the cancellous bone adjacent to the endplate. Wu and Chen created a similar model that incorporated posterior elements and found that under long-term creep loading their model demonstrated similar results as Simon and Wu (*87*). A poroelastic finite element model of the L_3-L_4 segment developed by Lee et al. (described in section 2.5) also supported Simon and Wu's results (*88*). In 2007, Natarajan et al. developed a poroelastic model that incorporated physiological parameters, such as the change in permeability in the disc due to strain, in order to predict the failure initiation and progression in a lumbar disc due to cyclic loads. The model accurately predicted change in disc height during loading and unloading that was comparable to in vivo experiments (*86*).

Many of the finite element models of the spine often assume a simplified geometry due to the complex nature of the vertebra and intervertebral disc. Facet joints and spinal ligaments may also be disregarded even though these structures have an important influence on motion segment behavior (*67*). In addition, trunk muscles are often not included in finite element models even though they are known to be primary stabilizers in the spine (*19*). Muscle forces have a strong influence on spinal load and stresses in the intervertebral discs and, in their absence, the applied stress distribution in the discs differs greatly (*85*). This elimination and simplification of geometry may influence which area will fail first under dynamic loading, altering the mechanism of failure. It is therefore essential that future efforts include the complex geometry of the vertebra and intervertebral discs, as well as the ligaments and muscles.

2.4 System Models

System level models of the spine include lumbar, thoracic, and cervical spine sections, as well as coupling to other body regions that have an influence on the loads transmitted to the spine, such as the torso. Developing a high-resolution finite element model of the entire human spine and its connected anatomy is computationally expensive and is difficult to validate. In the past, full models of the spine were referred to as simple models because they lacked a high degree of anatomic fidelity– many use just a few (usually one or three) elements to represent each motion segment. Many of these efforts were reviewed in the beginning of section 2. Now, with the advances in high performance computing, it is becoming easier to incorporate more accurate geometry and improved material constitutive models. However, from the published literature, it seems that there are limited high-resolution full spine models that capture the effects of high strain rate loading regimes. Nevertheless, there are a few research groups making significant progress achieving this goal, for example the Southwest Research Institute (*89*) and the Spine Research Center at Walter Reed Medical Center (*20*).

2.5 Failure Models

An adequate comprehension of fracture mechanisms is essential for effective prevention, mitigation and treatment. Predicting vertebral fractures, however, continues to be a challenge due to the complex and elaborate geometry of the vertebra and the different material properties of the vertebral components. While a number of researchers have studied the instability, spinal canal impingement, and treatment methods of vertebral fractures produced by spinal trauma, there is still a lack of knowledge on the internal transient change of the vertebral bodies during fracture (*19*). Lee et al. used a three-dimensional, poroelastic finite element model of the L_3-L_4 segment to predict changes in biomechanical parameters such as dynamic stiffness, intradiscal pressure, stress in the endplates, and the shock-absorbing mechanism of the spine under different impact durations and loading rates (*88*). Additionally, the authors analyzed the relation between the loading rate and the potential of vertebral body fracture. It was found that short-duration loading has a strong effect of initiating fractures in the endplates or the posterior wall of the cortical shell. The authors also concluded that pressure in the nucleus is independent of the impact duration and depends on the magnitude of the impact force.

To date, very little research has been performed to analyze the mechanism of fracture, especially on a predictive basis. Qiu et al. developed a finite element model of the thoracolumbar junction and subjected it to dynamic vertical impacts in order to investigate the vertebral burst fracture process (*19*). The authors found that an increase in pressure inside the nucleus forces each endplate to bulge towards the cancellous core of the vertebral body. When the endplates fail, the nucleus material is able to enter the vertebral body, pressurizing it more. This increase in pressure squeezes the contents of the vertebral body, including fat and marrow, out of the cancellous bone. When the nucleus material enters the vertebral body faster than the rate at which fat and marrow are being expulsed, a burst fracture is said to occur. Fracture sites predicted in this study were consistent with other studies as well as identical to the medical definition of burst fractures (*19*). Teo et al., developed a finite element model of the human atlas (C1) and were also able to accurately predict failure sites in comparison to experimental results (*90*).

In 2009, El-Rich et al. developed a finite element model of the L_2-L_3 spinal segment that analyzed the response of the bony structures and soft tissues to five degrees of sagittal rotation at different rates until fracture occurred. The model allowed for a detailed evaluation of failure occurrence and propagation over the bone. Results showed that the endplate, pedicle, and facet surface were the weakest regions when subjected to rapid movement in the sagittal plane and spinal injuries resulted from a sagittal rotational velocity exceeding 0.5°/ms (*56*). Segments subjected to the highest velocity resulted in a more severely damaged structural response compared to those subjected to the low and intermediate velocities, confirming that rapid movements increase the risk of injury (*91*). Under the highest velocity the model exhibited a considerable increase in intradiscal pressure in the nucleus and the stresses over the rest of the structures while there was only slight changes under lower and intermediate loading rates.

The spinal cord may be injured through various injury patterns; however, the relationship between spinal column injury pattern and spinal cord damage is not well understood. Greaves et al. developed a finite element model of the cervical spine and spinal cord in order to compare spinal cord distributions between three different injury mechanisms (*62*). Their model predictions showed that strain that induces distraction injuries were uniformly distributed lengthwise throughout the spinal cord, while dislocation injuries and contusion injuries (such as a burst fracture) led to focal strains.

A biomechanical description of the injury process can help predict subsequent acute and chronic pathophysiology of spinal cord trauma. Li et al. developed a finite element model of the cervical spinal cord to simulate the complex injury mechanism of a hyperextension injury (*92*). The results of their hyperextension injury simulation showed high localized stress at the anterior and posterior horn in the gray matter of the spinal cord. It is believed that these stresses account for the predominance of hand weakness in patients with central cord injury (*92*).

2.6 Uncertainty Quantification

Probabilistic analysis methods provide a tool in which the biomechanical response due to characteristic uncertainties and variations in biological structures can be studied. The uncertainty of these values lead to an associated degree of uncertainty in the results (*93*). In order to improve the precision of computer models, probabilistic models, in which uncertainties and natural variations are incorporated into the model, have been developed. The fundamental concept of this method is that the input parameters are defined by an appropriate statistical distribution, not a single value (*93, 94*). A Normal or Gaussian distribution is commonly used in probabilistic models and defines the input variables by a mean and standard deviation. The values of each input parameter are sampled at random from the distribution and used in the model. The model can then be solved multiple times, allowing researchers to develop an indication of the range of the results that can be found.

To date, this methodology has been used to study the influence of variations in material properties and geometry on spinal behavior (*95–97*). In 2005, the U.S. Naval Bio-dynamics Laboratory/U.S. Naval Air Warfare Center Aircraft Division used NESSUS, a computer software system that combines finite element methods with advanced reliability methods in order to model uncertainties in geometries, material properties, and other variables, to evaluate the risk of cervical spine injury from maneuvers experienced during flight, high speed ejection, and crashes (*97*). The group incorporated mechanical and geometric data in order to develop, verify, and validate a parameterized probabilistic spinal injury prediction model. In order to simulate heterogeneous and nonlinear materials the group defined the mechanical characteristics of vertebrae, discs, ligaments and muscles through data collected from QCT, bone mineral density data, and cryomicrotomy (*98*).

In addition, probabilistic models have been developed to study the impact spinal procedures, such as vertebroplasty and the implantation of spinal stabilization devices including artificial discs and pedicle screws, have on spinal functions (*94, 99, 100*). Progress in probabilistic biomechanics will depend upon the identification of injury modes, data collection, and the development of validated models (*39*). Additional research is necessary to integrate random loading schemes and random representations of geometric variations. As Thacker et al. (*101*) point out, uncertainty and error quantification play a key role in model verification and validation. Uncertainties associated with the model input parameters include material behavior, geometry, loads, initial conditions, and boundary conditions. Ideally, each input parameter would be varied according to statistically significant data obtained from experiments.

3. Conclusion and Recommendations

Developing an understanding of the high rate injury biomechanics associated with blast-induced loading to the spine, lower extremities, central nervous system, and peripheral nervous system is critical to optimize protective equipment for Soldiers. Finite element analysis is an essential part of understanding the biomechanics of spinal injury. These models provide a wealth of information that otherwise may not be available. While the research area of computational spinal injury biomechanics has received much attention in the past with regard to low strain rate degenerative effects, there are significant opportunities to enhance the state of knowledge of spinal injury in the area of high strain rate military loading. Through this literature review and careful contemplation, a list of six recommendations for future research directions has been developed. Future research concerning the finite element model of the spine should focus on the following aspects:

- **Develop high-resolution anatomic finite element models that capture the geometric complexity of the spine in a probabilistic manner.** Models should be three-dimensional and include all anatomic components of the spine necessary to capture the physics of interest, namely, injury mechanisms such as fracture and ligament rupture. Models should be constructed directly from high-resolution medical data such as x-rays, CT scans and MRI that enable individual-to-individual variations of the geometric representation. Finite element studies should use a probabilistic approach to include the variation of geometry ranges that are typical during postural changes, movement, gravity, aging, and extrinsic loading.

- **Develop a multilevel approach to apply loading conditions to the spine.** Currently the direct loads, their distributions, and strain rate that the spine experiences during a military underbelly blast event is unknown. There is detailed understanding about the explosion and the loading to the vehicle, as well as the spinal injury outcome observed from the field; however, little is known about the processes in between these endpoints. It is critical to understand the process of load transfer, injury initiation, and propagation. In addition to the complexity of considering the spine itself, it is suspected that the lower extremity reactions to underbelly blast and spinal injuries are coupled and cannot be considered isolated systems. Also, as discussed in section 2.4, a coupling between the spine and the torso exists. Thus, the coupled systems direct the research to face the challenging task of considering a wide range of length and times scales. While on one hand research must consider the large length scale (on the order of meters) for coupled systems that include the lower extremities, spine, and torso, we must also consider the smaller length scales (on the order of micrometers), such as the randomly organized collagen fibers and radially organized elastin fibers found in the nucleus pulposus, to understand the injury mechanisms. The military spinal injury research community would benefit from a set of high strain rate military loading paradigms for components (ligaments, processes), subassemblies (isolated vertebral bodies and intervertebral discs), assemblies (motion segments consisting of vertebral bodies, interconnecting ligaments, and intervertebral discs), and system level (whole spine) models. Loading paradigms should include rate, direction, and duration. Once again, finite element studies should use a probabilistic approach to include the variation of loading that are typical.

- **Develop, implement, and validate constitutive models for individual spinal component materials at the appropriate strain rate conditions.** As table 1 demonstrates, an abundance of literature sources describe a wide range of material properties of spinal structures. This variation in material property values has resulted in a substantial deviation of model predictions and results. Furthermore, results currently obtained from finite element models are often limited by the assumption of homogeneous, isotropic, and linear elastic material properties. While these assumptions simplify the analysis, they do not capture the realistic mechanical behavior of the spine (*19, 84, 85, 92*). Ultimately, models need to incorporate the anisotropic, nonlinear, heterogeneous, and porous material properties seen in the human spine. For example, we believe that a poroelastic model is essential to characterize changes of the annulus matrix from a solid to fluid phase during high-rate loading conditions. Recently, it has been shown that standard high strain rate material characterization techniques are not adequate for bones or soft tissues (*102, 103*). Additionally, standard gripping techniques for dynamic tension tests do not work for biological materials, and material inhomogeneity and anisotropy complicate the interpretation of measured stress and strain data. New dynamic test methods and data analysis techniques are needed to provide applicable high rate and failure data on spinal materials for constitutive model development and calibration.

- **Hierarchical verifications and validation.** Computational models and results should be compared at multiple length scales. Thacker et al. (*101*) provide a useful hierarchical verification and validation scheme that decomposes the spine into components (ligaments, processes), subassemblies (isolated vertebral bodies and intervertebral discs), assemblies (motion segments consisting of vertebral bodies, interconnecting ligaments and intervertebral discs), and system level (whole spine) models. The process should start at smaller length scales such as the components and progress to the last stage of system level models. This approach requires simulations and experiments to be conducted at each level.

- **Use novel computational approaches for capturing high strain rate compressive failure.** There are limited computational approaches to capture injury mechanisms associated with accelerative loading to the human spine. Current U.S. Army Research Laboratory (ARL) efforts are focused on using dynamic insertion of rate-independent cohesive elements to capture fracture mechanisms. The limitations of this approach are quickly being realized due to the large fragmentation, finite element mesh dependency, and material interface compatibility experienced during high rate accelerative loading. In addition, computational approaches to model muscle rupture, tearing, and other failure mechanisms are neither common nor validated within the tools used today. One may think of cohesive elements as a traditional approach to modeling failure in the computational setting. However, given the large deformations that biological materials exhibit during accelerative loading and the complexity of the anatomic geometry, cohesive approaches will only work within a limited regime of deformation. Therefore, novel simulation capabilities should be pursued. The Material Point Method (MPM) and the Extended Finite Element Method (XFEM) are two methods that seem feasible.

- **Integrate functional outcome predictions coupled with mechanical damage.** It is also important that computational models extend past mechanical descriptions to include functional measures. A multidisciplinary approach combining mechanics-based simulation and modeling with the clinical field to develop computational approaches for modeling structural and functional deficits that may arise from high strain rate spinal injuries is needed. The community should develop time-evolving predictive models to understand musculoskeletal stability, pain, and rehabilitation strategies.

At this time, the focus of numerous research groups is to develop a three-dimensional model under a full range of loading schemes, while simultaneously considering nonlinear material and geometry solutions. Current and past models are limited by their selection of material models and simplified geometries that are unrealistic in comparison to the human spine. With a full understanding of spinal injury mechanisms, design criteria based on human injury, such as vertebral body fracture tolerances, can be developed and implemented in order to develop spinal injury mitigation strategies for the warfighter.

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Appendix A. Verification, Sensitivity and Validation

Models must establish credibility before clinicians and scientists can use the simulations to extrapolate information and make decisions from information and results produced by the model. Specifically, an analyst must (1) assess the numerical accuracy of the underlying mathematical equations in the computation model, (2) evaluate the sensitivity of the model results to the input parameters, and (3) validate that the model results correspond to results of a real-world scenario (*54, 104*). Models are often validated through one of two methods: direct or indirect validation. Direct validation involves a comparison between a model's prediction and an experimental test result. Indirect validation compares model results to a physical case where it is not possible to determine if conditions are the same. This includes comparisons between data from the literature, clinical trial results, and historical medical data (*54*).

Since the first spinal finite element models were developed, significant advances have been made. Through adequate considerations of model sensitivity and validation, researchers have been able to develop more accurate and efficient simulations. Generating a precise model that incorporates verification, analysis sensitivity, and validation is a difficult task, especially for the complex spinal structure. However by doing so, the gaps between clinical medicine, experimental biology, and computational biomechanics will decrease.

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