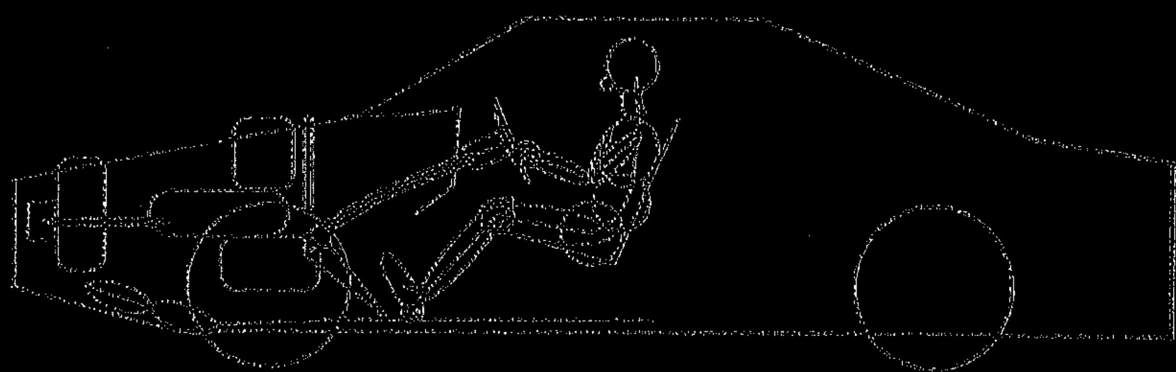
SECOND EDITION

Accidental Injury

BIOMECHANICS AND PREVENTION



Alan M. Nahum
John W. Melvin
EDITORS

Alan M. Nahum, M.D.
Medical-Legal Consultants
6361 Nancy Ridge Drive
San Diego, CA 92121
and
University of California at San Diego
School of Medicine
La Jolla, CA 92093, USA

John W. Melvin, Ph.D.
Tandelta, Inc.
Ann Arbor, MI 48103, USA

Library of Congress Cataloging-in-Publication Data
Accidental injury : biomechanics and prevention / edited by Alan M. Nahum, John W. Melvin – 2nd ed.
p. ; cm.
Includes bibliographical references and index.
ISBN 0-387-98820-3 (hard cover : alk. paper)
1. Crash injuries—Psychological aspects. 2. Wounds and injuries. 3. Human mechanics. I. Nahum, Alan M. II. Melvin, John.
[DNLM: 1. Wounds and Injuries—prevention & control. 2. Biomechanics, WO 700 A1714 2000]
RD96.6 .A23 2000
617.1'028—dc21 00-024950

Printed on acid-free paper,

Production managed by Terry Kornak; manufacturing supervised by Jerome Basma.
Typeset by Best-set Typesetter Ltd., Hong Kong
Printed and bound by Maple-Vail Book Manufacturing Group, York, PA.
Printed in the United States of America.

9 8 7 6 5 4 3 2 1

ISBN 0-387-98820-3 SPIN 10717154

Springer-Verlag New York Berlin Heidelberg
A member of BertelsmannSpringer Science+Business Media GmbH

18
Injury to the Thoracolumbar Spine and Pelvis

Albert I. King

Injury to the bony portion of the thoracolumbar is rare in automotive collisions. Soft tissue injuries appear to be more common. This chapter describes the major modes of injury to the spine, and discusses the biomechanical response of the spine to vertical ($+g_z$) and horizontal ($-g_x$) acceleration. A form of spinal injury due to the wearing of shoulder belts is discussed. The biomechanics and neurophysiology of low back pain form the foundation for an understanding of soft tissue injury. The relationship between disc rupture and impact loading is considered to be remote, as disc rupture is a degenerative process that occurs over a long period of time. Mathematical models of the spine are also reviewed.

The pelvis is a bony structure that transmits the weight of the torso to the lower extremities during normal locomotion and supports the torso in the seated position. In an automotive impact environment, it can sustain injury from both frontal and side impact, and, during aircraft ejection or vertical falls, it is called upon to take the entire inertial load from seat-to-head acceleration. Injuries to the pelvis, however, contribute only about 1% to the total Injury Priority Rating (IPR). This structure is important in this discussion, therefore, primarily for its response during load transmission.

The Spine

Functions

The human vertebral column is the principal load-bearing structure of the head and torso. There are also secondary functions performed by each portion of the spinal column. The cervical spine provides the head with a limited degree of mobility and a protected pathway for the proximal segment of the spinal cord. The thoracic spine offers the same protection to the cord, while it offers mobility to the upper torso and rib cage. The lumbar segment provides the lower torso mobility and encloses the distal end of the spinal cord. The protective role of the vertebral column is analogous to the function served by the skull to protect the brain. However, anatomic requirements dictate that the spine be flexible and yet strong so that it can serve a multitude of functions. Like the skull, it is strong but not strong enough to withstand mechanical insults of modern-day transportation systems. Injuries that affect the function of the spinal cord can result in death, quadriplegia, or paraplegia. Those who survive suffer permanent disabilities that cannot be restored as yet by modern medicine. Other biomechanical motivations to study the mechanical response of the spine include neckache and backache, osteoporosis, and scoliosis.

This chapter deals in part with the biomechanics of the spine, with particular emphasis on injury mechanisms and mechanical response to impact acceleration. Although spinal injuries are relatively uncommon in automotive accidents, they can often be rather severe and disabling. They are more common in aircraft accidents and constitute a special problem in aircraft ejection, which is the cause of anterior wedge fractures of the thoracolumbar spine.

In a review of 1988 National Automotive Sampling System (NASS) data on thoracolumbar spinal injuries, it was found that the frequency of injury was about 2% if vertebral fracture and back muscle strain were included, and 0.3% if only vertebral fractures were included. It is interesting to note that for an Abbreviated Injury Scale (AIS) range of 3 to 6, the frequency of spinal injury was 2.1% for lap-shoulder belted occupants, 1.5% for all cases, and 1.3% for unrestrained occupants.

Anatomy of the Thoracolumbar Spine

Familiarity with the anatomy of the vertebral column is necessary for understanding the biomechanics of the spine and its response to load. The ability to model this response also calls for an appreciation of the function of the various components of the column. From a macroscopic point of view, the vertebral column is made up of 24 individual bones, called vertebrae, that are joined together by several different types of soft tissue. The primary types of soft tissue are the intervertebral discs, ligaments, and skeletal muscle. As shown in Figure 18.1, the seven vertebrae supporting the head constitute the cervical spine, while the 12 vertebrae below it form the thoracic spine. The lumbar spine is the most inferior segment and is made up of five vertebrae. The entire column is supported by the sacrum, which is anatomically a part of the pelvic girdle. The thoracolumbar spine is located along the midline of the posterior aspect of the torso, and the cervical spine is along the posterior aspect of the neck. In general, each vertebra consists of a body, neural arch or pedicles, laminae, facet joints, spinous process, and transverse processes. The body is a cylindrically shaped bone consisting of a core of spongy bone surrounded by a thin layer of cortical or compact bone. The endplates above and below the centrum are cartilaginous. The sides of the body are usually slightly concave and form a narrow waist at midlevel. Figure 18.2 shows a typical lumbar vertebra, viewed laterally and posteri-

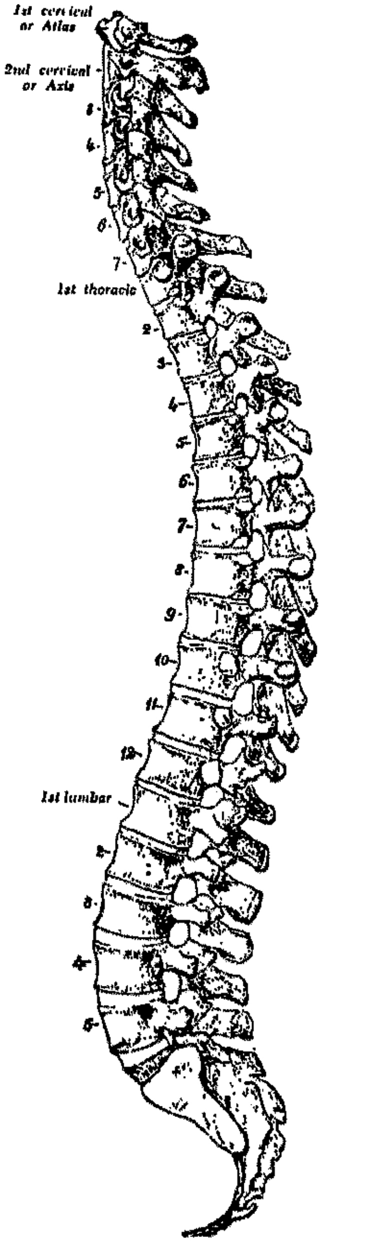


FIGURE 18.1. Lateral view of the spine. (From Gray, 1973.)

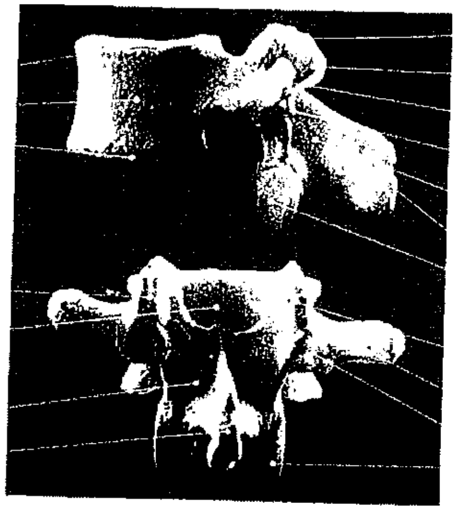


FIGURE 18.2. Lateral and posterior view of a lumbar vertebra.

orly. The pedicles arise from the posterolateral aspects of the body and are directed rearward. They form the lateral aspects of the spinal canal that surrounds the spinal cord and affords it mechanical protection. The laminae are quadrilaterally shaped pieces of compact bone that form the posterior aspect of the spinal canal. At the junction between the pedicles and the laminae are the articular facts. Each vertebra has four facets, two superior and two inferior. The facets are bony projections that articulate with mating projections (facets) of the vertebrae above and below. The joints formed by the facets are true synovial joints, encapsulated by capsular ligaments. The orientation of the facet joint surfaces varies from vertebra to vertebra and is of biomechanical interest because the facets share the load-bearing function of the spine with the vertebral bodies. The geometry of the facets will be described below.

Continuing with the general description of a typical vertebra, the transverse and spinous processes complete the posterior structure. They act as attachment points for muscles and ligaments and can be considered as short cantilever beams with free ends. The vertebrae gradually increase in size caudally, roughly in proportion to the weight they are expected to support. The precise description of each vertebra can be found in a text on human anatomy. The lateral view of the entire column in Figure 18.1 shows three principal spinal curves: the lordotic cervical and lumbar curves and the kyphotic thoracic curve. The normal spine is straight when viewed frontally. Abnormal lateral curves found in scoliotic spines tend to develop in adolescence and are more common in females than males. Mechanical explanations for this form of instability are not completely satisfactory. It should also be noted that the thoracic spine supports the posterior section of the rib cage. A pair of ribs arise from each thoracic vertebra. These ribs articulate with the vertebrae near the junction of the pedicles with the vertebral bodies and at the tips of transverse processes.

The spine has a few special features. In particular, the orientation of the facets is of biomechanical significance. To describe the orientation of the facet joint surfaces, it is convenient to use a unit to establish the approximate orientation of these surfaces. Some of the surfaces are slightly curved, and the description of their orientation assumes the unit normal to be located at the center of the surface. The unit normal for thoracic superior facets is directed generally posteriorly with a variable lateral component of about 30 degrees and an upward tilt of about 20 degrees. The lumbar surfaces are slightly curved, but the unit normal at the center is directed medially. Its orientation tends to shift to a posteromedial direction for the lower lumbar vertebrae. However, the vector tends to lie in a horizontal plane. A pictorial description of the orientation of facet surfaces can be found in Gray (1973). Another special feature of note is the inclination of the fifth lumbar vertebra (L5). The endplates are inclined at the L4-5 and L5-S1 level due to the lordotic curvature of the lumbar spine. The forward inclination of L5 can be more than 30 degrees in some individuals, resulting in high shear loads at these lower lumbar joints.

The vertebrae are joined together by soft tissue, anteriorly by ligaments and intervertebral discs, and posteriorly by ligaments and facet joint capsules. Intervertebral discs are cartilaginous in origin and consist principally of collagen, proteoglycans, and water. The disc can be divided into two main regions: the nucleus pulposus and the annulus fibrosus. The latter is a ring of primarily type I collagen (the type found in skin, tendon, and bone), made up of dense layers of collagen fibers that have an intricacy of pattern that almost defies description. In general, the direction of the fibers in adjacent layers cross each other at an oblique angle, but the direction of the fibers in any given layer can also change or the fibers can bifurcate and assume more than one direction. In the lumbar region, 12 to 16 layers can be found anteriorly. Type II collagen (the type found in hyaline cartilage) can be found in the nucleus, which has a higher concentration of proteoglycans, giving it a gel-like character. Proteoglycans have an affinity for water and are responsible for the maintenance of tension in the annular collagen fibers. The anatomy and function of the disc are affected by age. Disc degeneration begins at a very young age, and normal healthy discs are the exception rather than the rule in spines over the age of 25. The number and size of collagen fibrils increase with age, and the distinguishing features of the nucleus disappear as age transforms the entire disc into fibrocartilage. A detailed description of the anatomy of the disc can be found in Peacock (1952) and Buckwalter (1982).

A new finding regarding the microstructure of annular layers has been reported by Marchand and Ahmed (1990). They confirmed the fact that the fiber orientation within a single layer can indeed vary and that the change in orientation occurred at cleavage lines, which are possibly sites of mechanical weakness where a disc herniation can occur.

The articular facets are enclosed by a joint capsule and appear to allow the spine to flex freely while acting as motion limiters in spinal extension or rearward bending. Cavanaugh et al (1996) reported that these capsules can undergo a large amount of stretch, particularly when the lumbar spine is placed in extension. There is also neurophysiological evidence provided by Yamashita et al (1990) of pain-sensing fibers in the capsule that, if stimulated sufficiently, can be set off, resulting in low back pain. The joint surfaces are lined with articular cartilage and are lubricated by synovial fluid.

There are three spinal ligaments that run along the entire length of the spine. They are the anterior and posterior spinal ligaments, which line the anterior and posterior aspects of the vertebral bodies, and the supraspinous ligament, which joins the tips of the spinous processes. The ligamentum flavum, or yellow ligament, is a strong band that connects adjacent laminae behind the spinal cord. The interspinous ligament is a thin membrane located between adjacent spinous processes.

The spine is maintained in an erect posture with the help of the skeletal musculature. The extensor muscles of the thoracolumbar spine can be divided into two main groups: the superficial transversocostal and splenius group and the deeper transversospinal group. The former group contains muscles that arise from the pelvic region and insert at various levels from the 6th to the 12th rib. Others arise from the lower ribs and insert at the upper ribs or along the cervical spine. The deeper group contains muscles that join one vertebra to another or span one or more vertebrae. The principal flexors of the thoracolumbar spine are the internal oblique muscles and the rectus abdominus.

Injury Mechanisms

Injuries to the vertebral column can be roughly classified into seven different categories:

1. Anterior wedge fractures of vertebral bodies
2. Burst fractures of vertebral bodies
3. Dislocations and fracture-dislocations
4. Rotational injuries
5. Chance fractures
6. Hyperextension injuries
7. Soft tissue injuries

Anterior Wedge Fractures

These injuries occur at all levels of the spine and are common in both aircraft and automotive accidents. The mechanism of injury is combined flexion and axial compression. It is a mild form of spinal injury commonly identified with the pilot ejection problem. The region most susceptible to anterior wedge fractures during ejection is between T10 and L2, although they can occur in the upper thoracic region as well (T4–T6). Kazarian (1982) postulated that the mechanism of injury to the T4–T6 segment is forcible exaggeration of the normal upper spinal curvature. The fact that very little vertical ($+g_z$) acceleration is experienced in an automotive crash does not mean that wedge fractures cannot occur. Begeman et al (1973) have shown that subjects restrained by a lap belt and an upper torso belt, in a $-g_x$ environment, develop high spinal loads that can cause wedge fractures similar to ejection seat injuries.

Thoracic wedge fractures are seen in Scheuermann's disease, a condition not associated with an acute injury. Multiple levels are usually involved with mild wedging and disc space narrowing. A distinguishing feature is endplate irregularity, which is not present in acute fractures (Brandser and El-Khoury, 1997).

Burst Fractures

These injuries are due to higher levels of input acceleration or applied load, applied more directly over the vertebral body, causing it to break up into two or more segments. The integrity of the cord is threatened by the movement of the segments posteriorly into the spinal canal. The cord can also be injured by the retropulsion of the disc into the canal, particularly in the cervical spine. It should be noted that in many cases of paralysis, postimpact x-rays show a burst fracture with fragments that do not intrude into the spinal canal. This does not mean that spinal cord was not injured or contused by these fragments because such x-rays do not reveal the full extent of their dynamic retropulsion and because there is some retraction of the fragments after the impact. Oxland (1992) performed impact tests on 16 specimens of the thoracolumbar spine and was successful in creating 12 burst fractures. He used a weight drop device and obtained burst fractures at loads ranging from 3 to 8kN. The average was 6kN. He also measured encroachment of the canal by fragments of the vertebral body using a special strain gage device. The encroachment ranged from 2.4 to 16mm with an average of 7.5mm.

In severe cases of burst fracture, the dura of the spinal cord can be torn (Brandser and El-Khoury, 1997). This is a confirmation of the intrusion of vertebral body fragments into the spinal canal and an indication of the speed of these fragments.

Dislocations and Fracture-Dislocations

These are generally flexion injuries accompanied by rotation and posteroanterior shear. Unilateral dislocations require an axial rotational component, while bilateral dislocations can be due solely to flexion and compression. The essential difference between a simple wedge fracture and a fracture-dislocation is, according to Nicoll (1949), the rupture of the interspinous ligament. This observation is biomechanically significant and will be discussed later. There are varying degrees of dislocation. The inferior facets can be simply moved upward relative to the superior facets of the vertebra below or the facets can be perched on top of each other. There can also be a forward dislocation with fracture of the facets or the neural arch, and forward dislocation with locking of the facets. That is, the inferior facets have moved up and over the superior facets of the vertebra below and come back down so that they are now anterior to the superior facets. There is a high probability of neurologic damage in this type of injury because the cord is subjected to high shearing and stretching forces. If there is dislocation without wedging, the mechanism of injury is a high shear load in the posteroanterior direction (Kazarian, 1982).

Rotational Injuries

If the spine is twisted about its longitudinal axis and is subjected to axial and/or shearing loads, lateral wedge fractures can occur (Nicoll, 1949). Other forms of injury include uniform compression of the vertebral body and fracture of the articular facets and lamina. Kazarian (1982) indicated that lateral wedge fractures seem to gravitate to two spinal regions: T2 to T6 and T7 to T10. The damage to the posterior intervertebral joint is on the concave side, and this injury is often accompanied by fracture of the transverse process on the convex side. Unlike the anterior wedge fracture, this injury may result in neurologic deficit, including paraplegia.

Chance Fractures

This injury was first described by Chance (1948) as being a lapbelt–related syndrome in which a lumbar vertebra is split in the transverse plane, beginning with the spinous process. Subsequent studies, for example, by Smith and Kaufer (1967), attribute the injury to the improper wearing of the lap belt while involved in a frontal ($-g_x$) collision. An improperly worn lapbelt or a belt with too shallow an angle can ride over the iliac wings and acts as a fulcrum for the lumbar spine to flex over it, causing a marked separation of the posterior elements without any evidence of wedging (Steckler et al 1969). When the lapbelt is used in conjunction with an upper torso restraint, this injury does not occur.

Hyperextension Injuries

Hyperextension injuries of the cervical spine result in avulsion of the anterior aspect of the vertebral bodies, sometimes termed "teardrop fractures." Kazarian et al (1979) reported the occurrence of hyperextension injuries of the thoracic spine, resulting from ejection from F/FB-111 aircraft. The superior lip of one or more vertebrae is avulsed along with the rupture of the anterior longitudinal ligament. This injury is sometimes accompanied by loss of posterior vertebral body height. When this occurs, there may be injury to the articular facets, pedicles, and/or the laminae. The incidence was 23% over a 10-year period. The powered inertial reel and the seat back were considered responsible for this rare injury because of the large forces exerted on the front of the torso when the belts pulled the shoulders back.

Soft Tissue Injuries

The soft tissues involved are the intervertebral disc, the various ligaments around the intervertebral joint, the facet joints and their capsules, and the muscles and tendons attached to the vertebral column. The usual complaint of this type of injury is low back pain, which is often associated with radiating pain down the buttocks and the lower extremities. The incident provoking this complaint can vary from a mild bump in the rear by another vehicle while the victim is stopped at a light, to a bus going over a pothole, to a relatively severe collision of two cars at an intersection. If the x-rays taken in the emergency room are negative, a diagnosis of lumbar sprain or strain is made and the patient is sent home with some painkillers. In some cases, the pain persists and eventually a diagnosis of disc rupture, disc bulge, or other specific syndrome is made and the incident in question is generally blamed as the cause of the injury. This cause-and-effect relationship is invariably based on the history provided by the patient and not on the severity of the impact or the biomechanics of the loading on the spine.

Biomechanical Response of the Thoracolumbar Spine

Because of its flexibility, the vertebral column is frequently subjected to bending loads that are superimposed upon the axial load it bears to support the head and torso. There is no question that impact accelerations in the horizontal plane exert bending loads on the spine. However, vertical ($+g_z$) acceleration is also capable of subjecting the spine to a high level of bending due to the fact that the vertebral column is located along the posterior aspect of the torso.

It is perhaps interesting to trace the progress made in experimental research on spinal injury, beginning with this bending hypothesis made by King et al (1968). The development of countermeasures to prevent anterior wedge fractures from occurring in pilots who eject from disabled aircraft was somewhat hampered by simple spinal models of Latham (1957) and Hess and Lombard (1958). While they are admirable modeling efforts for their time and are sound from an engineering viewpoint, they unfortunately led subsequent researchers away from looking at the anatomy of the spine. The models were capable of simulating axial loading only. Experimental studies on the spine during whole-body acceleration of cadavers in the $+g_Z$ impact acceleration mode revealed that the spine was subjected to high bending loads even though it was restrained by a shoulder harness, and the input acceleration was in the seat-to-head (vertical) direction. This led to a more detailed study of the load-carrying capacity of the spine during $+g_Z$ acceleration. Ewing et al (1972) tested a series of embalmed cadavers on the Wayne State University vertical accelerator, using three different restraint configurations—the hyperextended, erect, and flexed modes. In the hyperextended mode the spine was pulled back at the shoulders by a pair of military-type harnesses, while the thoracolumbar spine was placed in extension by inserting a block of wood 50 mm thick behind the spine at the L1 level. In the erect mode, the spine was in its natural configuration while seated in a rigid seat, with the shoulder belts tightened manually to a tension of approximately 300 N. The shoulder harness was loosened in the flexed mode, permitting the torso to flex forward freely. The objective of the study was to determine the fracture level of the spine as a function of its spinal configuration. The results are shown in Table 18.1. By hyperextending the spine, the fracture g-level increased some 80%, and the observed difference was significant at the 95% level. In a subsequent search for this dramatic increase in spinal strength, it was determined that the spine did not receive external support from the hyperextension block and that the reason was an internal redistribution of the load borne by the spine.

TABLE 18.1. Increase in g-level to fracture due to hyperextension of the lumbar spine.

Spinal configuration	No. of specimens	Average fracture g-level (g)
Hyperextended	4	17.6
Erect	5	10.4
Flexed	3	9.0

Prasad et al (1974) embarked on a study to prove the hypothesis that the spine had two load paths and that the articular facets were indeed capable of transmitting load from one vertebra to the next. This facet load was difficult to measure directly but could be computed indirectly if the load borne by the disc was determined. An intervertebral load cell (IVLC) was designed to replace the inferior portion of a lumbar vertebra that was cut out by means of a double-bladed rotary saw. The IVLC shown in Figure 18.3 is 10 mm thick and has a diameter of about 40 mm. It is capable of measuring axial compression and the eccentricity of that load in the midsagittal plane. Figure 18.4 shows the facet load computed by subtracting the intervertebral load from the total load borne by the spine. The latter was assumed to be proportional to the measured seat pan load, with the proportionality constant equal to the ratio of the weight of the torso above the IVLC to the total weight of the body. At the beginning of the acceleration pulse, the facets were in compression, sharing the inertial load with the vertebral body and disc. As the head and torso flexed forward, the facets went into tension. These results were confirmed by Hakim and King (1976), who reproduced the IVLC loads on excised spinal segments in an *MTS* materials testing machine. By hyperextending the spine, the facets were prevented from going into tension, thus increasing the fracture level of the most vulnerable vertebral bodies in the

FIGURE 18.3. Photograph of an intervertebral load cell (IVLC) (thickness = 10mm).

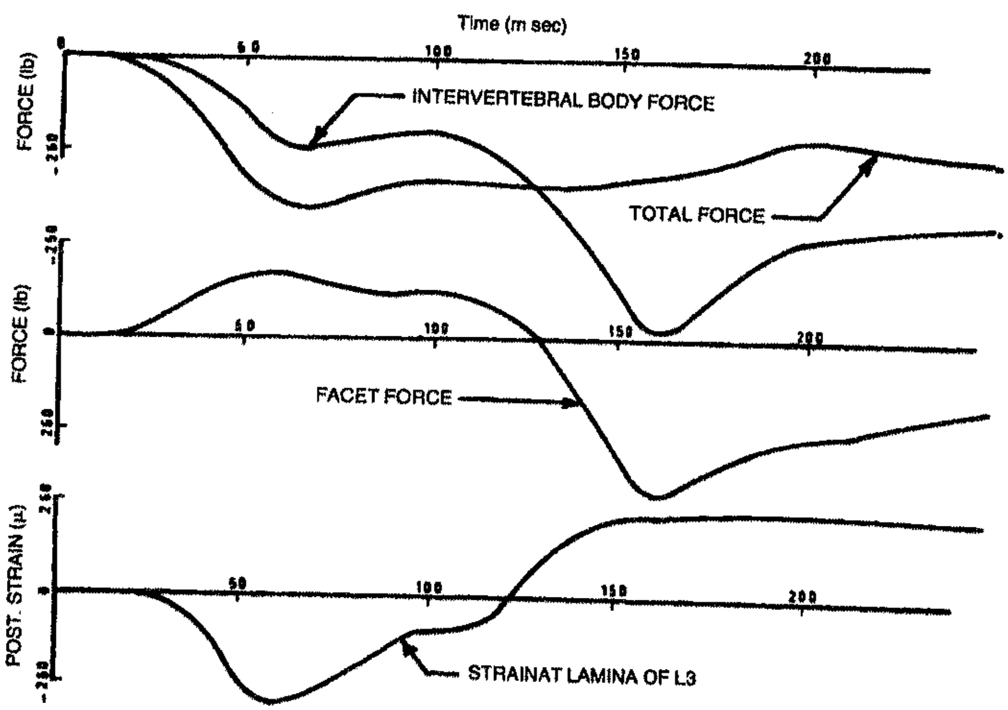


FIGURE 18.4. Facet load computed from measured IVLC data. (From Prasad et al, 1974.)

thoracolumbar spine. Furthermore, the facet load hypothesis provided an explanation for the frequently observed anterior wedge fractures. The additional compression load borne by the bodies was needed to balance the flexion moment caused by forward rotation of the head and torso. Since the moment arm is of the order of 25 mm and the flexion moment can be as high as 40 N.m, this additional compression is over 1,000 N. Such excessive compressive loads are the cause for anterior wedging of the vertebral bodies. Injury data from subhuman primates, obtained by Kazarian et al (1971), indicate that derangement of facets was due to locking of the facet joints to act as load paths during $+g_z$ acceleration. Although injuries to the posterior elements are rare in pilot ejection, these observations corroborate the load-bearing hypothesis of the facets.

Patwardhan et al (1982) measured contact pressure between the articular surfaces of lumbar facets and computed a facet force, reporting it to be the vertical facet force that was measured indirectly by Prasad et al (1974). This was felt to be erroneous since the articular surfaces are quite incapable of transmitting large shear loads. Yang and King (1984) performed loading experiments on isolated facet joints and obtained results that can explain the mechanism of load transmission through the facet joint. The posterior elements were separated from the body by cutting through the pedicles. The two facets were then loaded axially in their normal configuration in an Instron testing machine. In compression, they acted as a stiffening spring, as shown in Figure 18.5. In tension, however, they afforded very little resistance. Most of the tensile resistance was provided by the ligamentum flavum and the interspinous and supraspinous ligaments. Figure 18.6 shows the tensile load deflection curve of the isolated facets, with all ligaments severed. The mechanism of load transmission in compression is thus different from that in tension, and it is postulated that high compressive loads can be generated in the facet joint when the inferior tip of the inferior facet bottoms out on the pars interarticularis of the vertebra below it. In tension, the resistance is provided by soft tissues, such as the ligaments and the extensor muscles of the back. Figure 18.7 is a photograph of the lamina of someone who was possibly a weight lifter, showing graphic evidence of facet loading due to a life-

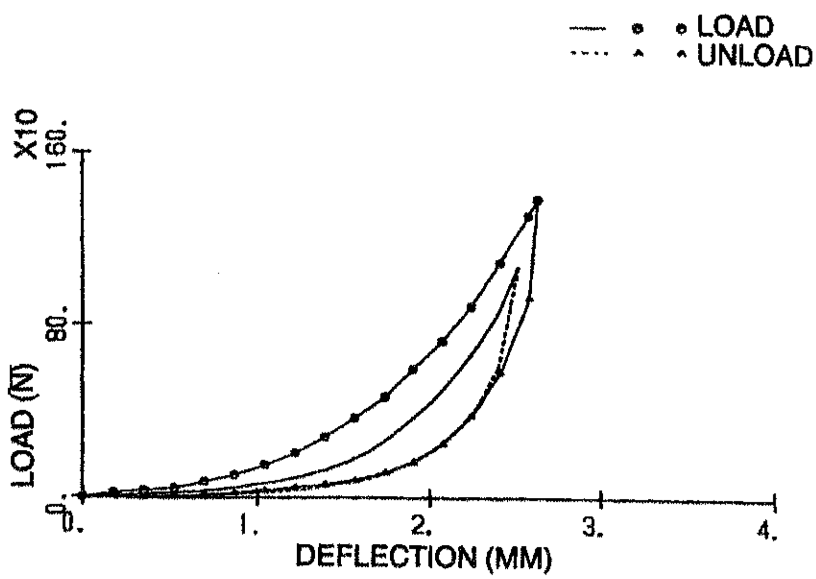


FIGURE 18.5. Response of an isolated facet joint to a compressive load. (From Yang and King, 1984.)

time of heavy weight bearing. It should be noted that the axial loading experiment on isolated facet joints was carried out because very little rotation of the facet joint could be detected when complete spinal segments were subjected to combined axial compression and forward flexion. Thus, for the isolated facets, axial compression is equivalent to spinal extension and axial tension to spinal flexion. These results have a significant impact on the understanding of injury mechanisms of the spine. The observation made by Nicoll (1949) that dislocations occur if the interspinous ligaments are ruptured is equivalent to saying that the capsules cannot provide much resistance in flexion and that facets can be easily subluxed if the posterior ligaments are torn. In fact, the geometry of the facet surfaces can be an important consideration as far as dislocation is concerned. Those with surfaces that are almost horizontal would be easier to dislocate than those with vertical faces, particularly in the presence of torsional loads and horizontal shear forces. The high frequency of dislocations at the C5-C7 level can be attributable to the facet geometry of those vertebrae.

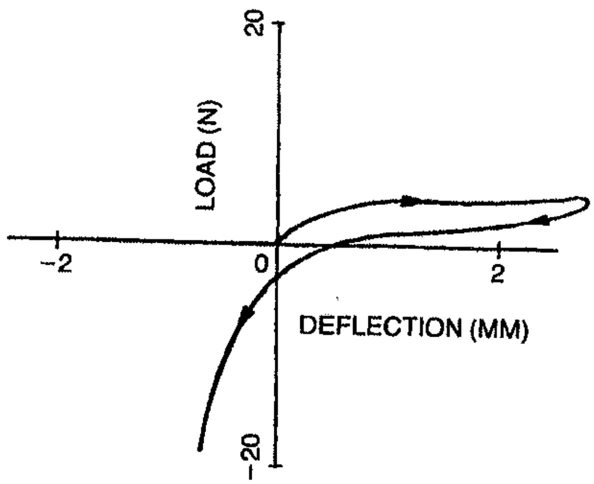


FIGURE 18.6. Response of an isolated facet joint to a tensile load. (From Yang and King, 1984.)

As further evidence of facet loading, contact pressure between the tip of the inferior facet and the lamina was measured by El-Bohy et al

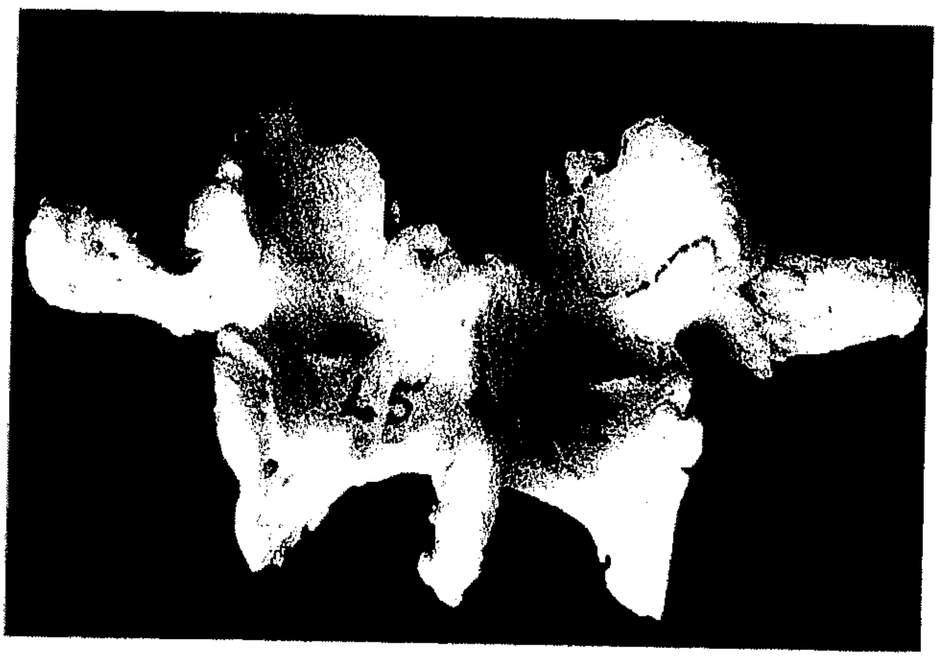


FIGURE 18.7. Heavy impression of facet tip contact with the lamina of possibly a weight lifter. (Courtesy of Dr. Jacob, Balgrist Hospital, Zurich.)

(1989). A miniature pressure transducer was inserted into the inferior facet through a predrilled hole so that its diaphragm was located at the very tip of the facet just above the lamina. The experimental setup is shown in Figure 18.8. The spinal segment was subjected to three types of loading: a simulated body weight that acted some 14 mm anterior to the center of the top disc, a simulated extensor muscle load that acted 50 mm posterior to the center of the top vertebral body, and an anterior eccentric load of 45 N simulating a weight borne by an individual. The pressure transducer was set to read zero pressure with the spine completely unloaded. A body weight equal to 50% of the body weight of the cadaver was then applied, causing the segment to flex. A simulated extensor muscle load was applied until the spine was returned to its erect configuration and the facet tip pressure was measured. In all six segments tested there was a measurable facet contact pressure. When the 45-N weight was applied at an anterior eccentricity of 340 mm, the segment again assumed a flexed configuration with no measurable facet pressure. However, when the extensor muscle force was used to bring the spine back to its erect position, a further increase in facet pressure was observed in five of the six segments tested. Statistically, this increase was significant at the 95% level.

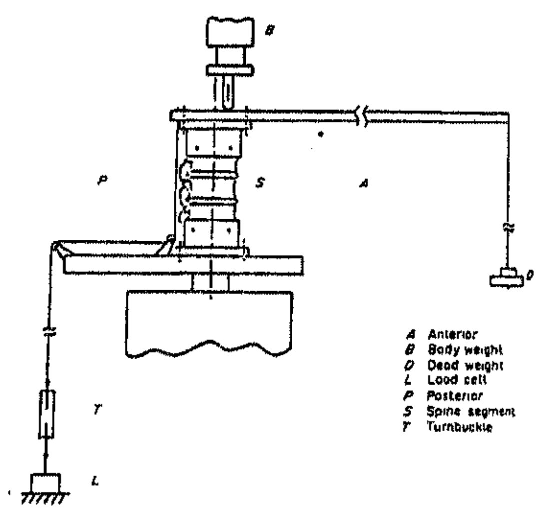


FIGURE 18.8. Schematic of the test setup for measuring facet contact pressure. (From El-Bohy et al, 1989.)

The biomechanics of a compressive force being generated in the vertebral column during a frontal impact ($+g_x$) acceleration needs to be discussed. This phenomenon was initially discovered when a two-dimensional model of the spine developed by Prasad and King (1974) was exercised to simulate a frontal impact. Cadaver tests were carried out by Begeman et al (1973) to verify the existence of this force, since vertical forces are not expected to be generated in a horizontal crash. Large seat pan loads were measured if the subject was restrained by an upper torso restraint, such as a cross-chest belt. This was the net force after accounting for all lap belt forces. Dummies did not generate this seat pan load. Subsequent tests by Begeman et al (1980) involving volunteer subjects confirmed these results. It was postulated that the seat pan load was a manifestation of spinal compression due to the tendency of the spine to straighten out during $-g_x$ acceleration. The same situation exists during a rear-end collision. The cervical spine is also subjected to a compressive force when the seat back pushes on the back of the torso. The neck ligaments and tendons are loosened by this compressive force, making it easier to cause shear damage to the soft tissues of the cervical spine, such as the facet capsules. With an inclined seat back the torso tends to ramp up the seat back, prolonging the compression of the neck. It may be necessary to decrease the distance between the head and the headrest to minimize the effect of shear on the neck. Research into the mechanisms of whiplash injury is ongoing at Wayne State University.

There is now anecdotal evidence of an increase in the frequency of thoracolumbar injuries among automotive crash victims, after the enactment of seat belt use laws in the United States. States et al (1989) reported a significant increase in spinal injuries as a result of the belt law in New York State. Hospital data from all hospitals in the Rochester area (Monroe County) were reviewed for injuries sustained 1 year before and 1 year after the effective date of the belt law. Assuming that the number of automotive related victims remained the same, there was a large increase in spinal injuries, as shown in Table 18.2. In addition, there is a case report of four anterolateral wedge compression fractures by Miniaci and McLaren (1989) that were attributed to the wearing of lap shoulder belts. Additional supporting evidence of compressive loading can be found in Anderson et al (1991), who found that patients with lumbar fractures were 5.7 times more likely to have been belted than patients with cervical fractures ($p < .001$). Orsay et al (1990) reported that there was a higher rate of spinal fractures in among belted occupants, 1.5% belted versus 0.9% unbelted. This difference was not statistically significant ($p = .29$). However, the rate of all spinal injuries was higher for belted individuals and significantly different from those sustained by unbelted occupants (61.6% belted versus 49% unbelted, $p < .001$). Porter and Zhao (1998) also found higher thoracolumbar injury rates among belted occupants compared to unbelted occupants, but the difference was again not statistically significant.

Soft Tissue Injuries to the Thoracolumbar Spine

Although the frequency of bony injuries to the thoracolumbar spine is relatively low as a result of vehicular crashes, that of soft tissue injuries to the intervertebral joints appear to be much higher. As mentioned earlier, the common complaint is low back pain (LBP), occasionally associated with pain radiating down the extremities. Since LBP is a very common complaint inasmuch as eight out of ten persons will have at least one such attack in their lifetime, it becomes extremely difficult to pinpoint the cause of any particular attack or to relate it to any given incident.

A difficulty in the diagnosis and treatment of LBP is its etiology. It is often called idiopathic, meaning it has no known cause. Recent research is homing in on at least a few of the causes of LBP, but by and large it is still difficult to determine the exact etiology. The pain can come from many sources and depends on the duration of the symptoms. The major culprits are believed to be the disc, the ligaments around the posterior structures of the spine, the facet joint capsules, and the tendons and muscles around intervertebral joint. Work by Yamashita et al (1990) focused on the source of pain, namely the presence and activation of pain-sensing nerve endings or nociceptors. The premise is that for someone to sense pain, there are three basic requirements:

1. Presence of nociceptors in the soft tissue in question
2. Adequate deformation of the soft tissue containing the nociceptors
3. Firing of the nociceptors in response to the deformation.

The first requirement is self-evident and needs no elaboration. However, the morphological study of the distribution of nerve endings in the intervertebral joint is still very much a topic of current research. Notably, one can point to the work of Giles and Taylor (1987)

TABLE 18.2. Change in lumbar injuries in Monroe County, NY, Between 1983–84 and 1985–86.

	Subjects[a]			Controls[b]		
	83–84	85–86	% Change	83–84	85–86	% Change
AIS 1–2	15	27	+80	16	19	+18.8
3–5	9	7	−22	2	4	+100.0

[a] Subjects are injured automotive occupants treated at county hospitals.
[b] Controls are nonautomotive traffic accident victims treated at the same hospitals (motorcyclists and pedestrians).

and Bogduk and Twomey (1987) as well as to a study by King et al (1990). There is some evidence that pain fibers are present in the outer layers of the intervertebral disc annulus. They are found in and around the capsules of the facet joints, extending to the borders of nearby ligaments and tendons.

The biomechanics of soft tissue deformation has also been studied. Stokes (1987) found that the strain of the anterior disc due to an applied compressive load was about 10% along the directions of the annular fibers. In Cavanaugh et al (1996), it was reported that the human facet joint capsule can undergo large deformations, particularly during spinal extension. In fact, during cadaveric experiments on spinal motion segments under static loading, the stretching of the superior lateral corner of certain facet capsules can be easily seen without optical magnification. The stretch can vary over a large range of strain, depending on the geometry of the facets. Table 18.3 shows data provided by Cavanaugh et al given in the form of stretch per unit moment (in N.m). The peak values of the flexion and extension moments applied were 24 and 18 N.m, respectively.

With respect to the third requirement, it is a well-known fact that nociceptors have high thresholds and do not go off unless stimulated vigorously. The large deformations noted in the facet capsules may be large enough but it is necessary to demonstrate this neurophysiologically. Yamashita et al (1990) and Avramov et a (1991) have mapped the distribution of nociceptors and other types of mechanoreceptors in and around the facet joint. They have also shown that spinal load and stretching of the facet capsule can cause the high threshold nociceptors to fire. This work confirms the previously reported facet syndrome by Mooney and Robertson (1976) and others. In fact, facet pain may be responsible for a large proportion of the LBP being reported. This combined neurophysiological and biomechanical technique is being extended to the disc, which appears to be sensitive to a change in the pH of its contents (Nachemson, 1969).

A significant finding of the research by Cavanaugh and colleagues was the lowering of the threshold of the nociceptors in the presence of spinal degeneration or of an inflammatory process. When the facet capsule was artificially inflamed with carrageenan, the nociceptors would fire spontaneously and would fire more rigorously if stimulated (Ozaktay et al 1994). However, the firing of nociceptors is non-adaptive. That is, they will continue to fire until the stimulus is removed, as opposed to other mechanoreceptors that cease to fire even though the stimulus is still acting. Thus, the pain resulting from a mechanical stimulus to a degenerated spine should be a temporary exac-

TABLE 18.3. Maximum tensile facet capsule stretch (%/N.m).

Cadaver no.	Extension tests			Flexion tests		
	X-axis[a]	Y-axis[b]	Z-axis[c]	X-axis	Y-axis	Z-axis
400	7.1	8.7	5.3	0.6	0.2	0.1
464	4.8	5.5	8.2	0.1	0.4	0.3
807	3.3	2.5	6.4	0.4	0.6	0.3
329	7.4	6.3	0.2	0.8	2.6	1.9
455	1.3	3.9	1.3	0.3	1.3	0.3
490	0.8	2.5	4.7	2.2	4.5	1.2
002	0.8	7.4	7.0	1.3	3.5	6.4
117	0.8	6.6	6.6	0.9	7.1	1.3
034	12.4	21.0	21.1	2.0	0.3	3.0
SD	4.0	5.6	6.0	0.7	2.4	2.0
CV	93.8	78.4	88.7	77.5	104.5	122.8

[a]The X-axis is directed anteriorly in the transverse plane.
[b]The Y-axis is directed laterally to the left in the transverse plane.
[c]The Z-axis is directed superiorly, normal to the transverse plane.

erbation of a painful condition. Another aspect of pain is its ability to initiate a vicious cycle of pain that is brought on by the production of neurotransmitters, such as substance P. These transmitters travel to the site of pain and further decrease the threshold of the nociceptor, resulting in more pain. This phenomenon explains the observed cycle of back pain in some individuals whose back pain gets worse day after day until they can no longer get out of bed. Then the cycle is reversed and they recover practically without any treatment. Nevertheless, the exacerbation is still temporary and should last from a few days to a few weeks. Continuous complaints of chronic pain without remission are possibly due to psychological or other physical causes. One such physical cause has been described by Ozaktay et al (1995) and Chen et al (1997). They have shown that the toxic chemical phospholipase A_2 (PLA_2) found in the nucleus pulposus can destroy nerve endings and cause the axons in the nerve root and the dorsal root ganglion to send nociceptor signals up the spinal cord, signaling referred pain in the lower extremities. The sympathetic nervous system can also be involved in the form of abdominal visceral pain at the level of L2-3 for a degenerative disc condition in the lower lumbar spine (Nakamura et al, 1996).

Research into facet pain may have provided an incentive for neurologists and anesthesiologists to treat back pain nonsurgically with facet injections and rhizotomies. There is, however, a predominant majority of physicians who attribute back pain solely to the intervertebral disc. Of course, surgical intervention usually involves the disc and there are no surgical procedures for facet pain. The reimbursement incentive is reinforced by the traditional belief that disc pain is synonymous with low back pain. This indeed is a concept that has outlived its scientific usefulness. While it is true that a degenerated disc can cause pain, it can induce facet pain due to loss of disc height, leading to an increase in facet load, and there are many other causes of back pain. Low back pain is complex problem that cannot be neatly explained by a single cause. Much research remains to be done, but for minor impact loads the complaint of pain cannot be equated with permanent injury of the soft tissue involved.

With regard to the relationship between disc rupture and impact loading on the spine, it can be safely said that disc ruptures do not occur as the result of a single loading event, unless there are associated massive bony injuries to the spine. This statement is based on a review of the literature on spinal response by Henzel et al (1968), who indicated that early researchers such as Ruff (1950), Brown et al (1957), and Roaf (1960) observed that the vertebral body always broke before the adjacent disc incurred visible damage. Moreover, Brinckmann (1986) has shown that a severely weakened lumbar disc, with the posterior elements removed, could not be ruptured and hardly even bulged when loaded in compression to 1kN. Additional loads causing fracture of the vertebral body did not result in herniation or excessive bulging. There are two reports of disc rupture due to a single loading event in the literature. Farfan et al (1970) applied torsional loads to intact lumbar motion segments without any compressive preload and was able to cause posterior and anterior disc ruptures after a rotation averaging 22.9 degrees for normal discs and 15.2 degrees for abnormal discs. They also tested facet joints and facet capsules to failure and found that the average angle at which they failed was 14 and 12 degrees, respectively. This meant that if the facets were allowed to slide over each other or fracture, resulting in a large rotation, then the rupture can occur. Normally, in the presence of a preload of the facets, a single torsional load that does not disrupt the facets or tear the capsules does not cause rupture. The second report is by Adams and Hutton (1982), in which they caused spontaneous rupture of several discs by compressing the spine while it was hyperflexed both laterally and sagittally. If the disc did not rupture on the first try, it was flexed 1 or 2 degrees more and loaded again with the same load. The average angle of flexion was 12.9 degrees, implying that the lumbar spine alone was flexed a total of 64 degrees. The average applied load was 5,449N (1,225lb). This situation is again not representative of a realistic loading condition, as it is extremely rare that a large compressive force would be applied to a spine that is virtually doubled over. Moreover, the herniation occurred between the disc and the endplate due to extreme tension on the posterior aspect of the disc. There was no rupture of the annulus. In fact, both of these reports tend to reinforce the point of view that a single loading event is unable to cause disc rupture.

Under repeated loading, Yang et al (1988) were able to produce disc herniation (extrusion of nuclear material to the outside of the disc) by the application of repetitive torsional loads combined with compression and flexion and with the facets removed. Figure 18.9 shows extrusion of nuclear material (white arrow) after approximately 20,000 cycles of torsional loading. In a subsequent study by Gordon et al (1991), nuclear extrusion occurred in four of 14 specimens that were tested under combined axial load, flexion, and torsion, with the posterior elements intact. The average number of cycles of loading was 36,750. Previous attempts by Hardy et al (1959) resulted in the extrusion of the nucleus into the endplates and compression fracture of the endplates and the vertebral bodies. Only one unembalmed specimen was tested and failure occurred at 1.29 million cycles. Liu et al (1983, 1985) performed fatigue tests on lumbar motion segments under axial and torsional loads. Under axial loading, failure occurred in the form of fracture of the vertebral body or of the endplates. Under torsional loading there were some annular tears observed but no extrusion of disc material. Figure 18.10 shows the tortuous path taken by the nucleus material before it can exit the disc. The annular layers do not rupture along a straight line and the nucleus material is too viscous to move through this maze under an impulsive load. Disc rupture appears to be the result of a slow degenerative process that takes a long time to develop. In other words, discs do not rupture like a balloon, allowing all of the material in the center to come out at once.

Case Histories of Claims of Back Pain Due to a Minor Incident

Cases of back injury claims following a minor incident, such as a fender bender crash, abound

FIGURE 18.9. Photograph of extrusion of nucleus pulposus following several thousand cycles of torsional loading. (Courtesy of Dr. K.H. Yang, Wayne State University, Detroit.)

FIGURE 18.10. Photograph of a coronal section of a ruptured lumbar disc showing the tortuous path created by the rupture of individual annular layers. (Courtesy of Dr. K.H. Yang, Wayne State University, Detroit.)

in the United States. In this section, examples of claims made in automobile crashes, elevator incidents, and other impact-loading conditions are given to illustrate the need for more engineering analysis and less dependence on the history provided by the patient or claimant. It is also important to rely heavily on the claimant's prior medical history when a decision has to be made as to whether the incident in question is causally related to the diagnosed condition.

Case 1—Rearend Impact

This typical case involves a 32-year old male driver who was stopped at a traffic light and was rearended by a car that failed to stop in time. The speed of impact was difficult to estimate because photographs of the bumpers of both cars show no damage, and in the absence of damage no reconstruction can be performed, particularly when the vehicles are no longer available. He complained of immediate back and neck pain and was seen in the emergency room shortly after the incident. X-rays showed spinal degeneration in his neck and low back, with arthritic lipping in both spines. Eventually, magnetic resonance imaging (MRI) showed that he had a ruptured lumbar disc at L4-5. He filed a claim for sustaining the injury as a result of this minor impact. However, his prior medical history indicated that he had made visits to chiropractors and orthopedic surgeons for complaints of back pain and that his degenerative condition preexisted the incident. A causal relationship could not be established because discs do not rupture as a result of a single impact, even though a rear-end impact exerts a compressive load on the lumbar spine. The immediate complaint of pain confirms the inflammatory condition of his degenerated spine. He may have sustained a temporary increase in pain level but no permanent injury resulted. The jury awarded him the cost of his ambulance ride to the hospital after the crash.

Case 2—Frontal Impact

On July 4, a 46-year-old male unrestrained driver hit an errant trailer, which had broken free from a vehicle going in the opposite direction. The entire front end of his vehicle was damaged and the estimated repair cost was about $4,000. However, he did not sustain any rib fractures and the only visible injury was a mid-forehead abrasion, with abrasion of the left extremity and a bruised left knee. He initially complained of neck pain and an MRI done on July 30 revealed the existence of bony spurs in his cervical spine at several levels. However, on July 17 he started to complain of low back pain. A computer tomography (CT) scan done on July 28 revealed a grade I spondylolisthesis, disc degeneration at L5-S1, and facet joint degeneration at L4-5 and L5-S1. His lumbar range of motion was limited, especially in extension. However, his back pain was cyclic, reaching a peak every 3 or 4 months. During periods of remission, he was able to do weight-lifting exercises. His pain became much worse 3 years after the crash and an additional MRI indicated a ruptured disc at L3-4 with dehydration in all of the lumbar discs. It can be concluded from this medical history that the disc rupture is a natural progression of his spinal degeneration and not causally related to the crash. His treating physicians failed to recognize that one of the sources of his pain was the lumbar facets, but they decided to treat him nonsurgically because his pain was over multiple spinal levels.

Case 3—Mining Vehicle

This was a landmark case in which a claim was made that a lumbar intervertebral disc was possibly ruptured as the result of a mining vehicle (a tractor) going over a crib block, which is a piece of timber with a $3\frac{5}{8}$-inch-square cross-sectional area. The tractor is a heavy four-wheeled vehicle that has no suspension and is controlled by an operator who is seated on the right side of the vehicle ahead of the front wheel and facing the left side of the vehicle. Its top speed is about 4 mph and when it climbs over a $3\frac{5}{8}$-inch obstruction, the maximum seat acceleration averaged $3.9g$. This peak was found to occur at the time of impact of the right front wheel with the ground after it has gone over the obstruction. These data were obtained from tests done above ground on an asphalt surface at 4 mph. The peak acceleration at the pelvis for an unbelted dummy averaged $5.86g$ over five tests. The measured compression load on the lumbar spine of the dummy averaged 220 lb. Such a load can be generated if the claimant had lifted a 20-lb bag of groceries 11 inches in front of his spine. The peak acceleration at the pelvis for the belted runs was $4.76g$, and the corresponding lumbar load was 250 lb.

The operator complained of immediate low back pain following the incident and had cyclic flare-ups subsequently. There was no clear evidence of a disc herniation but subjective complaints of pain persisted. The manufacturer of the tractor was accused of not installing a lap belt restraint system, which would have prevented the operator from being thrown into the canopy and landing hard onto the seat. However, the measured g-levels at the pelvis of the dummy were the same whether it was belted or not. Since the scientific literature does not support disc rupture following a single impact load, the issue of disc rupture became a dispute between the biomechanical expert and the claimant's treating physician. The fact that this 28-year-old man had a badly degenerated spine with evidence of preexisting facet pain and the fact that his pain was cyclic in nature pointed to the temporary exacerbation of his facet pain at the time of the incident and to the absence of a permanent injury. The discrepancy between the medical and bioengineering testimony was explained to the members of the jury. They were told that the treating physician based his/her opinion entirely on the history provided by the patient. This physician had no knowledge of the severity of the impact and was not trained to perform biomechanical analyses or tests to correlate the test data with human tolerance to injury. An important point that was made was the fact that the immediate onset of pain does not necessarily mean that it was a disc rupture. There are many sources of back pain, and disc pain is only one of the sources. The jury found for the defense, and subsequent mining vehicle cases were not pursued based on the theory that lap-belts would have prevented these alleged injuries.

Case 4—Railroad Injury

A railroad worker claimed that he was injured while he was trying to line up drawbars between freight cars so that they can be connected up to form a train. The drawbars are about 40 inches above the ground. They weigh about 350 lb and are about 5 ft long. They pivot about a pin under the car and most of their weight is concentrated at a knuckle at the free end that consists of a block of cast iron 15 by 14 inches in cross-section and 12 inches long. The drawbars slide over a steel wear plate, over a distance of about 19 inches because the pivot is 33 inches from the wear plate. It takes about

70 to 80lb to get the bar to move, and the old way of moving the bars was to have the back against the knuckle with the back facing the direction of motion of the knuckle. Both hands were to be used to lift it slightly so that it could be made to slide and be centered. The worker claimed that the bar was all the way to one side of the train, and that when he tried to move it at first it would not budge, but as he applied more force it slid all the way over to the other side, a distance of about 32 inches. This caused him to fall to the ground, hitting the rail with his left side. He complained of immediate low back pain and claimed a disc rupture. He did not seek medical treatment for about 3 months, and when he did no disc rupture was ever diagnosed. However, he complained of intermittent but severe low back pain and was declared totally disabled. He was found to have a similar history of intermittent low back pain prior to the incident.

A biomechanical analysis was made as to the time needed to push the knuckle from one side to the other. If a constant 100-lb force was used and if the coefficient of friction was assumed to be only 0.2, the time it took to move the knuckle 32 inches was about 0.6 second. There was no time for him to take a step during this period but he would not fall down because the knuckle would hold him up. If the force was less and the coefficient of friction was higher than 0.2, the time interval could be as long as 2.9 seconds. That is, he could have taken a step if the time interval was longer than 1 second and would lean against the knuckle if it was less than a second. In either case, he could not have fallen to the ground as he claimed. Furthermore, even if he did, the load on the lumbar spine would not be directly compressive and would not be able to herniate a disc. The arguments described in case 3 above were used to explain the source of his pain and to show that the alleged incident was not causally related to his claim of injury.

Case 5—Stepping into Wet Cement

A young woman, carrying a baby in one arm and a child seat in the other, stepped into some wet cement because no barrier was put up to warn pedestrians that the sidewalk was being repaved. She had just left a chiropractic clinic where she was receiving treatment for back pain and numbness in her leg. Her prior employment history includes service in the U.S. Army and in the army reserves. She claimed to have ruptured a lumbar intervertebral disc as a result of this mishap and justified the claim on the basis that this incident was the "last straw" that broke her disc. The accelerations she experienced would be less than those she would have sustained had she placed her foot onto solid concrete. The viscous wet cement would definitely cushion her foot contact with the ground. The fact that such incidents produce acceleration levels at or below those normally experienced during activities of daily living should be pointed out. Allen et al (1994) did a series of experiments involving activities of daily living and measured head acceleration in two directions—anteroposterior and superoinferior. They found resultant acceleration levels in the range of a few g's up to $10g$ when one of his subjects plopped heavily into a chair. Thus, there is no viable "last straw" theory when the impact g-level is within those experienced in activities of daily living. Furthermore, Brinckmann (1986) experimentally punctured the annulus of a human lumbar disc with a spinal needle and loaded it after the needle had been withdrawn. No nucleus material exited the disc through the original puncture. He used the needle as a guide to insert a small blade that was used to cut the annular fibers of the disc on the far side of the puncture. He left the last millimeter of the outer layer of the annulus intact but the rest of the fibers were cut. No ruptures occurred at this weakened section even when the vertebral body was loaded to fracture. At 1,000N, the disc did not show any significant bulge at that location. This study, along with others mentioned above, constitute the bulk of the evidence that discs are not ruptured as the result of a single loading event. The zigzag path shown in Figure 18.10 and taken by the nucleus pulposus to exit the disc explains why disc ruptures are a slow, degenerative process.

Case 6—Bus Going Over a Pothole

A bus driver claimed he ruptured one of his lumbar intervertebral discs when his bus drove over a large pothole. The bus manufacturer and the manufacturer of the driver's seat were both held liable because it was alleged that the seat magnified the chassis acceleration, causing the driver to sustain this back injury. Tests were done to re-create the alleged incident and the g-levels measured on the bus as well as on the seat cushion of the driver's seat were all in the range of 0.8 and $1.2g$. Thus, the argument of load magnification is moot and g-levels of this magnitude could not produce a disc rupture. The driver's prior medical history was devoid of any back complaints. However, postincident CT scans revealed degenerated facet joints at all levels from L3 down, but there was no evidence of any disc rupture. The claim was based solely on subjective complaints of pain. Here again, the existence of facet pain was ignored by his treating physicians, who were at a loss as to what treatment should be given. Nevertheless, they testified that his back "injury" was causally related to the pothole incident.

Case 7—Sudden Drop of an Adjustable Seat

Drop of an Office Chair

In this case, a secretary claimed she sustained a low back injury because the seat she was using suddenly dropped from its highest position to the lowest without any intervention on her part. The total distance of the drop was about 3 inches. Tests were conducted using an exemplar chair with the feet on and off the ground and with the chair on a hard floor as well as on a carpeted floor. The measured vertical acceleration on the seat pan or seat cushion did not exceed $0.6g$ in any of the 17 tests that were made. The plaintiff's attorney was present during the tests because they were conducted at the site where the incident allegedly occurred. He arrived at a mutually acceptable settlement on the spot.

Drop of a Bus Driver's Seat

A bus driver claimed that one of his lumbar discs was ruptured when his bus driver's seat sudden dropped 4 inches without any intervention on his part. The seat height, in fact, could only be adjusted if a foot pedal was depressed. In spite of this, a series of tests was conducted to measure the vertical acceleration experienced by a Hybrid III dummy for this 4-inch drop with the foot pedal fully depressed. The measured seat cushion accelerations were between 2.0 and $2.9g$. The conclusion was that no disc rupture could have occurred as a result of this drop. After the jury rendered a defense verdict, the treating physicians of this bus driver became very irate and wrote to the attorneys that they have seen thousands of disc ruptures from impacts and loads much less severe than this drop, such as picking up a potato chip. Such statements demonstrate that these physicians not only are ignorant of the many causes of back pain but also have erroneously relied solely upon the history provided by their patients to establish causality. They have not seen those discs rupture in front of their eyes. What they saw, eventually at surgery, was a ruptured disc, which was attributed to a precipitating event that could be blamed on someone else. They also must know that there are just as many if not more cases of disc rupture that occur without any precipitating event.

Mathematical Models of the Thoracolumbar Spine

Mathematical models can often be used as an experimental tool to study the response of a system to a variety of input conditions. The premise is that the model must provide reliable predictions before this type of study can be of value. In other words, models that have been validated against experimental data are required. King and Chou (1976) reviewed mathematical models of impact developed before 1975 and discussed in detail models of the spine that were available at that time.

Two-dimensional models developed after 1975 include discrete parameter models by Tennyson and King (1976) and by Pontius and Liu (1976). Both were extensions of previous models without muscles, and both had a capability of simulating a delayed response of the musculature following a stretch stimulus. Validation against human volunteer impact data was provided by Tennyson and King (1976, 1977).

Since 1975, many three-dimensional (3D) models have been developed to simulate the response of the spine to impact acceleration. Belytschko and Privitzer (1978) demonstrated the capability of a 3D model of the entire spine. It is basically a discrete parameter model in which the vertebrae are represented by rigid bodies interconnected by deformable elements. It was capable of simulating ejection seat dynamics was well as spinal response to a horizontal crash ($-g_x$ impact acceleration). There was, however, no validation against experimental data.

A finite element model of a motion segment was developed by King and Yang (1985) to simulate the response of the articular facets, using the data detailed by Yang and King (1984). This model is based on an earlier model of Hakim and King (1979) for a single vertebra and computes stress distribution in the body, disc, and posterior structures. The disc has a fluid nucleus surrounded by a low modulus annular material. It is still a static model, which can be made to respond to dynamic inputs at a later date. The computed intradiscal pressure compared favorably with experimental data. In a parametric study, it was found that the sensitive variables were the modulus of elasticity of the annulus fibrosus and spongy bone.

More recent models include the work of Shirazi-Adl (1984), who developed a comprehensive 3D finite element model of a lumbar motion segment and has been simulating a variety of static loading conditions, such as the prediction of facet load in the lumbar spine (Shirazi-Adl and Drouin, 1987).

Discussion

Because spinal injuries are relatively infrequent in automotive accidents, research on the biomechanics of spinal injury and response is not as advanced as that for body regions that are frequently injured, such as the head or the thorax. For the thoracolumbar spine, the tolerance information is over 30 years old for $+g_z$ acceleration, and there has been no new information since the publication of the curves by Eiband (1959). The fact that tolerance is dependent on the restraint system used and the age of occupant renders the problem of defining it as a single parameter virtually impossible. Furthermore, the configuration of the spine plays an important role in the injury pattern it sustains as a result of an impact in a given direction. In the absence of a restraint system, more spinal injuries are likely to occur, especially for motorcycle riders and occupants involved in rollovers. Thus, it is also extremely difficult to arrive at a limited set of injury criteria for the spine, particularly since the failure of the spinal components is not restricted to the bony portions of he spine. Much of the spinal resistance to bending and torsional loads is provided by soft tissues, ligaments, muscles, and cartilage (disc). For each set of loading conditions, injury criteria need to be formulated. Such a task is indeed formidable.

The mechanisms of injury to the spine are relatively well understood. Failure of the various spinal components can be attributed to a combination of axial and bending loads. The central role played by the articular facets cannot be overemphasized. Together with the vertebrae, they provide a dual-load path for the transmission of axial load. In forward bending, the compressive load may be borne entirely by the body. However, new data being acquired from spinal segments appear to indicate that the facets are load bearing until there is excessive flexion. In the living spine, the role of the musculature of the back acts to increase facet loading. Torsional loads and shear loads are also resisted to a great extent by the facets. In the lumbar region, Patwardhan et al (1982) may have measured this torsional resistance, using pressure-sensitive Fuji film.

The mechanism of disc rupture has been studied extensively, and all scientific evidence points to a slow, degenerative process. The inability of a disc to rupture under a single load or impact is frequently contradicted by physicians, particularly by the treating physicians of the claimants. Their opinion that the diagnosed ruptured disc resulted from a specific incident is based on two factors: patient history, which is generally not supported by a quantitative analysis of the loads involved; and the patient's prior medical history. Frequently, the treating physician is unaware of the patient's complaints of back pain prior to the incident in question. The fact that the many causes of back pain are still unknown leads physicians to hold on to the few things they do know about back pain, namely, that disc pain is the result of a disc bulge or a disc rupture. The use of this simple-minded approach to establish a causal relationship between disc rupture and a possible precipitating event is not only irresponsible but also a remarkable demonstration of scientific ignorance. Even though some U.S. courts have erroneously barred expert testimony from biomechanical engineers regarding causality, it is still the opinion of this author that physicians without expertise in impact biomechanics lack the qualifications to determine causality, and that the only professionals who can do so are impact biomechanicians. In fact, there have been court decisions barring physicians from giving testimony regarding causality. This issue is now more critical because clearer guidelines have been drawn by the U.S. Supreme Court regarding the admissibility of scientifically based testimony under the Daubert decision. That is, biomechanical opinions should be based on peered reviewed publications and/or on computational models that have been validated against experimental data. By the same token, medical opinions on causality based solely on the history provided by the patient should be excluded under these same guidelines. Finally, the biomechanician need not apologize for arriving at an opinion that is different from that of the treating physician. As long as that opinion can be backed up with good science, the biomechanician can simply explain that differences can arise from the use of a totally different source of information—analysis versus history—and the former is far stronger than the latter.

Conclusions

1. Spinal anatomy is extremely complex. A thorough knowledge of its construction and function is essential to understanding its biomechanical role in load bearing and protection of the spinal cord.

2. The articular facets play a central role in the mechanism of spinal support and in the mechanisms of injury to the spine. The orientation of the facet surfaces may explain the tendency of subluxation and dislocation of certain vertebrae. It should be noted that subluxation is an unstable condition that endangers the integrity of the cord.

3. The manner in which facets resist compressive and tensile loads has been studied by means of tests on isolated facet joints. It can be hypothesized that compressive resistance is generated by the bottoming out of the tip of the inferior facets onto the pars interarticularis of the vertebra below. Proof of this hypothesis will require additional research. The facets offer virtually no resistance to tensile loads. The smooth surfaces of the facets and the relatively weak capsular ligaments are unable to provide the tensile resistance necessary for countering such loads. It is postulated that the ligaments and muscles along the posterior spine assume this role.

4. Impact loads are not likely to produce ruptures of the lumbar intervertebral disc. Injuries to the soft tissues of the intervertebral joint are, however, difficult to diagnose. At present, there is adequate evidence to state that the facet capsule is a source of low back pain.

5. Tolerance data are woefully lacking. A carefully worked-out research plan is needed to address this complex problem of injury tolerance and injury criteria. In particular, there is a need to study the tolerance of the facets to dislocation, which is a high-risk injury.

The types of dislocation and the injury mechanisms involved need to be identified at each spinal level and for each combination of applied loads.

6. The following observations can be made from the case presentations discussed above:

a. Temporary exacerbation of back pain is not a permanent injury. It is merely the firing of sensitized nerve endings, which should not last more than a few days or weeks.

b. Facet pain is a common source of back pain. One can readily assume why they are seldom diagnosed by surgeons who operate on ruptured discs.

c. Disc degeneration causes a loss of disc height. This increases the load on the facets, which in turn causes the facet capsule to stretch, setting up the conditions necessary for the nociceptors to fire.

d. Performing an engineering analysis can often demonstrate the impossibility of an alleged event. It is always comforting to know that no lawyer has the ability to repeal Newton's laws of motion.

e. There is no "last straw" theory when it comes to disc ruptures. They are a slow, degenerative process that can take many thousands of cycles of loading and that does not have a precipitating event.

f. Conducting relevant tests that re-create the alleged incident may be costly but is frequently the most effective way to establish or dismiss causality.

g. Physicians with no expertise in impact biomechanics lack the necessary background and qualifications to opine on causality and injury mechanisms, particularly if their opinions are based entirely on the history provided by the patient. Their job is to treat the injury or complaint. The impact biomechanician is the only professional qualified to determine causality and the mechanisms of injury.

7. The use of mathematical models to study the response of the spine is a viable approach because of the flexibility of the models and the low cost involved in comparison with that incurred in experimental research. Whenever possible, the combined approach of using models and obtaining experimental data based on model predictions will enhance the understanding of spinal response.

The Pelvis

Anatomy

The pelvis (Latin for "basin") is a ring of bone interposed between the flexible spinal column, which it supports, and the movable lower limbs, upon which it rests. Mechanically, it is the only load path for the transmission of the weight of the head, arms, and torso to the ground. Thus, the pelvic structure is more massive than that of the cranial or thoracic cavities. The pelvic ring or girdle is composed of four bones. Two hipbones form the side and front walls of the ring, while the sacrum and coccyx make up the rear wall. Figure 18.11 shows a frontal view of the male and female pelvis. There are many differences between the male and female pelvis, but the principal functional difference is the shape of the inner cavity that is completely surrounded by bone. This aperture is almost circular in the female, while it is wider in the side-to-side direction in the male. In the female, this aperture is the birth canal through which the fetus must pass. The orientation of the pelvis is less variable in the standing posture than that of the seated posture. A method for defining pelvic orientation will be discussed below.

Hipbone

The hipbone is a large, relatively flat, and irregularly shaped bone that forms the greater part of the pelvic girdle. It is formed by three fused bones called the ilium, ischium, and pubis. The fusion occurs around a cup-shaped articular cavity called the acetabulum (hip socket), which is situated near the middle of the outer surface of the bone. The ilium makes up the flank of the hipbone and is the upper broad and expanded portion that extends upward from the acetabulum. The ilium is divided into two parts—the large wing-like ala and the body of the ilium that forms a part of the acetabulum.

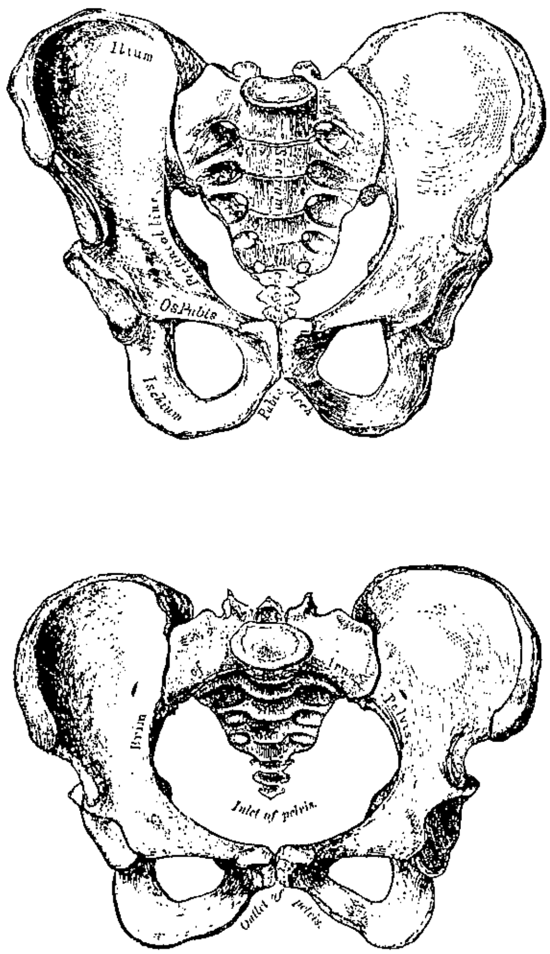


FIGURE 18.11. Frontal view of the male and female pelvis. (From Gray's Anatomy, 1973.)

Most of the landmarks and surface features are of little biomechanical significance and will not be discussed. However, it is necessary to point out the anterior-superior iliac spine (ASIS), which acts as an anatomic anchor point to prevent the lap belt from slipping over the top of the pelvis, a phenomenon called "submarining." The ilium is attached to the sacrum along its medial (inside) surface called the auricular surface. This surface is covered with cartilage, but the joint is not synovial. It is therefore postulated that only a limited amount of relative motion can occur here.

The ischium forms the lower and rearward part of the hipbone and is divided into two parts—a body and a ramus. The former constitutes the rearward third of the acetabular cup. The lowest portion of the body is the ischial tuberosity, which supports the upper torso in a seated posture. The ramus is a thin flattened part of the ischium that joins with the inferior pubic ramus. The pubic bone is an irregularly shaped bone composed of a body and two rami, the superior and inferior pubic rami. The body forms the front third of the acetabulum. The upper ramus extends from the body to the midsagittal plane, where it articulates with the corresponding ramus on the opposite side. The joint formed by the two superior pubic rami is called the pubic symphysis, which is a slightly movable joint containing a cartilaginous disc between the two bones. The lower pubic ramus joins with the ramus of the ischium to form the bottom arch of the obturator foramen. The inferior pubic rami likewise join each other through the pubic symphysis.

Sacrum and Coccyx

The rear wall of the pelvic girdle is composed of the sacrum and the coccyx. The former is a fusion of five sacral vertebrae and is triangular in shape. Its auricular surfaces form a solid joint with the pelvis through which the weight of the upper torso is transmitted to the legs or the ischial tuberosities. The bone is concave toward the front and supports the lumbar spine at the top.

The coccyx is a vestigial tail made up of three to five fused vertebrae. The vertebrae do not have all of the features of normal vertebrae, inasmuch as they are devoid of all the posterior structures and the last vertebra is a mere nodule of bone. The sacral-coccygeal joint is a slightly movable joint, interposed by a thin disc of cartilage that is stiffer than a normal intervertebral disc.

Orientation of the Pelvis

This orientation varies from person to person and is different in the standing and seated posture. Quantification of orientation requires a plane of reference that appears as a straight line when the pelvis is viewed laterally. Nyquist and Murton (1975) proposed that the plane formed by the two ASIS and the pubic symphysis can be used as a reference. The three landmarks are palpable, and x-ray is not needed to identify them. Although a method exists to define pelvic orientation, it is not known whether a large body of data exists for either the standing or seated posture. According to Gray (1973), the variation in pelvic orientation in the standing posture is about 10 degrees.

Proximal Femur

Since the femur articulates with the pelvis and injuries are often located in the vicinity of the hip joint, it is necessary to discuss briefly the anatomy of the proximal femur. The femur is the longest and strongest bone in the body and is roughly cylindrical in shape along its shaft. However, the shape of the two ends is different from that of the shaft. In particular, the proximal or upper portion of the femur is made up of a head, neck, and trochanteric region. A view of the proximal femur from the rear is shown in Figure 18.12.

The head of the femur has a spherical shape, forming a synovial joint with the acetabulum. This joint is commonly known as the hip joint. The surface of the head is lined with a layer of hyaline cartilage, typical of a synovial joint. Its blood supply comes through the neck of the femur, which is the structure below the head.

The neck of the femur is shaped like a truncated cone and connects the head with the rest of the femur. It is almost cylindrical at the base of the head, and its axis is at an angle of about 125 degrees with the femoral shaft. In the female, this angle can be as low as 90 degrees. The lower boundary of the neck is defined by the intertrochanteric crest in the rear and by the intertrochanteric line in the front. Both boundaries are well-defined ridges that act to demarcate the neck from the trochanteric region.

The trochanteric region is an enlarged portion of the proximal femoral shaft distin-

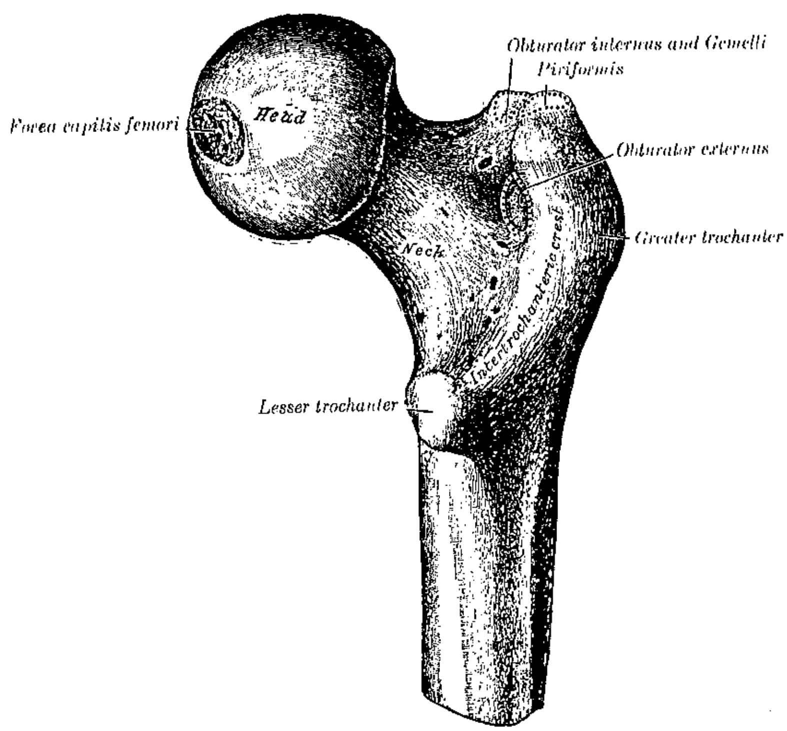


FIGURE 18.12. Rear view of the proximal (upper) femur. (From Gray's Anatomy, 1973.)

guished by two landmarks—the greater and lesser trochanter. The former is a prominence that extends laterally from the base of the neck and is easily palpable in most people. The level of its upper boundary corresponds to the center of the femoral head or the center of the hip joint. The lesser trochanter is a smaller prominence on the inside of the femur. It is visible on x-ray but not palpable.

The proximal femur is covered by a layer of cortical or compact bone and contains spongy or cancellous bone within. The strength and density of the spongy bone varies with age and sex. Hip fractures, defined as fractures to the head, neck, or trochanter, are often attributed to a fall or to impact trauma and the fracture types are termed subcapital, transcervical, intertrochanteric, and pertrochanteric depending on the location of the fracture. However, Yang et al (1996) have proposed a new mechanism of injury to the hip. They hypothesize that these fractures are due to sudden and strong muscular contractions of the hip muscles, resulting in a fall to the side of the fracture. Many arguments were provided to show that falling to the side and fracturing one's hip is less likely than a spontaneous fracture of the hip, culminating in a fall. One of the most convincing pieces of evidence is the virtual absence of hip fractures among occupants of cars that were broadsided. The nearside occupant is struck in the greater trochanter by the intruding side door but they are found to sustain pubic rami fractures instead of hip fractures.

Pelvic Injuries in Clinical Experience

A survey of pelvic injury patterns as reported in the clinical literature was made in an effort to understand the various types of injuries to this body structure. Although the more recent clinical literature tends to report injuries to the pelvis due to auto accidents, the reports generally cover pelvic injuries from all causes, such as crushing injuries to miners, falls, and excessive muscular activity. Many of the papers describe injuries to the hip joint and the proximal femur along with those of the bony pelvis. The extension into the hip joint has a sound clinical basis, since the treatment of the hip and upper femur involves the pelvis, and the hip and pelvis are frequently injured simultaneously. This discussion therefore includes injuries to the hip joint and the upper femur.

This review is divided into pelvic injury types, hip injury types, and clinical incidence. A comprehensive summary of injuries sustained by the entire anatomical region was found in Watson-Jones (1976), and that classification is presented here.

Pelvic Injuries

Injuries to the pelvis, not involving the hip joint, can be classified under four distinct headings: avulsions of muscle insertions, isolated fractures of the pelvic ring, double or multiple fracture of the pelvic ring, and fractures of the sacrum and coccyx.

Avulsions of Muscle Insertions

Due to excessive muscular activity, a portion of the pelvis can be avulsed or separated from the main bone at the point of muscle insertion. For example, the ASIS can be avulsed by the sartorius muscle, or the hamstrings can avulse the ischial tuberosity. These injuries occur in young adults who are engaged in athletic activities, and they are not the result of impact-type accidents.

Isolated Fractures of the Pelvic Ring

If there is a single fracture around the pelvic ring, significant displacement of the fractured segments does not arise. Unilateral fractures of the superior or inferior pubic rami, or of both rami, are examples of isolated fractures that do not require surgical intervention. Other isolated fractures include single fractures of the ilium with minor separation of the pubic symphysis, and sacroiliac subluxation (slippage). These injuries appear to be the result of minor impacts. For example, in a side impact aimed at the greater trochanter, pubic rami fractures frequently occur.

Multiple Fractures or Fracture-Dislocations

With multiple fractures, the pelvic ring becomes unstable. Large displacements of the fragments are possible. The two main types of injury are fractures of the pubic segments only and fractures of the pubic bone associated with fractures of the ilium.

Multiple injuries of the pubis consist of two or more fractures of the rami with dislocation of the pubic symphysis. They occur in pedestrians struck on the side of the pelvis by an automobile. In general, fractures of the rami occur on the nonimpacted side. If there are bilateral multiple fractures of the rami, the impacted side cannot be firmly established. Although surgical intervention is not required, the patient can suffer urinary tract injuries, such as rupture of the urethra, a duct that conducts urine out of the body from the bladder (Wiggishoff and Kiefer 1968).

The most common type of combined injuries of the iliac and pubic segments is the dislocation of the pubic symphysis with dislocation of the sacroiliac joint. Other types of injury include the fracture of the ilium accompanied by a dislocation of the pubic symphysis or fracture of both rami on one side with sacroiliac dislocation on the same side. Bladder and urethral injuries are common in these cases. The mechanism of injury is a force applied from the front to the back as opposed to a sideward force in the previous case.

Injuries to the Sacrum and Coccyx

Extensive injuries to the pelvis can result in fractures of the sacrum. These usually occur in regions of stress concentration, that is, across the foramina or holes through which the sacral nerves pass. There may be associated nerve injury. Vertical acceleration can cause compression of the sacrum with a loss in height. Coccygeal injuries are sustained in falls in a sitting position. The injury can be a contusion, fracture, or dislocation of the coccyx. Although it is not a serious injury, it can be extremely painful.

Associated Soft Tissue Injuries

The most serious soft tissue injury associated with pelvic fractures is hemorrhage. It comes from large blood vessels in the pelvic wall as well as from the fractured surfaces themselves. The amount of blood loss can be large, and one of the methods to stop the hemorrhage is to ligate the internal iliac arteries. Injuries to the bladder and urethra and to the abdominal viscera are also common. These injuries are discussed in detail below.

Hip Injuries

These injuries occur in and around the hip joint. They involve one or more of the following structures: acetabulum, femoral head, and proximal femur. In the adolescent, damage to the epiphysis in the proximal femur can occur, and in the aged injuries to the hip are the result of bone loss and decreased resilience. Hip injuries can be grouped under three main headings: avulsions of the proximal femoral epiphysis, traumatic hip dislocations, and fractures of the neck of the femur.

Avulsions of the Epiphysis

These injuries occur in adolescents due to powerful muscular exertions. The greater and lesser trochanter can be avulsed by muscular violence during athletic competition.

Traumatic Hip Dislocations and Fracture-Dislocations

These injuries are described below according to whether the dislocations are accompanied by injuries to the pelvis (acetabulum), femoral head, femoral neck, and femoral shaft.

The easiest way to dislocate a hip is to flex and adduct the hip and to apply a force along the femur rearward. In this position the hip joint is not supported by any bony structure. Automobile occupants who undergo a frontal impact with their legs crossed are liable to sustain this dislocation injury. If the hip is not fully adducted, fracture of the lip of the acetabulum along the back of the hip joint can occur together with a backward dislocation of the hip joint. Central dislocation with fracture of the acetabulum occurs when the hip is struck from the side, such as in a side-impact collision. The near-side occupant is struck by the door at the level of the greater trochanter, causing the femoral head to punch through the thin wall of the acetabulum. The mechanism of a central acetabular fracture due to a side impact to the greater trochanter has been reproduced in a series of cadaveric experiments by Pearson and Hargadon (1962), who used 56-lb (25.5-kg) and 150-lb (68.2-kg) pendulums to cause these fractures. Grattan and Hobbs (1969) reported that six out of seven cases of central acetabular fracture were due to side impact, while Eichenholtz and Stark (1964) asserted that the mechanism was a blow to the greater trochanter. This injury can also occur in the standing posture with the hip joint in a neutral position. A heavy blow to the shoulders or back can cause the hip to abduct, forcing the head of the femur into the acetabulum. If the hip is abducted, a rearward impact to the femur can cause an anterior (forward) dislocation. A rear-seat passenger in a frontal crash can be forced into the seat back with the thighs abducted and suffer such a dislocation (Beaupre, 1973). If the knee is wedged between the front seat back and the wall of the vehicle, separation of the sacroiliac joint can result (Markham, 1972).

In rearward dislocations of the hip joint involving fracture of the rim of the acetabulum, the femoral head can also be fractured. There is the prospect of developing a long-term disability in the form of traumatic arthritis. According to Armstrong (1948), the incidence is as high as 60%.

Hip dislocations accompanied by injuries to the femoral neck are rarely seen and occur in the elderly group. Necrosis of the head of the femur due to loss of blood supply (avascular necrosis) tends to result from these injuries. The mechanism of injury is a high tensile bending stress applied to the femoral neck causing it to fracture, as the joint is dislocated.

Dislocations with fracture of the femoral shaft are also rare and are easy to miss on x-rays (Schoenecker et al 1978). The mechanism of injury is unknown, but bending of the shaft is involved because the fractures tend to be transverse.

Fractures of the Neck of the Femur

There are two major classes of femoral neck fractures: intracapsular and extracapsular. The hip joint capsule, or the fibrous tissues enclosing the joint, extends from the pelvis to the intertrochanteric line of the femur. If a fracture occurs within the capsule, it can be a subcapital fracture of the head or a transcervical fracture of the neck. Extracapsular fractures can be either intertrochanteric or pertrochanteric.

Clinical Incidence of Pelvic Injuries

Six large series of clinical studies involving pelvic injuries are summarized. Three of the six were concerned with skeletal injuries, one with massive hemorrhage, and two with abdominal injuries.

Ryan (1971) surveyed 713 patients between 1956 and 1967 with fractures and fracture-dislocations of the pelvis. X-ray films of 387 patients admitted between 1958 and 1963 were studied in detail. A large majority (321 of the 387 patients) were traffic accident victims, including 154 pedestrians and a large group of 140 of unknown status. Only 20 were known to be occupants. The most common fracture was that of the pubic ramus (305), followed by the acetabulum (89), and the ilium (79). There were a total of 116 dislocations divided approximately equally among those of the hip, pubic symphysis, and sacroiliac joint.

Conolly and Hedberg (1969) reported on 200 patients with pelvic injuries. They classified them as major or minor. Injuries that involve the line of weight transmission and bilateral fracture of the pubic rami were considered major. A total of 109 patients sustained major injuries. The four major injuries were acetabular fracture (49), bilateral pubic rami fracture (34), multiple fracture of the hemipelvis (21), and separation of the pubic symphysis (5). Among the minor injuries, there were 82 cases of unilateral fracture of rami and an occasional case of isolated fracture of the ilium. In the third study, Kulowski (1962) reported on 145 cases of hip, femur, and knee injuries sustained by automotive occupants. There were a total of 184 lesions, with 23 patients having multiple injuries. Twelve cases of hip injury were associated with femoral shaft fractures or knee injuries, and there were only three cases of pelvic fractures.

Hauser and Perry (1966) looked at 196 patients with pelvic fractures and analyzed the role played by hemorrhage. About two-thirds of the patients were traffic accident victims. The total mortality rate was 19.4%. However, those with significant blood loss suffered a mortality rate of 30% (19 out of 63 patients). There was more blood loss in the more severely injured patients who required more transfusions. It was suggested that ligation of the hypogastric arteries saved three of four patients so treated.

There were two surveys of abdominal injuries. Levine and Crampton (1963) reported on a series of 425 patients with pelvic girdle fractures. Only 35 patients had severe or major abdominal injuries (8.2%). This is much lower than the 22% to 29% rate reported previously. Thirty-three of the 35 patients had traffic-related injuries. Fifteen types of complicating injuries were listed. They can be grouped under the following categories:

- Injuries to the bladder, urethra, and vagina with or without vascular injuries
- Injuries to the hard abdominal organs (liver, spleen, and kidney)
- Injuries to the small intestine, mesentery, and diaphragm
- Hemothorax
- Retroperitoneal hemorrhage
- Nerve injuries (frequently to the sciatic nerve)

This survey shows a low percentage of urinary tract injuries (3.5%) compared to 9% to 21% in other surveys. This may be due to the heavy concentration of cases related to automotive trauma. In the other survey of 1,309 cases by Moore (1966), there were only 26 cases of intraabdominal injuries not involving the urogenital organs. The paper was concerned with perforation of the large and small intestines and laceration of the mesentery. There were 11 patients with perforation of the intestines and not other injuries. Ten of the 26 patients had multiple intraabdominal injuries, including perforations and lacerations. No mechanisms of injury were proposed.

Pelvic Impact Response and Tolerance to Injury

Research on pelvic response to static and dynamic loads and associated injury thresholds can be categorized in terms of direction of loading: vertical, frontal, and lateral. Early work on pelvic response to vertical loading constituted a significant contribution to the literature, but, when viewed from current perspectives, it appears to be largely qualitative due to the lack of instrumentation at that time. Pelvic response to frontal loads applied via the femur was studied by several investigators. Although the data are rather sparse, it is possible to determine a response relationship between knee load and pelvic acceleration. In side impact, there are recent data that can be used to establish injury tolerance of the pelvis as a function of pelvic deformation, age, and body weight.

Vertical Loading

Data on pelvic response to vertical loads were provided by Evans and Lissner (1955) and Fasola et al (1955), who performed vertical loading tests on human pelves. In the former study, stress-coat lacquer was used to determine the areas of high tensile strain during the application of static and dynamic loads to the pelvis. Input energies from 3.7 to 12.8 N.m (33 to 113 in-lb) were applied to the ischial tuberosities of 22 isolated pelves, 16 of which were embalmed. Tensile strain patterns were found on both surfaces of the iliac wing, around the acetabulum, and on the pubic rami. In a second series of test, the entire body was dropped vertically onto the ischial tuberosities resulting in input energy levels of 22.6 to 50.9 N.m (200 to 450 in-lb). Fracture of the ischiopubic ramus occurred in a 79-year-old man's pelvis at 27.1 N.m (240 in-lb). The strain pattern was similar to that observed in the first series.

The study by Fasola et al (1955) involved the dropping of a weight onto the lumbar spine of cadaveric specimens. The lower portion of the pelvis, including the proximal femur, was embedded in cement to hold the pelvis upright. A force of 3.7 kN (830 lb) was necessary to cause a bilateral dislocation of the sacroiliac joint. The disjunction was due to avulsion of the bone at sites of ligamentous attachment near the joint. Static loading resulted in a fracture dislocation of the joint at a load of 3.5 kN (775 lb).

Frontal Loading

Evans and Lissner (1955) also carried out stress-coat studies on the pelvis, which was loaded frontally at the pubic symphysis, and identified regions of tensile strain. Fasola et al (1955) produced bilateral fracture of the superior and inferior pubic rami with a load of 2.7 kN (595 lb).

Sled tests to elicit the response of the knee-thigh-pelvis complex to frontally applied impact loads were carried out by Patrick et al (1966) using embalmed cadavers. Most of the study was concerned with the tolerance of the knee and femur, but there were data on the fracture loads of the proximal femur and the pelvis. The hip sustained fractures at loads ranging from 4.2 to 17.1 kN (950 to 3,850 lb) among the 10 cadavers tested. The load range for pelvic fractures was 6.2 to 11.8 kN (1,400 to 2,650 lb). The authors considered these load limits to be conservative because of the age of the cadavers. However, the wide range over a limited number of tests renders statistical analysis difficult. A normalization procedure for cadaver data needs to be formalized before these data can be used.

Melvin and Nusholtz (1980) performed six sled tests on unembalmed cadavers during which knee loads and pelvic accelerations were measured. Two knee load cells were used, but they were tied together by a rigid plate to minimize errors due to excessive bending. Impact load peaks varying from 8.9 to 25.6 kN (2,000 to 5,760 lb) resulted in hip and/or pelvic fractures. There were also femoral shaft fractures and injuries to the patella and condyles. There was only one case in which the hip or pelvis was not involved, and in a lightweight individual the loads generated were insufficient to cause any fracture (6.2 to 8.1 kN, 1,400 to 1,820 lb). Pelvic accelerations were indicative of load transfer to the pelvis, and abrupt changes in the acceleration signal were representative of pelvic fractures occurring. Acceleration traces were provided in the report by Melvin and Nusholtz (1980), but their peaks were not listed or correlated with injury. It is also interesting to note that the knee force curves showed a double peak if there was fracture of the femoral neck followed by that of the shaft. There was only a single peak if the shaft fractured first.

In another study by Nusholtz et al (1982), 37 knee impacts were carried out on 16 cadavers, using a pendulum impactor. The test subject was suspended in a restraint harness and impacted frontally at the knee or laterally at the level of the greater trochanter. Only the frontal impacts are discussed here. The impactor surface was either rigid or padded by three different types of foam: 25 mm (0.5 in) of Styrofoam or 25 to 100 mm (0.5 to 4 in) of Ensolite. Impactor velocity varied from 3.4 to 21.3 m/s (11.2 to 69.9 ft/s). Pelvic and trochanteric acceleration were monitored along with impactor force. No pelvic or hip fractures occurred at force levels as high as 37 kN (8,300 lb). Response data in the form of pelvic linear and angular acceleration were provided. However, acceleration data were difficult to analyze. For example, the magnitudes of the three components of angular acceleration of the pelvis were inconsistent in that the component about the spinal axis (z-axis) was not always predominant. Initial rotation of the pelvis appeared to be about the contralateral trochanter, but subsequent motion of the pelvis and femur were rather complex. Peak pelvic accelerations were less than those of the greater trochanter and were found to lag the trochanteric acceleration. Mechanical impedance was used as an analytical tool, and resonance was estimated to occur between 180 and 280 Hz. Impedance corridors were plotted for both the pelvis and the trochanter. However, because of the large scatter in the data, it was difficult to obtain average values of mass and spring rate for the pelvis.

Brun-Cassan et al (1982) conducted 10 whole-body impacts on unrestrained and unembalmed cadavers at collision speeds of 49.5 to 67.1 km/h (30.7 to 41.7 mph). Peak knee loads ranged from 3.7 to 11.4 kN (830 to 2,560 lb). There was only one fracture injury noted, that of the right patella and iliac crest at a knee load of 8.8 kN (1,980 lb). Finally, Doorly (1978) impacted isolated pelves with a drop weight causing acetabular and hip injuries. The impact force was not measured but was computed from energy considerations. The computed average force could not be correlated to the observed injuries.

Lateral Loading

Cadaveric research on lateral impact response of the pelvis was performed at Organisme National de Securité Routiere (ONSER), Association Peugeot-Renault (APR), University of Michigan Transportation Research Institute (UMTRI), the University of Heidelberg, and Wayne State University. Results from each laboratory are discussed separately without regard to chronology. Repeated impacts were carried out on ONSER test subjects, while the APR studies were single drop tests on a variety of surfaces. The UMTRI lateral tests were single impacts. At the University of Heidelberg, whole-body sled tests were conducted in the side-impact mode. There were two series of tests at Wayne State University pendulum impacts sponsored by General Motors (GM) and whole-body impacts sponsored by the Centers for Disease Control (CDC).

ONSER

Pendulum impacts on cadaveric subjects were reported by Ramet and Cesari (1979), Cesari et al (1980), and Cesari and Ramet (1982). The latest report was given by Cesari et al (1983), which contains all of the results reported since 1979. A total of 22 cadavers were used in this effort by ONSER. There were a total of 60 tests, using a 17.3-kg (38-lb) pendulum impactor with a 175-mm (6.9-in) diameter impact face having a spherical radius of curvature of 600 mm (23.6 in). The impact speeds ranged from 21 to 44.6 km/h (13.0 to 27.7 mph). Each cadaver sustained multiple (two to five) impacts that were administered at increasing velocities until fracture occurred. In 55 of these tests, a rigid impactor was used. The remaining tests were padded impacts. All impacts were aimed at the greater trochanter. Impact force and pelvic acceleration were measured. Peak force data as well as force and acceleration data, except for cumulative durations less than 3 ms (the so-called 3-ms clip), were provided.

Most of the injuries were multiple fractures of the pubic rami. In order of frequency of occurrence, these were followed by fractures of the proximal femur, dislocation of the sacroiliac joint, fractures of the iliac wing, and fractures of the acetabulum. The authors stated that this pattern of injury compared favorably with data from 14 accidents, the source of which was not given. In general, the experimental injuries were less severe because the tests were suspended fractures of the rami while standing subjects (pedestrians) are more prone to acetabular fractures. No reason was given for the observed difference in injury pattern. There were three femoral neck fractures out of the 22 cadavers tested. This is more frequent than that in the accident data, in which there was only one case of femoral neck fracture out of a total of 14 cases.

In terms of quantitative data, the results were separated into two groups by sex. For males, the force for fracture with the 3-ms clip, ranged from 4.9 to 11.9 kN (1,100 to 2,900 lb). The corresponding range for females was 4.4 to 8.2 kN (1,000 to 1,840 lb). The average force for an AIS-2 or -3 pelvic injury was 8.6 kN (1,930 lb) for males and 5.6 kN (1,260 lb) for females. The age range for males was 54 to 85 years, and that for females was 54 to 84. An index was introduced to correct for the body weight and height of the test subjects. It was termed the Livi index, which is the ratio of the cube root of the body weight in kilograms and 10 times the standing height in meters. The Livi index is used to obtain a corrected body weight. The Livi index for a 50th percentile male is 23.5. Thus, the correction factor is the ratio of 23.5 to the computed Livi index of the test subject. The corrected weight of the subject is the product of the correction factor and its actual weight. By plotting the fracture-producing force at 3 ms against actual body weight, the correlation coefficient was found to be 0.75. It increased to 0.89 if the corrected weight was used. The least squares line for impact force as a function of corrected body weight is given by:

$$\text{Impact Force} = 193.85\ W_c - 4{,}710.6$$

where W_c is the corrected body weight. For a 50th percentile male weighing 75kg (165lb), the impact force for fracture is 9.8kN (2,210lb), which is higher than that of the average of the male cadaver data. The measured acceleration values with the 3-ms clip were less than 100g. The authors defended their multiple impact procedure vigorously, stating that the force-time histories were consistent in pattern for multiple impacts and that the impact force increased consistently with increasing impactor speed.

The effect of using a padded impactor was studied in five tests on three cadavers. Two cadavers sustained AIS-3 injuries while one had AIS-2 injuries. The force required to cause fracture was about the same as that for a rigid impactor, but the speed of the impactor could be increased by 40%.

Following dynamic testing, the nonimpacted hemipelvis was instrumented with nine strain gates and subjected to a static lateral load. High strains were measured in the pubic rami followed by strains in the ischiopubic rami. Strains were low in the acetabulum and ilium. Section moduli of the iliopubic rami were measured and plotted against the impact force at fracture. The resulting least squares line had a correlation coefficient of 0.96.

The proposed tolerance force with the 3-ms clip is 10kN (2,250lb) for a 75-kg (165-lb) person. This reduces to 4.6kN (1,030lb) for a 5th percentile female. The authors recommended that side-impact dummies should be fitted with a lateral force transducer. This transducer should be located in the anterior region of the pelvis, such as near the pubic rami where many injuries occur.

In a later paper presenting the development of an improved ATD pelvis for lateral impact, Cesari et al (1984) discussed the use of the ATD pelvic acceleration with the 3-ms clip as an injury criterion. They concluded that injury does not correlate with impactor maximum force as well as it does with maximum pelvic acceleration ($r = .891$ versus $r = .986$). However, their pelvis design was also shown to produce higher accelerations than those obtained with cadavers. A linear regression analysis of cadaver maximum pelvic lateral acceleration versus impactor force produced a correlation coefficient of 0.77.

APR

The bulk of the APR data are contained in reports by Fayon et al (1977) and Tarriere et al (1979). The latter contains all of the side-impact data acquired at APR via a series of 26 cadaveric drop tests. The subjects ranged in age from 25 to 71 years and in weight from 41 to 75kg (90 to 165lb). Each subject was dropped only once, but there were a variety of test configurations and impact surfaces for the thorax as well as the pelvis. The subject was suspended with cables and dropped on its side onto a load-measuring plate. The arm was either at a 45-degree angle or was pulled out of the way during thoracic impact. The impact surfaces were rigid planes or padded surfaces. Some supported the shoulder at impact while others did not. When energy-absorbing pads were used, the thorax and pelvis impacted separate blocks. The thoracic impact load was measured by a load cell. Pelvic response was given in terms of pelvic acceleration, measured at the sacrum in the midsagittal plane, 90mm (3.5in) distal to the iliac crests. The condition of the bone was characterized by its ash content given as a percentage of wet weight (C/M) and ash content in 10mm (0.4in) of rib given in g/cm (C/L). Undamaged ribs were also subjected to standard bending and shear tests, where C is the weight of the ash content, M is the weight and L is the length of rib used.

The drop height ranged from 0.5 to 3m (20 to 118in). Fractures of the pubic rami were observed in four subjects. There were multiple injuries in one subject who sustained fractures of the iliac wing and acetabulum in addition to the fractures of the rami. In the other 22 subjects, there were no pelvic or hip injuries. There were no femoral neck fractures. If data from a 68-year-old female subject are ignored, the lowest fracture level is 50g with the 3-ms clip and the highest is 90g with the 3-ms clip. The proposed tolerance level is 80 to 90g with the

3-ms clip. Note that this is pelvic acceleration and not that of the impactor.

UMTRI

Published UMTRI data consist of 12 tests reported by Nusholtz et al (1982) using a flat pendulum impactor. Most of the impacts were aimed at the greater trochanter of a cadaver suspended in a restraint harness. Impact force and pelvic acceleration were measured. The peak force ranged from 3.2 to 14 kN (720 to 3,150 lb), and the peak linear acceleration varied from 38 to 135 *g* for impact speeds of 18.4 to 31.0 km/h (11.4–19.2 mph). Six of the 12 cadavers sustained pelvic fractures. Four of them had fractures of the pubic rami, and one sustained fractures to the ilium. These injuries did not occur at high force or acceleration levels. The absence of femoral neck fractures was again noted. The lack of correspondence between observed injuries and measured data was attributed to variation among subjects. It was also concluded that these data were not comparable to those of ONSER, since the test setup and impactor configuration were all different.

University of Heidelberg

Marcus et al (1983) presented data from 11 cadaver tests carried out at the University of Heidelberg. They were whole-body side-impact tests in which the thoracic and pelvic impact surfaces were instrumented with load cells to measure contact force. Eight of the 11 tests were rigid wall impacts at 24, 32, and 40 km/h (15, 20, and 25 mph). The other three were padded impacts at 32 km/h (20 mph), using foam developed by APR. Subjects ranged in age from 17 to 61 years and in weight from 50 to 99 kg (110 to 218 lb). The body weight was corrected in the same manner as that proposed by Cesari and Ramet (1982), using the Livi index. The pelvic impact force was then plotted against corrected body weight. Among the 11 data points, there were three pelvic injuries with an AIS rating of 2 or 3, but the type of injury was not disclosed. The range of pelvic forces at 3 ms was 3.6 to 28.9 kN (800 to 6,500 lb). It was concluded from this study that the tolerance limits proposed by Cesari and Ramet may be too conservative. It was also determined that 28% of the inertial force of impact was transmitted via the pelvis.

Wayne State University (GM Tests)

Viano et al (1989) performed a series of 14 pendulum impact tests on unembalmed cadavers at Wayne State University. The mass of the 150-mm diameter pendulum was 23.4 kg and its speed varied from 16.2 to 33.8 km/h (10.0–21.0 mph). The impact location was the greater trochanter. Lateral pelvic response was reported in the form of force-deflection curves, as shown in Figure 18.13. A variety of parameters were used to determine their suitability as a measure of tolerance. It was found that peak pelvic acceleration and pelvic deformation were not reliable measures. However, the ratio of pelvic deformation to pelvic width (% compression) was found to correlate well with pubic rami fracture, which was the only type of injury observed. The tolerance level for 25% probability of serious injury to the pelvis was found to be 27% of pelvic compression, based on the entire width of the pelvis.

Wayne State University (CDC Tests)

A total of 17 whole-body side-impact tests were run, using a modified University of Heidelberg setup in which the lateral impact force at the shoulder, thorax, abdomen, and pelvis were measured separately. It should also be noted that the transmission of impact force to the pelvis was through the greater trochanter, as the pelvic load cell plate was only 100 mm high and was designed to miss the wing of the ilium. Cavanaugh et al (1990) have already reported on data from the first 12 tests. The percent compression of the pelvis was again found to be the best measure of tolerance, based on a logist analysis of the data. For a 25% probability of fracture, the tolerance is 32.6% of the struck side half width. Peak impact force and pelvic acceleration did not perform well as injury indicators. Pubic rami injuries were observed but again, there were no femoral neck fractures.

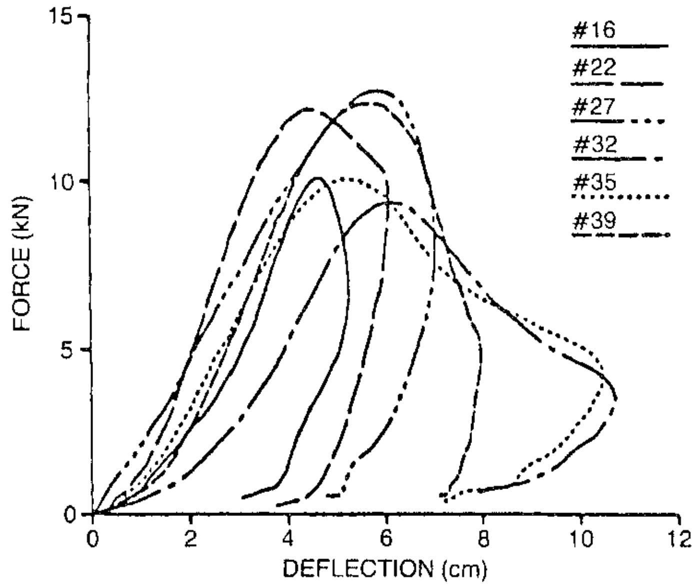


FIGURE 18.13. Force-deflection response of the pelvis during lateral impact. (From Viano et al, 1989.)

Pelvic Response to Frontal and Lateral Impacts

If pelvic response is defined as an acceleration output at the pelvis for a force input to either the knee, for frontal response, or the greater trochanter, for lateral response, the only published data that can provide this information are found in Nusholtz et al (1982). For frontal response, the peak force and acceleration data in Table 7 of Nusholtz et al (1982) were analyzed by linear regression techniques. (Run 80L098 was considered to be an "outlier," and the data from this run were not used in the regression analysis.) For all 15 data points, including both rigid and padded knee impacts, the correlation coefficient is 0.78. If only rigid impacts are used, the result is a correlation coefficient of 0.80, a value similar to that obtained by Cesari et al (1984) for lateral impacts. There is thus not a strong relationship between pelvic acceleration and knee loads.

A similar analysis was performed for all available lateral impact data provided by Nusholtz et al (1982). A meaningful linear regression line could not be drawn for the lateral impact data. Its slope was almost zero, and the correlation coefficient was less than 0.1. The low correlation of maximum pelvic acceleration and peak input load was attributed to the highly variable response of the system to the point of load application and pelvis-leg orientation at the time of impact. However, in tests in which pelvic deformation was measured, there was excellent correlation of this variable with pelvic AIS, as was shown by Viano et al (1989) and by Cavanaugh et al (1990).

Summary and Conclusions

As a major load-bearing structure between the upper torso and the lower extremities, the pelvis plays an important role in controlling body kinematics. Injuries to the pelvis contribute only about 1% to the total IPR. Lateral response of the pelvis has been studied for both impactor and flat-wall impacts and has been described in terms of force-time histories and pelvic acceleration-time histories. Lateral loading tolerances for the pelvis are available in terms of peak deformation, force, and acceleration limits, with the deformation limit showing the best correlation to injury severity, as quantified by the pelvic AIS.

The infrequency of femoral neck fractures due to side impacts to the greater trochanter is a surprising finding. It has been hypothesized that the elderly sustain such injuries when they fall laterally and impact the greater trochanter. More research is necessary to determine if femoral neck fractures are spontaneous and that the fall is the result of the fracture and not the cause.

References

Adams MA, Hutton WC. Prolapsed intervertebral disc. A hyperflexion injury. Spine 1982;7:184–191.

Allen ME, Wein-Jones I, Motink DR, et al. Acceleration perturbations of daily living. A comparison to "whiplash", 1994, Spine 19:1285–1290.

Anderson PA, Rivara FP, Maier RV. The epidemiology of seat belt associated injuries. J Trauma 1991;31:60–67.

Armstrong JR. Traumatic dislocation of the hip joint. J Bone Joint Surg 1948;30B:430–445.

Avramov AI, Cavanaugh JM, Ozaktay AC, King AI. Effects of controlled mechanical loading on group III and IV afferents from the lumbar facet joint: an in vitro study. Proceedings of the Eighteenth Annual Meeting of the Society for the Study of the Lumbar Spine, Heidelberg, University of Heidelberg, pp. 23–24, 1991.

Beaupre A. Trochanteric fractures. Proceedings of the Tenth Traffic Injury Research Foundation of Canada Annual Meeting, TIRF, Ottawa, pp. 15–18, 1973.

Begeman PC, King AI, Levine RS, Viano DC. Biodynamic response of the musculoskeletal system to impact acceleration. In: Proceedings of the Twenty-fourth Stapp Car Crash Conference, SAE paper no. 801312, Warrendale, PA, Society of Automotive Engineers, 1980.

Begeman PC, King AI, Prasad P. Spinal loads resulting from $-g_x$ acceleration. In: Proceedings of the Seventeenth Stapp Car Crash Conference, SAE paper no. 730977, Warrendale, PA, Society of Automotive Engineers, 1973.

Belytschko T, Privitzer E. A three-dimensional discrete element dynamic model of the spine head and torso. In: von Gierke HE, ed. *Models and analogues for the evaluation of human biodynamic response, performance and protection*, AGARD conf. proc. no. 253. Paris Technical Editing and Reproduction Ltd., London, pp. A9–1 to A9–15, 1978.

Bogduk N, Twomey LT. *Clinical anatomy of the lumbar spine*. Churchill Livingstone, Melbourne, 1987.

Brandser EA, El-Khoury GT. Thoracic and lumbar spine trauma. Radiol Clin North Am 1997;35:533–557.

Brinckmann P. Injury of the annulus fibrosus and disc protrusions. An in vitro investigation on human lumbar discs. Spine 1986;11:149–153.

Brown T, Hansen RJ, Yorra AJ. Some mechanical tests on the lumbosacral spine with particular references to the intervertebral discs. J Bone Joint Surg 1937;39A:1135–1164.

Brun-Cassan F, Leung YC, Tarriere C, Fayon A, Patel A, Got C, Hureau J. Determination of knee-femur-pelvis tolerance from the simulation of car frontal impacts. Proceedings of the Seventh International Conference on the Biomechanics of Impacts, IRCOBI, pp. 101–115, 1982.

Buckwalter JA. The fine structure of human intervertebral disc. In: White AA, Gordon SL, eds. *Idiopathic low back pain*. Mosby, St. Louis, pp. 108–143, 1982.

Cavanaugh JM, Ozaktay AC, Yamashita T, King AI. Lumbar facet pain: biomechanics, neuroanatomy and neurophysiology. J Biomech 1996;29:1117–1129.

Cavanaugh JM, Walilko TJ, Malhotra A, Zhu Y, King AI. Biomechanical response and injury tolerance of the pelvis in twelve sled side impact tests. Proceedings of the Thirty-fourth Stapp Car Crash Conference, SAE paper no. 902307, Warrendale, PA, Society of Automotive Engineers, 1990.

Cesari D, Bouquet R, Zac R. A new pelvis design for the European side impact dummy. Proceedings of the Twenty-eighth Stapp Car Crash Conference, SAE paper no. 841650, Warrendale, PA, Society of Automotive Engineers, 1984.

Cesari D, Ramet M. Pelvic tolerance and protection criteria in side impact. Proceedings of the Twenty-sixth Stapp Car Crash Conference, SAE paper no. 821159, Warrendale, PA, Society of Automotive Engineers, 1982.

Cesari D, Ramet M, Bouquet R. Tolerance of human pelvis to fracture and proposed pelvic protection criterion to be measured on side impact dummies. Proceedings of the Ninth International Technical Conference on Experimental Safety Vehicles, Washington, DC, U.S. Department of Transportation, National Highway Traffic Safety Administration, pp. 261–269, 1983.

Cesari D, Ramet M, Clair PY. Evaluation of pelvic fracture tolerance in side impact. Proceedings of the Twenty-sixth Stapp Car Crash Conference, SAE paper no. 801306, Warrendale, PA, Society of Automotive Engineers, 1980.

Chance GO. Note on a type of flexion fracture of the spine. Br J Radiol 1948;21:452–453.

Chen CY, Cavanaugh JM, Ozaktay AC, Kallakuri S, King AI. The effects of phospholipase A_2 on nerve root structure and function. Spine 1997;22:1057–1064.

Conolly WB, Hedberg EA. Observations on fractures of the pelvis. J Trauma 1969;9:104–111.

Doorly TPG. Forces imposed on the hip-joint in car collisions. J Traffic Med 1978;6:44–46.

Eiband AM. Human tolerance to rapidly applied accelerations: a summary of the literature. NASA Memorandum No 5-19-59E, 1959.

Eichenholtz SN, Stark RM. Central acetabular fractures a review of thirty-five cases. J Bone Joint Surg 1964;46:695–714.

El-Bohy AA, Yang KH, King AI. Experimental verification of facet load transmission by direct measurement of facet/lamina contact pressure. J Biomech 1989;22:931–941.

Evans FG, Lissner HR. Studies on pelvic deformations and fractures. Anat Rec 1955;121:141–165.

Ewing CL, King AI, Prasad P. Structural considerations of the human vertebral column under $+g_z$ impact acceleration. J Aircraft 1972;9:84–90.

Farfan HF, Cossette JW, Robertson GH, Wells RV, Kraus H. The effects of torsion on the lumbar intervertebral joints: the role of torsion in the production of disc degeneration. J Bone Joint Surg 1970;52A:468–497.

Fasola AF, Baker RC, Hitchcock FA. Anatomical and physiological effects of rapid decelerations. WADC-TR-54-218. Wright-Patterson AFB, Ohio, 1955.

Fayon A, Tarriere C, Walfisch G, Got C, Patel A. Contributions to defining the human tolerance to perpendicular side impact. Proceedings Third International Conference on Impact Trauma, IRCOBI, pp. 297–309, 1977.

Giles LGF, Taylor JR. Innervation of lumbar zygapophysial joint synovial folds. Acta Orthop Scand 1987;58:43–46.

Gordon SJ, Yang KH, Mayer PJ, Mace AH, Kish VL, Radin EL. Mechanism of disc rupture—a preliminary report. Spine 1991;16:450–456.

Grattan E, Hobbs JA. Injuries to hip joint in car occupants. Br Med J 1969;11:71–73.

Gray H. *Anatomy of the human body*. Lea & Febiger, Philadelphia, 1973.

Hakim NS, King AI. Programmed replication of in situ (whole-body) loading conditions during in vitro (substructure) testing of a vertebral column segment. J Biomech 1976;9:629–632.

Hakim NS, King AI. A three-dimensional finite element dynamic response analysis of a vertebra with experimental verification. J Biomech 1979;12:277–292.

Hardy WG, Lissner HR, Webster JE, Gurdjian ES. Repeated loading tests of the lumbar spine: a preliminary report. Surg Forum 1959;9:690–695.

Hauser CW, Perry JF Jr. Massive hemorrhage from pelvic fractures. Minn Med 1966;49:285–290.

Henzel JH, Mohr GC, von Gierke HE. Reappraisal of biodynamic implications of human ejections. Aerospace Med 1968;39:231–240.

Hess JL, Lombard CF. Theoretical investigations of dynamic response of man to high vertical accelerations, Aviat Med 1958;29:66–75.

Kazarian LE. Injuries to the human spinal column: biomechanics and injury classification. Exerc Sports Sci Rev 1982;9:297–352.

Kazarian LE, Beers K, Hernandez J. Spinal injuries in the F/FB-111 crew escape system. Aviat Space Environ Med 1979;50:948–957.

Kazarian LE, Boyd D, von Gierke H. The dynamic biomechanical nature of spinal fractures and articular facet derangement. AGARD publication CP-88-71, paper no. 19, 1971.

King AI, Chou CC. Mathematical modelling, simulation and experimental testing of biomechanical system crash response. J Biomech 1976;9:301–317.

King AI, Vulcan AP, Cheng LK. Effects of bending on the vertebral column of the seated human during caudocephalad acceleration. In: Proceedings of the Twenty-first Annual Conference on Engineering in Medicine and Biology, Houston, TX, p. 32, 1968.

King AI, Yamashita T, Ozaktay AC, Cavanaugh JM. A morphological study of the lumbar facet joint capsule and its innervation. Proceedings of the Sixth International Conference on Biomedical Engineering, Singapore, National University of Singapore, pp. 32–35, 1990.

King AI, Yang KH. Biomechanics of the lumbar spine. In: Schmid-Schonbein G, ed. *Frontiers in Applied Mechanics and Biomechanics*. Springer-Verlag, New York, pp. 210–224, 1985.

Kulowski J. Interconnected motorist injuries of the hip, femoral shaft and knee. Proceedings of the Fifth Stapp Automotive Crash and Field Demonstration Conference, University of Minnesota, Minneapolis, pp. 105–124, 1962.

Latham FA. A study in body ballistics: seat ejection. Proc R Soc Lond [Biol] 1957;147:121–139.

Levine JI, Crampton RS. Major abdominal injuries associated with pelvic fractures. Surg Gynecol Obstet 1963;11:223–226.

Liu YK, Goel VK, Dejong A, Njus G, Nishiyama K, Buckwalter J. Torsional fatigue of the lumbar intervertebral joints. Spine 1985;10:894–900.

Liu YK, Njus G, Buckwalter J, Wakano K. Fatigue response of lumbar intervertebral joints under axial cyclic loading. Spine 1983;8:857–865.

Marchand F, Ahmed AM. Investigation of the laminate structure of lumbar disc annulus fibrosus. Spine 1990;15:402–410.

Marcus JH, Morgan RM, Eppinger RH, Kallieris D, Mattern R, Schmidt G. Human response to and injury from lateral impact. Proceedings of the Twenty-seventh Stapp Car Crash Conference, SAE paper no. 831634, Warrendale, PA, Society of Automotive Engineers, 1983.

Markham DE. Anterior-dislocation of the hip and diastatis of the contralateral sacroiliac joint: the rear-seat passenger's injury? Br J Surg 1972;59: 296–298.

Melvin J, Nusholtz G. Tolerance and response of the knee-femur-pelvis complex to axial impacts. UM-HSRI-80-27. University of Michigan, Highway Safety Research Institute, Ann Arbor, 1980.

Miniaci A, McLaren AC. Anterolateral compression fracture of the thoracolumbar spine. A seat belt injury. Clin Orthop 1989;240:153–156.

Mooney V, Robertson J. The facet syndrome. Clin Orthop 1976;115:149–156.

Moore JR. Pelvic fractures: associated intestinal and mesenteric lesions. J Surg 1966;9:253–261.

Nachemson A. Intradiscal measurements of pH in patients with lumbar rhizopathies. Acta Orthop Scand 1969;40:23–42.

Nakamura SI, Takahashi K, Takahashi Y, Yamagata M, Moriya H. The afferent pathways of discogenic low back pain.: Evaluation of L2 spinal nerve infiltration. J Bone Joint Surg 1996;78B:606–612.

Nicoll EA. Fractures of the dorso-lumbar spine. J Bone Joint Surg 1949;31B:376–393.

Nusholtz G, Alem NM, Melvin JW. Impact response and injury to the pelvis. Proceedings of the Twenty-sixth Stapp Car Crash Conference, SAE paper no. 821160, Warrendale, PA, Society of Automotive Engineers, 1982.

Nyquist GW, Murton CJ. Static bending response of the human lower torso. Proceedings of the Nineteenth Stapp Car Crash Conference, SAE paper no. 751158, Warrendale, PA, Society of Automotive Engineers, 1975.

Orsay E, Dunne M, Trumbull T, et al. Prospective study of the effects of safety belts in motor vehicle crashes. Ann Emerg Med 1990;19:258–261.

Oxland TR. Burst fractures of the human thoracolumbar spine: a biomechanical investigation. Ph.D. thesis, Yale University, 1992.

Ozaktay AC, Cavanaugh JM, Blagoev DC, Getchell TV, King AI. Effects of a carrageenan-induced inflammation in rabbit lumbar facet joint capsule and adjacent tissues. Neurosci Res 1994;20:355–364.

Ozaktay AV, Kallakuri S, Cavanaugh JM, King AI. Phospholipase A_2-induced electrophysiologic and histologic changes in rabbit dorsal lumbar spine tissues. Spine 1995;20:2659–2668.

Patrick LM, Kroell CK, Mertz HJ Jr. Forces on the human body in simulated crashes. Proceedings of the Ninth Stapp Car Crash Conference, Minneapolis, University of Minnesota, pp. 237–259, 1966.

Patwardhan A, Vanderby R Jr, Lorenz M. Load bearing characteristics of lumbar facets in axial compression. In: Thibault L, ed. Advances in bioengineering. ASME, Phoenix, pp. 155–160, 1982.

Peacock A. Observations on the postnatal structure of the intervertebral disc in man. J Anat 1952;86: 162–179.

Pearson JR, Hargadon EJ. Fractures of the pelvis involving the floor of the acetabulum. J Bone Joint Surg 1962;44B:550–561.

Pontius UR, Liu YK. Neuromuscular cervical spine model for whiplash. In: *Mathematical modeling biodynamic response to impact*, SAE paper no. 760770, SAE publication no. SP-412, Warrendale, PA, Society of Automotive Engineers, 1976.

Porter RS, Zhao N. Patterns of injury in belted and unbelted individuals presenting to a trauma center after motor vehicle crash: seat belt syndrome revisited. Ann Emerg Med 1998;32:318–424.

Prasad P, King AI. An experimentally validated dynamic model of the spine. J Appl Mech 1974;41: 545–550.

Prasad P, King AI, Ewing CL. The role of articular facets during $+g_z$ accelerations. J Appl Mech 1974;41:321–326.

Ramet M, Cesari D. Experimental study of pelvis tolerance in lateral impact. Proceedings of the Fourth International Conference on the Biomechanics of Trauma, IRCOBI, pp. 243–249, 1979.

Roaf R. A study of the mechanics of spinal injuries. J Bone Joint Surg 1960;42B:810–823.

Ruff S. Brief acceleration: less than one second. In: *German aviation medicine, World War II*. U.S. Government Printing Office, Washington, DC, 1:584–597, 1950.

Ryan P. Traffic injuries of the pelvis at St. Vincent's Hospital, Melbourne. Med J Aust 1971;1:475–479.

Schoenecker PL, Manske PR, Sertl GO. Traumatic hip dislocation with ipsilateral femoral shaft fractures. Clin Orthop Rel Res 1978;130:233–238.

Shirazi-Adl A. Three-dimensional nonlinear finite element stress analysis of a lumbar intervertebral joint. PhD Thesis, Montreal, McGill University, 1984.

Shirazi-Adl A, Drouin G. Load bearing role of facets in a lumbar segment under sagittal plane loadings. J Biomech 1987;20:601–613.

Smith WS, Kaufer H. Lumbar seat belt fracture. In: Selzer ML, Gikas PW, Huelke DF, eds. *The prevention of highway injury*. University of Michigan, Ann Arbor, MI, p. 218, 1967.

States JD, Annechiarico RP, Good RG, et al. A time comparison study of the New York State safety belt use law utilizing hospital admission and police accident report information. Proceedings Thirty-third Annual Conference of Association for the Advancement of Automotive Medicine, Des Plaines, IL, AAAM, pp. 265–281, 1989.

Steckler RM, Epstein JA, Epstein BS. Seat belt trauma to the lumbar spine: an unusual manifestation of the seat belt syndrome. J Trauma 1969;9:508–513.

Stokes IAF. Surface strain on human intervertebral discs. J Orthop Res 1987;5:348–355.

Tarriere C, Walfisch G, Fayon A, et al. Synthesis of human tolerance obtained from lateral impact simulations. Proceedings of the Seventh International Technical Conference on Experimental Safety Vehicles, U.S. Department of Transportation, National Highway Traffic Safety Administration, Washington, DC, pp. 359–373, 1979.

Tennyson SA, King AI. A biodynamic model of the human spinal column. Transactions of SAE, paper no. 760771, 1976.

Tennyson SA, King AI. Mathematical models of the spine. In: Avala XJR, ed. *Proceedings of the First International Conference on Mathematical Modeling*, vol II. Rolla, MO, pp. 977–985, 1977.

Viano DC, Lau IV, Asbury C, King AI, Begeman P. Biomechanics of the human chest, abdomen, and pelvis in lateral impact. Proceedings of the Thirty-third Annual Conference of Association for the Advancement of Automotive Medicine, Des Plaines, IL, AAAM, pp. 367–382, 1989.

Watson-Jones R. *Fractures and joint injuries*. London, Churchill-Livingstone, 1976.

Wiggishoff CC, Kiefer JH. Urethral injury associated with pelvic fracture. J Trauma 1968;8:1042–1048.

Yamashita T, Cavanaugh JM, El-Bohy AA, Getchell TV, King AI. Mechanosensitive afferent units in the lumbar facet joint. J Bone Joint Surg 1990;72A:865–870.

Yang KH, Byrd AJ III, Kish VL, Radin EL. Annulus fibrosus tears—an experimental model. Orthop Trans 1988;12:86–87.

Yang KH, King AI. Mechanism of facet load transmission as a hypothesis for low back pain. Spine 1984;9:557–565.

Yang KH, Shen K-L, Demetropoulos CK, et al. The relationship between loading conditions and fracture patterns of the proximal femur. J Biomech Eng 1996;118:575–578.