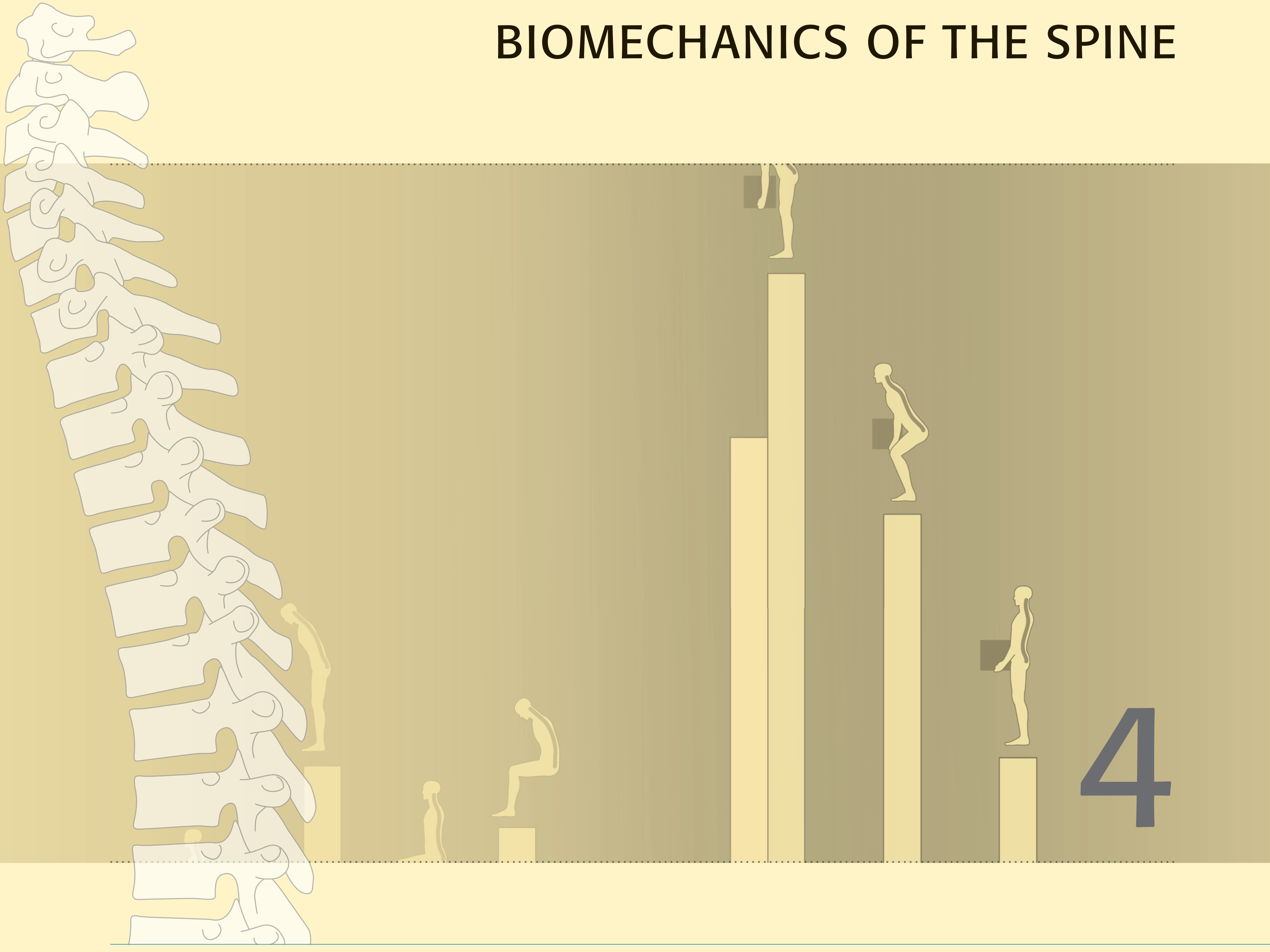
Thomas Steffen

BIOMECHANICS OF THE SPINE



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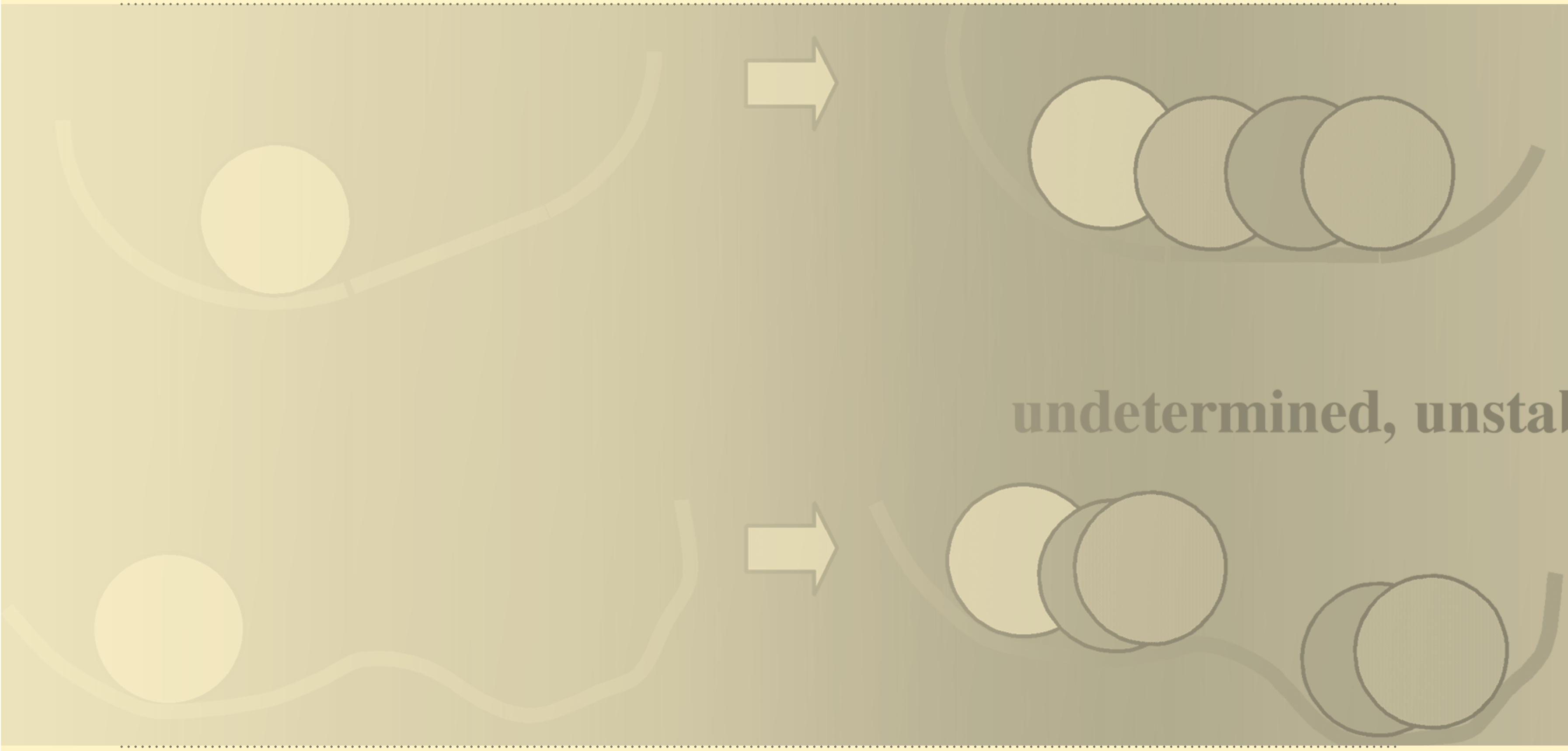
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4 BIOMECHANICS OF THE SPINE

INTRODUCTION

This chapter provides basic anatomical and biomechanical knowledge of the spine. Anatomy uses descriptive language to teach physical structure and biomechanics applies engineering methods to the analysis of human motion. This chapter will also explore what biomechanics can teach us about spinal function. Figuring out the relationship between structure and function is the key to grasping how the healthy spine functions properly. Knowledge of this relationship is also important when trying to rationalize why a specific type of instrumentation used to alter the spine's mechanical behavior is believed to ultimately help the patient.

Understanding the anatomy and biomechanics of the spine helps physicians determine the likely source of a patient's spinal complaint. The diagnostic workup of a spine patient is often challenging—due mostly to the complexity of the spinal anatomy, the multisourced generation of pain, and the sheer number of joints involved. The physician needs to have deep and thorough knowledge to be able to narrow a patient's complaint down to a specific problem, which is often mechanical in nature.

It is like everything else in life: We will see only what we are looking for, we will only find what we already know, and we will only have a grasp of what we understand.

It is the author's goal to provide the reader with a certain level of biomechanical knowledge so that they can become experts in diagnosing spinal problems. Using this knowledge and combining it with a history and physical examination can guide clinicians in determining the likely cause of a patient's spinal complaints and the ways in which it can be treated.

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1 INTRODUCTION

The spine is essentially a curved stack of 33 vertebrae, which can be divided based on structural differences into five distinct regions: cervical (7 vertebrae), thoracic (12 vertebrae), lumbar (5 vertebrae), sacral (5 fused vertebrae), and coccygeal (4 fused vertebrae). An obvious difference between the distinct regions is the curvature in the sagittal plane. The thoracic and sacral regions of the spine feature a kyphotic curvature. Kyphotic curvatures are considered primary because they already exist at birth. Later, to allow the growing child an upright posture, a secondary lordotic curvature develops in the cervical and lumbar regions.

Individual vertebrae, including all interposed structures, show reasonable similarities along the entire spine. This is mostly true, except for the upper cervical spine which has an adapted anatomy to allow for larger head movements, and for the sacral and coccygeal regions in which mobility was largely lost due to aberrant discs. The remaining spine, reaching from C3 to S1, is often referred to and looked upon, for diagnostic and therapy-related decisions, as being composed of individual motion segments. The subsequent sections describe the individual structural components of such motion segments, their interplay with neighboring components, und how this ultimately enables the spine to fulfill a complex function.

The primary purpose of the entire spine is to provide axial support for the head and trunk, while allowing for bending and twisting movements, and to protect neural structures encased in a bony canal running from the head to the sacrum. Not unique to the spine, but nevertheless significant, is its contribution to blood cell formation through a noteworthy total mass of bone marrow.

2 MOTION SEGMENT

2.1 VERTEBRA

Vertebral anatomy—regional differences

A vertebra is composed of the vertebral body and posterior elements, which include the paired pedicles, superior and inferior articular processes with interposed intraarticular mass, the lamina, transverse processes, and the singular spinous process.

Vertebrae are primarily composed of cancellous bone, an anisotropic viscoelastic material. Fortunately, for non-injurious values of strain and over a wide range of strain rates, cancellous bone behaves elastically. The vertebral body has an approximated cylindrical shape. A thin shell of increased density trabecular bone surrounds a core of cancellous bone. Posterior elements are made from true cortical and cancellous bone.

Regional differences between vertebrae are obvious:

- Typical cervical vertebrae have a small broad body, a large triangular canal, laterally directed pedicles lying just anterior to the transverse foramen, and medially directed laminae ending in a bifid spinous process. The seventh cervical vertebra (C7), with its long spinous process palpable through the skin, is known as the vertebra prominens. The lowest portion of the nuchal ligament attaches to this spinous process.
- Thoracic vertebrae have a triangular cross section at the cranial levels, but they gradually become more circular at the caudal levels. Most vertebrae feature paired superior and inferior demifacets for the rib head articulation. The canal is small relative to the body and is circular in outline, but there are no transverse foramina in the thoracic transverse processes. The typical spinous processes are long, straight and narrow. They overlap like roof tiles. Transverse processes are equally prominent and articulate with the tubercle of the ribs. The superior articular processes are vertical, flat, and face backward and laterally.
- Lumbar vertebrae have a large size vertebral body. The vertebral foramen is triangular, and the spinous process is hatchet-shaped and blunt. The superior articular surfaces are vertical, curved, and face backward and medially inward. The posterior rim features a mammillary process.
- The sacrum is typically formed from five fused sacral vertebrae and is triangular in shape. It has a superior, posterior, anterior, and lateral surface. The anterior surface is curved. The superior surface is formed by the superior end plate of the first sacral vertebra. Vertebral bodies are much wider transversely than anteroposteriorly. The S1 superior articular processes are concave and directed posteromedially, to be congruent with the L5 lower articular processes.
- The lower end of the sacrum is often fused with the coccyx, the four lowest vertebrae which are small and rudimentary. The coccygeal vertebrae, of course, form the tail in animals, with the aberrant human version of it gently reminding us of our origins. The coccyx can be a common source of postpartum pain, but also as a result of falling on the buttocks.

How is axial load being carried?

Between 70% and 90% of static axial load is carried by the cancellous vertebral body. The role of the shell and core in providing mechanical strength varies with age. They carry, modulated by sagittal posture variations, the remaining axial load. Processes serve as lever arms to provide mechanical advantage for muscles inserting along their surfaces.

Vertebrae are loaded in series. Caudal vertebrae must support a greater share of the body weight and this accounts for an increasing cross-sectional area of the vertebral bodies. In healthy adults the bone density remains reasonably constant throughout the entire spine. Ultimate failure of the individual vertebral bodies similarly increases from cranial to caudal vertebrae (**Table 4.1-1**).

Spinal region	C3–7	T1–6	T7/8	T9–12	L1–5
Strength (N)	1,600	2,000	2,300	3,600	5,600

Table 4.1-1
Approximate ultimate compressive strength values of different vertebral bodies for a nonosteoporotic adult male.

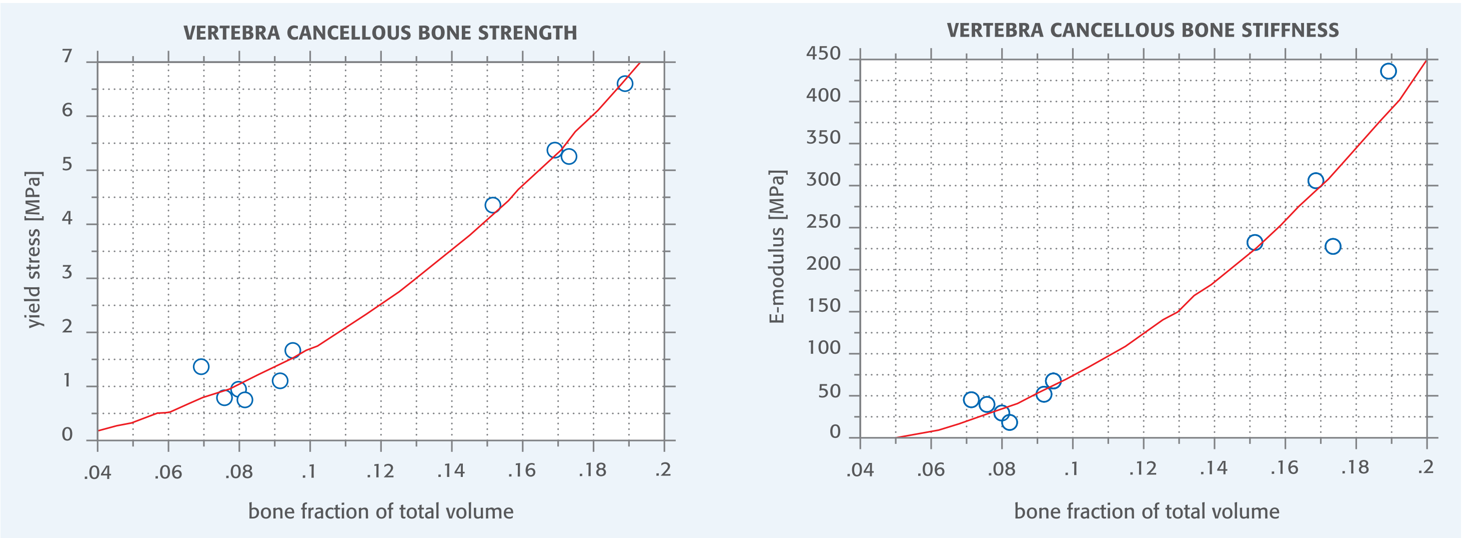


Fig 4.1-1
Compressive strength (yield stress, left graph) and stiffness (Young modulus, right graph) of vertebral cancellous bone as a function of bone density (2nd order polynomial function curve fit). The data were recorded from 12.5 mm diameter cancellous bone plugs compressed at a 0.1% strain/sec rate. Fractional bone volume was measured from bone samples of the same vertebral body (mirrored site) using the Archimedes principle.

Normal vertebral cancellous bone has a bone density of about 15%. Axial compressive material properties for the cancellous bone of a normal vertebral body (ie, 15–18% fractional bone volume) are estimated around 5 MPa yield stress and around 300 MPa elastic modulus (**Fig 4.1-1**). The strength and elastic modulus (Young modulus) of cancellous bone are roughly dependent on its density to the second power. That is, a 25% decrease in density may result in a 50% decrease in strength. The dense shell takes a greater share of the load as the cancellous core density is gradually lost due to osteoporosis. In severe osteoporotic conditions, bone density can be drastically reduced, maybe as low as one third of its original density. Not surprisingly, the overall load carrying capacity of osteoporotic vertebrae can then be reduced by almost a magnitude. From initially being fairly uniform, with progressive loss of bone density, regional differences in trabecular bone strength become more obvious. Trabecular bone strength decreases from the anterior to the posterior and from the medial to the lateral. Thanks to the bony end plate, the axial load is more uniformly distributed across the cancellous bone cross section. The strongest part of the bony end plate is the peripheral epiphyseal ring, which makes this region best suited to resist localized axial loads. Intervertebral spacers (cages) best resist subsidence when they are seated on this epiphyseal ring [1].

2.2 FACET JOINTS

Facet orientation and joint loads

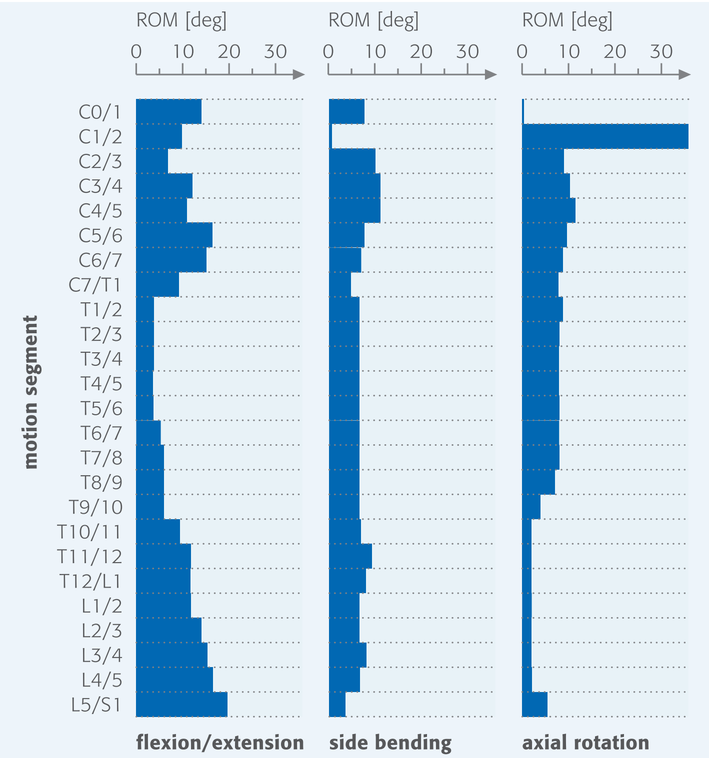
The paired facet joints form, along with the intervertebral disc, the intervertebral joint. Whereas the disc is a fibrocartilaginous syndesmosis, the facet joints are diarthrotic joints with sliding cartilaginous surfaces lubricated with synovial fluid. Facet joints channel and limit the range of motion in anteroposterior shear and axial rotation directions (**Fig 4.1-2**), which is an important and differentiating aspect of spinal regions.

- Cervical spine: facet orientation is roof-tile shaped, coupling lateral bending and axial rotation motions in an opposite direction (bending the head to the left results automatically in an axial rotation to the right).
- Thoracic spine: coronal plane orientation with a slight inward tilt in the transversal plane, permitting easy axial rotation movements with the center of rotation projected into the vertebral body.
- Lumbar spine: sagittal plane orientation of facet joint surfaces, effectively blocking axial rotation movements.

Facet joints carry, in an upright standing posture, between 10% and 20% of the axial body load. In hyperextension, the joint load increases up to 30%. In a flexed posture, the facet joints carry up to 50% of the anterior shear load (compressive loads transmitted by surface contact and tensile loads resisted by joint capsule). Facet joint capsules are highly innervated and have been shown to be a source of low back pain.

Facet joint tropism

Facet joint tropism is defined as asymmetry in the facet joint angles, with one joint having a more coronal orientation than the other. It has been reported [2] that the incidence of tropism in patients with degenerative disc disease is higher than in the normal population. When tropism is present, the segment tends to rotate toward the more oblique facet when axial loads are applied. Rotation resulting from the joint asymmetry can place additional torsional stress on the anulus fibrosus, thus, possibly contributing to intervertebral disc injury.



2.3 INTERVERTEBRAL DISC

Disc anatomy

The nucleus pulposus is located approximately in the disc's center. Through a transitional zone the disc's appearance gradually changes toward the periphery, with concentric annular fiber layers making up its outer border. The nucleus, particularly with age, is difficult to delimit. It covers an estimated 30%–50% of the cross-sectional area of the total disc. The healthy nucleus contains almost exclusively type II collagen fibers in an aqueous gel rich in proteoglycans. The latter attracts water, leading to natural swelling of the nucleus. Water content in the normal nucleus decreases from about 90% of its total volume during the first year of life to around 70% at old age. Water is gradually replaced with a nondirectional fibrous matrix, associated with an overall loss in tissue elasticity and an increase in stiffness.

The end plates are composed of a dense layer of trabecular bone, further covered with a layer of hyaline cartilage (**Fig 4.1-3**). Vascular channels within the vertebral bodies have been observed to run directly along the end plates, representing the predominant nutrient source for the adult disc cells. Some blood vessels approach the annulus at the periphery but do not penetrate the disc. A healthy intervertebral disc is the human body's largest avascular structure. The cartilaginous end plates

Fig 4.1-2
Ranges of motion (ROM) for each spinal motion segment of the entire spine. Values are given separately for flexion/extension, lateral bending, and axial rotation. Significant differences are apparent for the cervical, thoracic, and lumbar spine. These differences are largely the result of distinct regional facet joint orientations.

undergo progressive calcification with age, which impedes nutrition and contributes to a progressive degeneration of the disc throughout adulthood.

The anulus fibrosus is composed of concentric layers of collagen fiber bundles. The fiber orientations alternate from layer to layer, with the fibers generally oriented at an angle of approximately 30° with respect to the horizontal plane and in any two adjacent layers at 120° with respect to each other. The fiber orientation from outer to the inner annulus gradually becomes more horizontal. The anterior anulus fibrosus band is thickest, the posterolateral and, most significantly, the posterior anulus fibrosus band is thinnest. The number of distinct fiber layers varies from over twenty in the anterior regions to less than twelve in the posterior regions. The individual fiber layers are in the order of 50–300 µm thick, with the outer layers generally being smaller.

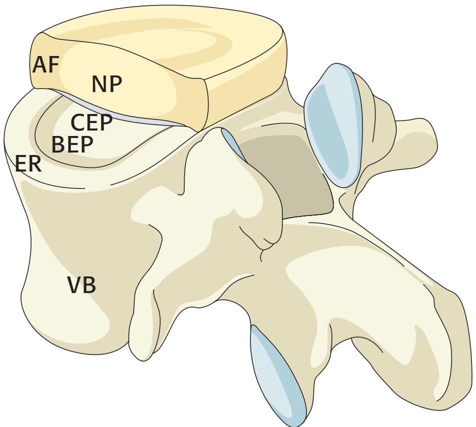


Fig 4.1-3
Vertebral body (VB) with bony end plate (BEP) and cartilaginous end plate (CEP). The intervertebral disc is comprised of the layers of the anulus fibrosus (AF) and the central nucleus pulposus (NP). Sharpey fibers insert from the anulus fibrosus directly into the epiphyseal ring (ER).

The fibers are almost exclusively of type I collagen for the outer annular portions, but they gradually change to a 40% type I and 60% type II fiber ratio for the inner portions. With degeneration, type I collagen fibers are replaced with type II fibers.

Many fiber layers are discontinuous. This fact may be responsible for interlamellar stress peaks predisposing the annulus to fail with the formation of circumferential or radial tears. SEM (scanning electron microscopy) imaging has shown the fibers in the inner third of the annulus to interconnect loosely with the cartilaginous end plate; the fibers in the outer portion are firmly bonded to the epiphyseal ring of the bony vertebral end plate. Thus, the inner annulus is most prone to initial mechanical failure.

Seemingly, intervertebral disc morphology predisposes injury through sites of high stress. Disc cells are not as readily serviced with nutrients as other tissues in the body, a critical factor that only gets worse with age. Degeneration and/or injury decrease the functional ability of the disc to distribute axial forces through hydrostatic pressure. Finally, degenerative changes are accompanied by an ingrowth of nerve fibers into the outer annular regions, sensing and transmitting pain.

Anulus fibrosus is made to resist hoop stresses

The intervertebral disc is an inhomogeneous, anisotropic, porous, and nonlinearly viscoelastic structure. Mechanical characterization of discs can either be performed on isolated anulus fibrosus or nucleus pulposus material, or on intact whole discs. Since the annulus is physiologically loaded in tension (at least for the nondegenerated disc), its tensile properties are best documented. Test data on whole discs reflect the predominant compression loading to which they are subjected in vivo and their exhibited viscoelastic behavior.

Quasistatic tensile modulus and failure strength have been determined for small rectangular anulus fibrosus specimens. The modulus of the outer annulus is greater in the circumferential direction compared to the vertical direction (approximately 3–4 MPa compared to 0.5 MPa for circumferential and vertical directions, respectively). The anterior annulus consistently reveals a larger tensile modulus than the posterior annulus, regardless of depth and loading direction. This implies a potential weakness of the postero-lateral annulus, making it more prone to bulging or protrusion of disc material. Degenerated discs have lower moduli compared to nondegenerated ones. Tensile failure strength of the annulus shows a similar trend, with the maximum values of healthy discs being around 5–10 MPa for loads applied in a circum-ferential direction.

In a healthy disc, with the annular fibers cyclically loaded in circumferential directions, the endurance limit for such hoop stresses is around 1.5 MPa, but it can drop quickly for degener-ated discs. The annulus is weakest in a radial direction, with tensile strength values consistently below 0.5 MPa. The an-nulus is poorly designed to resist tensile radial forces which tend to separate the laminar layers. When the annular band is being compressed on the side of bending (eg, anteriorly dur-ing flexion), inner fiber layers are bulging inward and outer layers are bulging outward, in effect separating annular fiber layers (**Fig 4.1-4**).

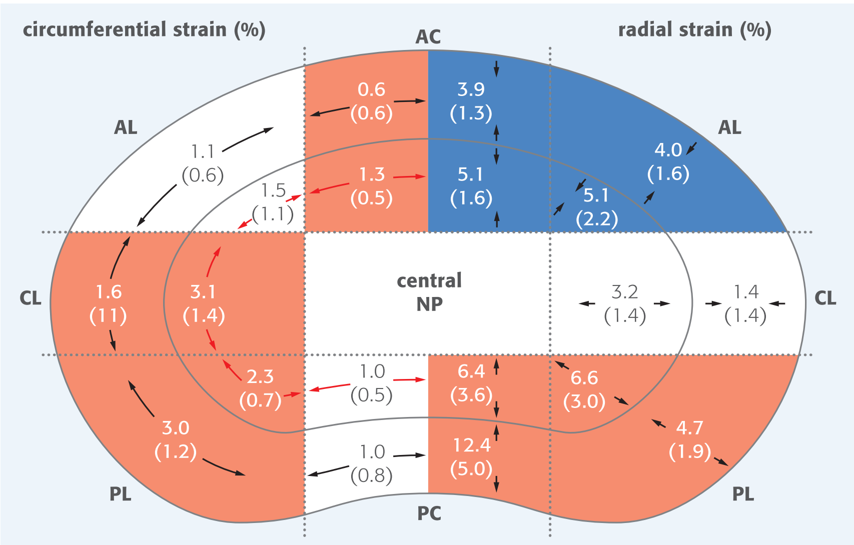


Fig 4.1-4
Transversal disc maps [3] showing internal disc strain during compression in extended posture, measured for the posterocentral (PC), posterolateral (PL), centro-lateral (CL), anterolateral (AL), and anterocentral (AC) annular regions. All data are from normal discs (Thompson grade I or II). The left side displays strain (expressed in %, with standard deviations in brackets) recorded in a circumferential direction, the right side in a radial direction. Red areas were identified to demon-strate significant strain in a tensile mode, blue areas in a compressive mode. Whereas the circumferential tensile strain in the annular regions is expected, the tensile radial strains in the PL and PC regions are not. Tensile radial strains were even amplified in degenerated discs. They might be responsible for the progressive annular fiber delamination seen in degenerated discs.

Nucleus pulposus pressure varies with external loads

Due to the high proteogycan content of the disc matrix, the nucleus pulposus (NP) has a base swelling pressure of about 0.1–0.3 MPa [4]. When a normal disc is cut through the center, the NP is immediately protruding from the cutting surface. When the disc is put in a saline solution the NP matrix will continue to swell.

Besides this base pressure, the NP pressure measured at its center is greatly modulated by external trunk loads and paraspinal muscle tension balancing those external loads. Nachemson [5] and later Wilke [6] both measured intradiscal pressure in healthy subjects using an invasive method (**Fig 4.1-5**). Moderate activities such as walking or stair climbing, compared to upright standing, may already double the NP pressure. Carrying a 20 kg load, depending on the technique used, can increase the NP pressure by a factor of four.

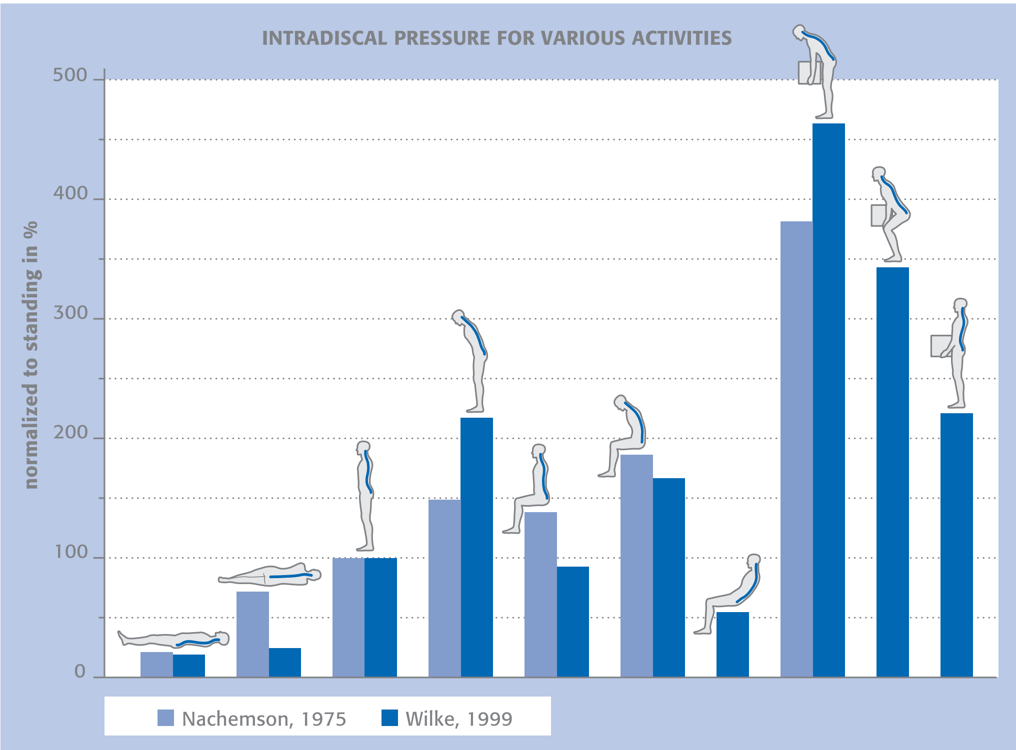


Fig 4.1-5
Intradiscal pressure measurements using a percutaneous needle placed in the center of the disc [5, 6]. Pressures are normalized to the average pressure recorded in an upright standing position. The pressure may rise four to five times for moderate physical activity such as the lifting of a 20 kg weight. Knee and lower back positions play a major role.

McNally and Adams [7] introduced a new technique called "stress profilometry" that allowed them to quantify pressure within the nucleus and annulus under axial load application. A strain-gauged sensor mounted to a needle was passed incrementally through the disc along a straight path (mid-sagittal or mid-coronal plane). Measurements were obtained also for flexion, extension, and side bending. The authors found abnormal stress concentrations in the posterior annulus [8], which suggests a predisposition for the prolapse of an intervertebral disc (IVD) at this location. This stress peak in the posterior annulus was found to increase even further after minor damage to the trabecular arcades was induced by supramaximal compression to the vertebra.

The author [9] measured intradiscal pressures using three needles with a total of nine strain-gauged pressure sensors mounted on them. The needles were placed in the anterior, as well as in the left and right posterolateral disc regions, with the sensors positioned in the intermediate disc zone between nucleus and annulus. "Pressure maps" were recorded for axial loads applied in the flexed or extended postures, combined with axial rotation (**Fig 4.1-6**). The largest pressure increase was found in the posterolateral regions during flexion, predominantly on the side of axial rotation (eg, left posterolateral region for left axial rotation). It seems that the disc's posterolateral regions are subjected to even higher stress peaks when axial rotation is added to the loading modality.

Volume shifts in the nucleus pulposus matrix

The water content of the matrix is variable and represents an equilibrium between two opposing pressures: mechanical, which dehydrates the gel-like matrix; and swelling of the hydrophilic proteoglycans, which causes the matrix to absorb fluid. Changes in the load applied to the motion segment will disturb the equilibrium and subsequently cause a net outward fluid flow, until a new balance is reached. Net fluid flow can occur within the disc (eg, from the anterior to the posterior areas) or from the disc to the outside. Fluid exchange between the disc and the surrounding tissue occurs through both the peri-annular route [10] and through the end-plate route [11, 12].

In this scope, diurnal changes in the intervertebral disc height can easily be explained. At night, in a supine position, reduced axial load acting on the disc allows relatively unopposed swelling of the disc's matrix. During the daytime, in a predominantly upright position, the absorbed fluid is again expelled from the disc.

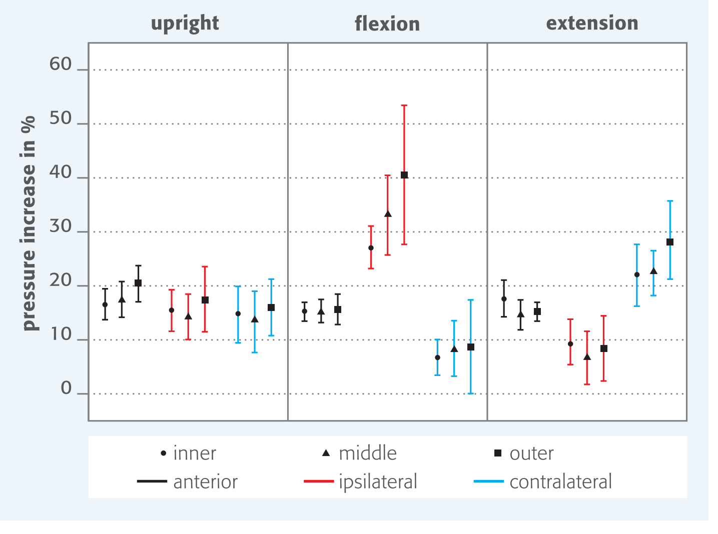
Apart from circadian rhythms modulating the IVD volume, postural load changes can be responsible for an internal fluid shift [13]. Because the IVD's permeability is very low [14], a considerable amount of fluid shift can only be achieved by large changes in postural loads acting over long periods of time. Maintaining a specific posture over an extended period (as might be the case in a workplace) will produce fluid shifts within the IVD. These shifts in turn can influence the spine's mechanical behavior, and, more importantly, increase the disc's vulnerability to localized mechanical overload [15].

Disc mechanical failure

The intradiscal pressure in a healthy individual is proportional to the compressive load applied to the motion segment. The maximum pressure is about 1.5 times the applied force, divided by the disc's transverse cross-sectional area. Because of this proportionality, the disc pressure (in a healthy disc) can be used to estimate compressive loading of the spine. Pressure causes the end plate to bulge about 0.5 mm toward the vertebral body [16]. Excessive disc load, seen mostly during flexion and while carrying loads, may fracture the end plate's central region.

A degenerated or aged disc undergoes gradual changes. Common structural changes include loss of nucleus pulposus volume, disc height loss, radial fissures, circumferential clefts and rim tears in the annulus, inward bulging of the inner annulus, increased radial (outward) bulging of the annulus, reduced disc height, and possible end-plate defects with disc material herniated into the adjacent vertebral body. A healthy disc contains a relatively soft, hydrated hydraulic nucleus pulposus, which distributes load and stress evenly between vertebrae. The degenerated disc with a diminished hydrostatic region exhibits high stress concentrations in the posterior annular regions [3].

Disc degeneration also affects other elements of the motion segment. Through disc height loss and apparent changes in axial load transmission, the facet joints become incongruent and may become mechanically overloaded. They gradually develop arthritic changes as is the case with all diarthrotic joints. Also, loss of disc height will initially result in ligament



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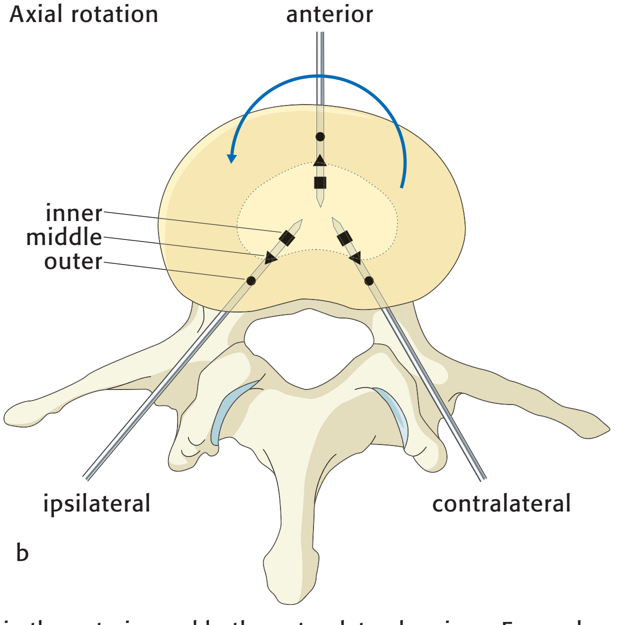

b

Fig 4.1-6a–b

Intradiscal pressure increase measured in a multisegmental cadaveric lumbar spinal test setup. Pressures were recorded at 10 Nm with a left-sided axial rotation load, recorded either in a neutral, flexed, or extended posture [9]. Three needles were placed in the anterior and both posterolateral regions. For each needle three sensors were spread out across the disc's transitional zone (b). The highest pressure increase was seen in the posterolateral region, facing the side of axial rotation (ie, ipsilateral).

laxity, hypermobility, and a loss in segmental stiffness. Only much later, with almost complete collapse of the disc space and the formation of bridging osteophytes, the segmental mobility will be reduced and the stiffness increased. Ultimately, the segment will spontaneously fuse.

As disc structure deteriorates so does disc function. Annular fibers progressively fail through fissuring. The degrading nucleus pulposus no longer transfers pressure across the disc's center. Consequently, the annulus which is usually loaded in tension due to hoop stress, no longer fails mechanically in tension, because it becomes primarily loaded in compression. In healthy discs, the underlying end plate will fail due to compressive loading before the disc does, but this is no longer the case in the degenerated disc. The remaining nuclear material is pressed against the annulus, this causes bulging and ultimately a disc prolapse (herniation). Disc prolapses occur mostly in the cervical and lower lumbar spine. Bulging or herniated discs may compress nerve roots or the thecal sac.

Gross structural disruption certainly appears to represent mechanical failure, but tissue composition is usually altered and it is not clear whether the disc structure is weakened by biochemical changes or whether those changes represent a response to mechanical failure. Further cadaveric experiments have attempted to link mechanical factors to disc degeneration by showing that disc prolapse and radial fissures can be simulated in apparently normal discs if the loading is sufficiently severe [17, 18]. Animal experiments show that biological degeneration always occurs after minimal structural damage was induced by means of a scalpel blade stab incision into the annulus [19, 20]. There is considerable interest in identifying biochemical and metabolic abnormalities in degenerated disc tissues, but these abnormalities may be the consequences of disc failure rather than the cause.

2.4 MUSCLES

Along with ligaments, muscles initiate and guide spinal movements. Larger muscles are important in balancing external forces. Small segmental muscles, too small in their cross section to produce relevant force, are densely packed with muscle spindles and therefore believed to be important for proprioception of the spine.

The spinal musculature may be divided, based on location, into six major groups:

- Posterior spinal muscles (erector spinae, multifidus, lumbar part of the longissimus thoracis).
- Anterior spinal muscles (psoas major, quadratus lumborum).
- Small segmental muscles (interspinales, intertransversarii).
- Respiratory or intercostal muscles.
- Abdominal wall muscles (intertransversarii, internal and external oblique, rectus abdominis).
- Superficial trunk muscles (broad muscles including the rhomboids, latissimus dorsi, pectoralis, trapezius, transversus abdominis).

The greatest risk of injury occurs during maximum muscle tension. For example, when tripping and during the subsequent fall, a reflex triggers the forcibly and involuntarily lengthened erector spinae muscle to contract. Following deceleration of the forward movement, the erector spinae muscle will eventually accelerate the spine back into an extended posture. Muscle injury most likely occurs during such forcible lengthening while the muscle is maximally activated (ie, eccentric muscle contraction).

2.5 LIGAMENTS

Spinal ligaments, like most soft tissues of the body, are visco-elastic in nature with nonlinear elastic responses. Ligaments connect between adjacent vertebrae (intersegmental), but individual fibers may well stretch over multiple levels. They transmit tensile loads only and, as a result, they specifically limit excessive motion.

Intersegmental ligaments are:

- Anterior and posterior longitudinal ligaments
- Yellow ligament (ligamentum flavum)
- Interspinous and supraspinous ligaments
- Intertransverse ligaments
- Facet joint capsules

Spinal ligaments do not enjoy the same margins of fail safety as bones do, as they physiologically may operate under conditions relatively close to their failure strength. Ligaments mechanically fail at approximately 10 to 20 MPa of tensile stress (translates to approximately 180 N failure load for the posterior longitudinal and 340 N failure load for the anterior longitudinal ligament). For ultimate bending postures, strains in ligaments farthest from the axis of rotation can reach 20%. Hormonal concentrations can affect ligament laxity. For instance, pregnancy systemically increases ligament laxity.

2.6 SPINAL CORD, NERVE ROOTS

The spinal cord runs down the foramen magnum of the skull base to the sacrum, protected in a bone delimited canal formed by the stacked neural arches and the vertebral body posterior walls. Paired nerve roots exit from between vertebrae through the intervertebral foramen. The dura further protects neural structures that are floating inside this sheet in cerebrospinal fluid. Vertebral fractures, disc ruptures, or bony impingements can all potentially affect neural performance, resulting in pain and/or paralysis.

3 SPINAL MOTION

Kinematics is the study of the motion of bodies, regardless of the cause. Kinetics or dynamics, on the other hand, are concerned with the effects of forces on the motion of objects. Motion patterns of the entire spine are complex. Again, to simplify matters, motion is meaningfully and best described for isolated motion segments. Nonlinear motion behavior, as for most soft tissues, is characteristic for all regions of the spine. The spine also exhibits viscoelastic behavior, due to viscoelastic qualities of its tissue constituents. The hysteresis, expressed as deviate paths of the load-displacement curve for the forward and reverse directions, is a direct result of viscoelasticity.

Segmental flexibility

The spinal motion segment is, compared to other joints in the body, relatively unconstrained, exhibiting relevant motion in all six degrees of freedom (DOF). Motion can be adequately described by specifying the angular (rotational) and linear (translational) relative displacements for three orthogonal axes (**Table 4.1-2**).

Rotational or angular displacement description		Translational or linear displacement description	
Plane	**Rotation in plane**	**Axis out of plane**	**Translation along axis**
Sagittal	Flexion/ extension	Transverse	Left/right lateral shear
Coronal	Left/right side bending	Frontal	Anterior/ posterior shear
Transverse	Left/right axial rotation	Axial	Compression/ distraction

Table 4.1-2
Nomenclature for angular and linear displacements for the spine.

Forces (expressed in Newton=N) lead to linear displacements; moments (expressed in Newton × meter=Nm) lead to angular displacements. Forces and moments are both called loads. The segmental flexibility for each degree of freedom can be characterized with a load-displacement relationship. Three parameters have been particularly effective in characterizing the typically nonlinear load-displacement relation (**Fig 4.1-7**) of motion segments: neutral zone (NZ), elastic zone (EZ), and range of motion (ROM). For the biomechanical testing of implant performance, the three parameters are often compared between test groups.

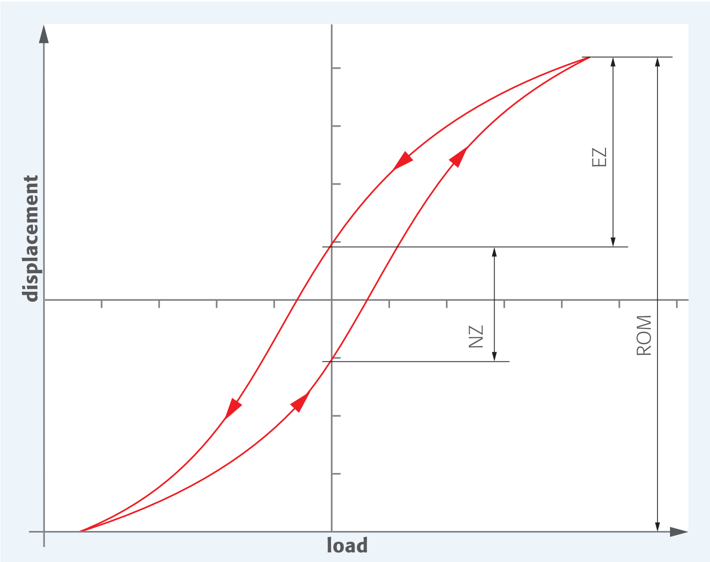

Fig 4.1-7
Schematic sample for typical segmental load-displacement curve. The forward and reverse paths show a hysteresis. Neutral zone (NZ), elastic zone (EZ), and range of motion (ROM) are indicated with double arrows.

If a load is applied to a motion segment (or similarly to the entire spine), the segment quickly displaces from a neutral position to a position where an appreciable resistance is first encountered. This initial displacement is called NZ, comparable to the “toe-in region” generally seen in soft-tissue elastic responses. Within the NZ, movements of a few degrees or millimeters are observed with practically no muscular effort. An extended NZ is indicative of abnormal joint laxity, also known as clinical instability. When moving beyond the NZ, stiffening of the motion segment is encountered. Loads and resulting displacements are largely proportional. This region is called EZ, a stiffness value is given by the slope of the curve. Eventually, respecting physiological magnitudes of load, a maximum displacement is reached. The span covered between minimum and maximum loads is called ROM.

Kinematic description of multidirectional motion

Motion rarely involves a single degree of freedom only. Typically, the spine exhibits complex motions when external loads change. Complex motions are simultaneous displacements in multiple degrees of freedom (translation and rotation). Complex motions, however, are still not unpredictable, and are sometimes referred to as a motion pattern. It describes a typical “motion path” a vertebral body follows under external load changes (eg, a person sitting up from a chair or climbing stairs).

The instantaneous center of rotation (ICR) is an easy kinematic notation useful in describing complex motion. Any motion in a plane (eg, in the sagittal plane for flexion/extension) can always be expressed as an angular displacement around a center point. This point, denoted for some instant in time (ie, for an infinitesimally small displacement) is called the ICR. All ICR points connected for a larger movement form the centrode path or area.

For a healthy motion segment, the centrode path, even for full ROM segmental movements in the principal directions, is confined to a relatively small area usually overlaying the inferior vertebral body. In the degenerated or unstable spine [21], the centrode area can enlarge dramatically (**Fig 4.1-8**). Despite

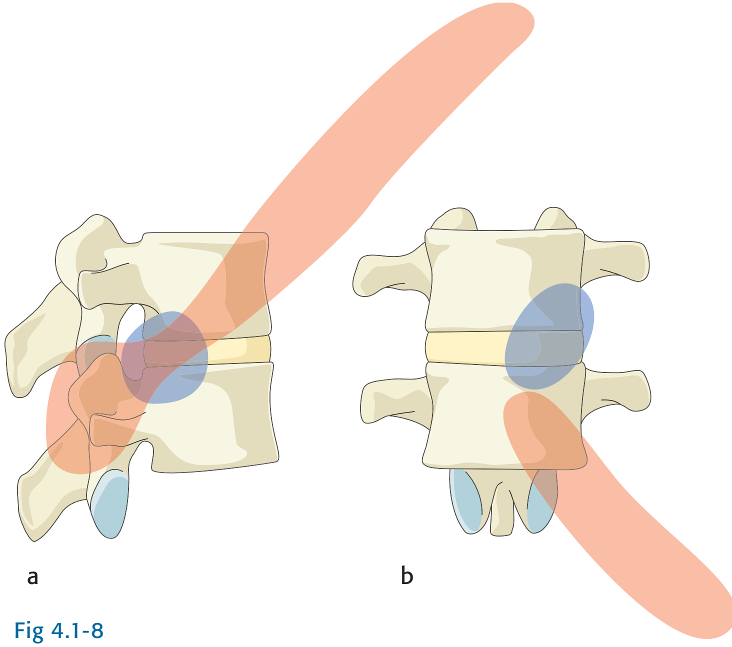

Fig 4.1-8
Instantaneous center of rotation paths (centrode areas) for a specimen of the lumbar spine in lateral (a) and anteroposterior projection (b). The red area represents a healthy motion segment. The blue area depicts the much larger centrode for a degenerative motion segment. Larger centrode areas are indicative of segmental instability.

the ICR and centrode path being a very effective criterion in identifying abnormal segmental motion, clinicians so far have not adopted it. The wide error margin associated with tracking skeletal landmarks in subsequent radiographic views seriously impedes the reconstruction of the centrode path with reasonable fidelity.

More complex kinematic descriptions to numerically analyze motion of a rigid body in 3-D space, either use Cardanian or Eulerian sequences, direction cosign matrices, or helical axis descriptions. Complex 3-D kinematic descriptions are mostly used for body animation or 3-D visualization of joint kinematics.

Coupled motion

Coupling refers to motion about or along axes secondary to those of the axis of applied load (**Fig 4.1-9**) [22]. For example, in the middle and lower cervical spine, left lateral bending produces a concomitant left axial rotation due to the orientation of the articulating surfaces of the facets. Coupling in the lumbar spine is more complex. In the normal spine left lateral bending causes right axial rotation in the upper lumbar segments, but it causes left axial rotation in the lumbosacral joint.

The L4/5 segment constitutes a transitional level [23]. In kinematic measurements conducted by the author for asymptomatic individuals walking on a treadmill, most subjects had axial rotation and lateral bending movements in the same direction, but about one third of the subjects were in opposite directions (**Fig 4.1-10**) [22]. Seemingly, coupled motions have considerable variability between subjects, but they may perhaps be altered as a result of structural changes leading to increased or decreased segmental laxity, or causing segmental instability. It is also possible that coupled motion patterns may be due to specific contractile patterns of intrinsic and extrinsic paraspinal muscle groups.

Segmental instability

Segmental instability has been described in many different ways. White and Panjabi [24] described it as "the loss of the ability of the spine under physiological loads to maintain its pattern of displacement so that there is no initial or additional neurological deficit, no major deformity, and no incapacitating pain".

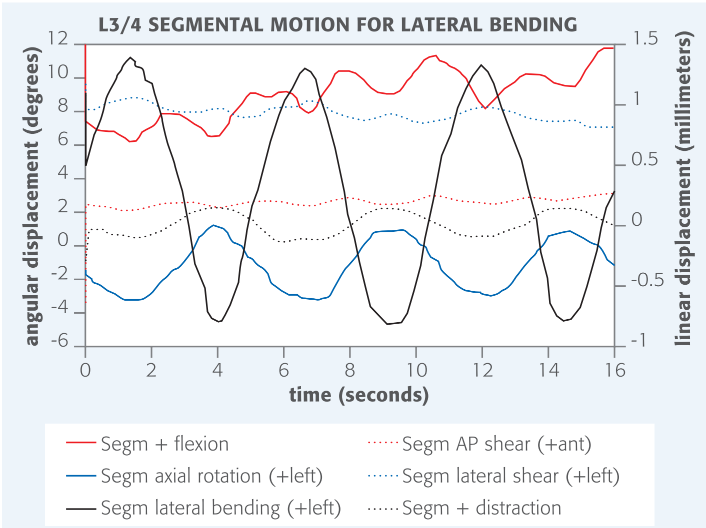

Fig 4.1-9
L3/4 segmental kinematics analyzed for three cycles of voluntary lateral bending (full range) recorded in a healthy subject. Invasive measurements were obtained with a 3-D electromagnetic motion tracking system affixed to the spinous processes [22]. Solid lines represent angular displacement, dashed lines linear displacement. Coupled motion can be seen in axial rotation—and to a lesser degree in the other motion directions.

Spinal stability refers to a state in which there is adequate control and support between two adjacent vertebral segments in the spine. This is accomplished by both, passive structures (ie, disc, ligaments, joint capsules) and active structures (ie, muscles). When there is a breakdown in passive structures, usually resulting in pain, it is essential that the muscles (active structures) are trained to compensate. This can best be accomplished with abdominal muscle training, spinal stabilization exercises administered by a physiotherapist, or by similar means.

Segmental instability might be caused by congenital anomalies such as spondylolisthesis/-lysis, but it also could be the result of an acquired disease, such as the early to mid stages of

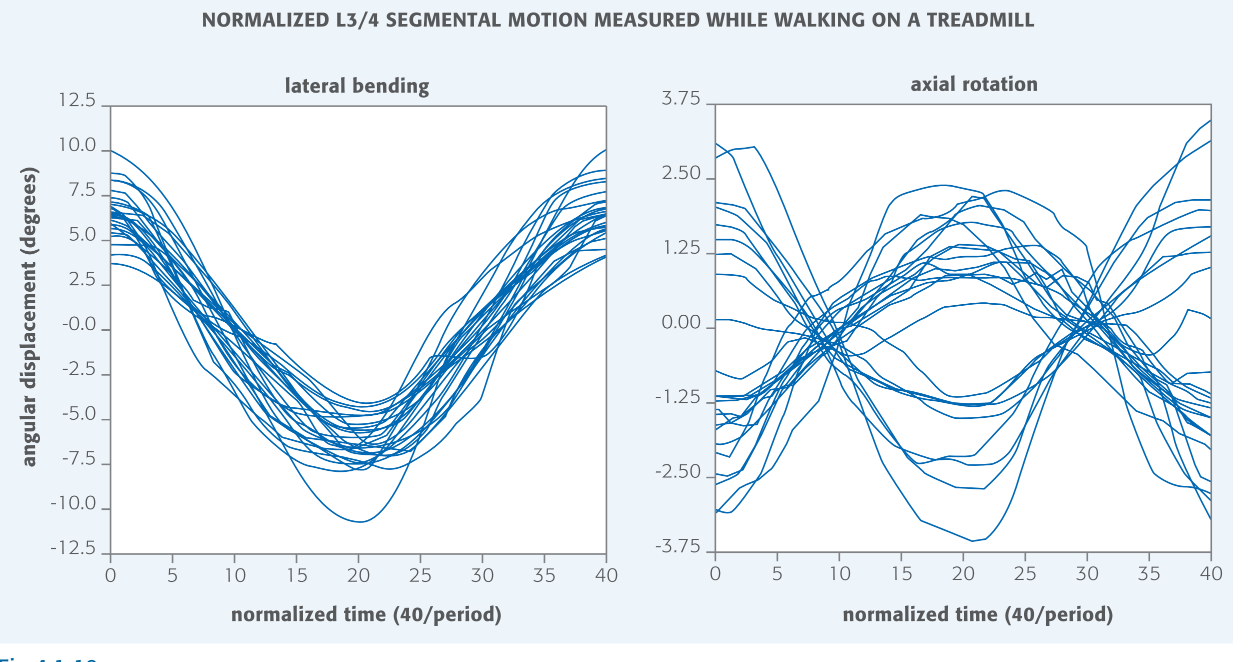

Fig 4.1-10
Segmental kinematics for the L4/5 motion segment were recorded while walking on a treadmill at 2.5 mph. An invasive tracking method was used [22]. The angular displacement curves normalized for a single gait cycle are overlaid for 23 asymptomatic subjects measured in this study. The left curves display lateral bending, which was fairly uniform across the measured population. Interesting is that the coupling in axial rotation (right curves) was determined to be toward the contralateral direction for 16 subjects, and toward the ipsilateral direction for 7 subjects. This disperse result illustrates intersubject variability for coupling patterns.

degeneration (**Fig 4.1-11**), neoplasm, or infection with destruction of passive structures. Instability can also be a complication of spinal trauma or it can be iatrogenic resulting from spine surgery. Any surgery cutting or removing passive structures (ie, facetectomy, laminectomy, discectomy) can potentially lead to segmental instability.

To diagnose segmental instability clinicians early on used increased range of motion manifestations such as for antero-posterior shear [25] or flexion/extension. But, segmental instability cannot be comprehensively described when using static end range of motion x-rays (ironically called "dynamic x-rays"). Losses in segmental stiffness [26] as well as changes in coupled motion patterns were equally described [27] as clinical manifestations of segmental instability.

Segmental instability is probably best identified under increased axial loading, provoking abnormal segmental motion [28]. Sudden changes in load may result in unanticipated or unpredictable intersegmental motion. Even during continual and wide trunk movements, particularly at low ligament pretension, abrupt segmental motion might occur. The manifestation of instability is more likely to be observed in the mid ranges of spinal motion, and can have a wide range of

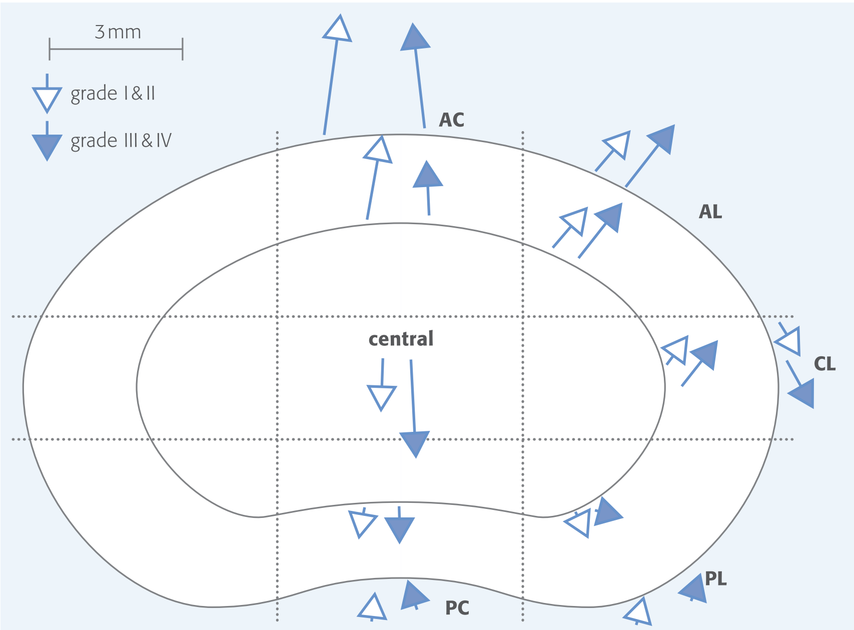

Fig 4.1-11
Transversal disc map showing internal disc displacements during flexion-compression measured for the posterocentral (PC), posterolateral (PL), centrolateral (CL), anterolateral (AL), and anterocentral (AC) annular regions, as well as for the central nuclear region. The internal displacements were obtained from fine wire markers inserted into the disc, followed with sequential high resolution x-rays while loading the specimen [3]. Discs with a Thompson degeneration grade I and II (line arrow) are compared with discs featuring a grade III or IV (solid arrow). The larger nuclear displacement during loading is evident for the more progressed degeneration stages. It is also indicative of segmental instability.

manifestations (**Fig 4.1-12**). Kaigle et al [29] have demonstrated midrange segmental instability motion in a porcine instability model as well as in symptomatic patients.

A specific manifestation of instability is due to axial load. A linked and tall structure such as the ligamentous spine is inherently unstable and may buckle already at very low vertical loads [30]. Instability or buckling occurs when a displacement perturbation from an equilibrium position results in a force tending to increase the displacement. But, the spine cannot be analyzed as an independent structure. In vivo, the spine does not collapse easily because the trunk (ribs, connective tissue, passive musculature) stiffens the ligamentous spine already by about a factor of ten. Dynamic muscle actions stabilize the spine further, which is particularly important for the cervical spine. Muscle activation alone can increase cervical spine stiffness by a factor of five.

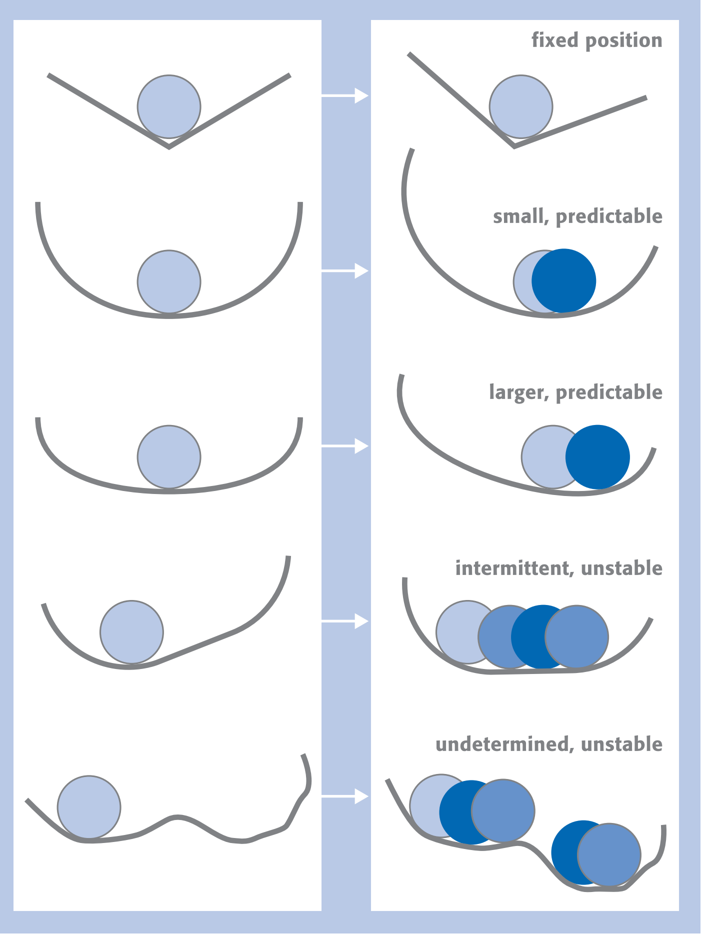

Fig 4.1-12
A mechanical model for segmental instability is presented. The drawing on the left shows a bowl with a rolling ball placed inside, coming automatically to rest at its lowest position. When slightly tilting the bowl, the ball will find a new position. Clearly defined ball positions that are reached after a relatively small excursion depict a stable situation. Undetermined positions and relatively large excursions of the ball are typical for instability. From top to bottom drawings, instability is increasing.

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4.2 BIOMECHANICS OF SPINAL STABILIZATION

1 INTRODUCTION

The following chapter is intended to provide a brief overview of current biomechanical concepts in spinal stabilization. It would not be possible in such a short review to cover the myriad of spinal stabilization devices and techniques available to the surgeon today. Indeed, whole volumes have been written about clinical instability of the spine and the biomechanics of spinal stabilization [1, 2], and the reader is encouraged to explore these resources, not only for a more in-depth treatment of the subject, but also for the historical perspective that can be gained into the rapid development of the field over the last two decades.

Each surgical procedure for spinal stabilization has its own unique structural and biomechanical characteristics. It is important that one chooses the appropriate implant and technique, by understanding the specific nature of each case.

With the exception of recent developments in spinal arthroplasty, spinal stabilization is a means to achieve the end goal of solid bony fusion. The goals of conventional spinal arthrodesis are:

- To support the spine when its structural integrity has been severely compromised (to reestablish clinical stability).
- To maintain correction following mechanical straightening of the spine (scoliosis, kyphosis, osteotomy).
- To prevent progression of deformity (scoliosis, kyphosis, spondylolisthesis).
- To alleviate or eliminate pain by stiffening a region of the spine (diminishing movement between various spinal segments).

To choose a stabilization method which will best achieve these goals, the surgeon requires an understanding of the mechanics of load transfer through the spine, how this load transfer is altered by the specific injury or pathology, and the relative merits of each particular surgical technique. There is no single solution for spinal stabilization, and in the following sections, the biomechanical aspects of some of the more common techniques will be discussed. These include: posterior stabilization, anterior stabilization, and intervertebral cages, as well as newer concepts in nonrigid stabilization.

2 POSTERIOR STABILIZATION

Pedicle screws have dramatically improved the outcomes of spinal fusion. Short segment surgical treatment using pedicle screws and rigid connecting plates or rods has proven to be safe and effective for the treatment of neoplastic, developmental, congenital, traumatic, and degenerative conditions [3].

The stabilizing potential of posterior spinal fixation systems has been demonstrated in many biomechanical studies. For example, a comparison of the internal fixator and the USS [4] has shown that motion of the stabilized spinal segment is reduced by up to 85% in flexion, 52% in extension, 81% in lateral bending, and 51% in axial rotation. Additional stability can be achieved by adding cross-links [4]. Similar results have been reported in other studies of the stabilizing potential of different posterior fixation systems [5, 6].

Posterior systems derive their stability from a solid anchorage in the pedicle and the inherent rigidity of the connecting instrumentation. The pullout strength of pedicle screws is directly related to the bone density [7]. It is possible to achieve an increase in pullout strength by choosing convergent screw trajectories (**Fig 4.2-1**) and by the addition of cross-links. Furthermore, it has been shown that with parallel pedicle screws in short-segment constructs, an unstable "four-bar" mechanism can result in the absence of adequate anterior column support (**Fig 4.2-2**); therefore, triangulation of pedicle screws is recommended for better stability. The same rationale applies for cross-linking the rods. Here, diagonal cross-linking is preferable to the horizontal configuration in terms of rotational stability [8]. The stiffness of the fixator construct depends heavily on the diameter of the connecting rods. Compared with a system using 7 mm rods, a 10 mm rod has a 4.1 times higher bending stiffness, while a 3 mm rod has a nearly 30 times lower bending stiffness [9]. Should these systems be made as rigid as possible? An increase in rod diameter provides a more stable construct, but at the same time it produces higher internal loads in the implant, on the clamping device, and on the pedicle screws, and thus a higher risk of screw breakage [9]. Therefore, a compromise between an absolutely rigid fixation and a minimal risk of implant failure must be achieved, and this compromise is reflected in current implant designs which provide stable fixation [4] with a proven service life.

While pedicle screws have been accepted as a reliable and safe method for stabilizing the thoracolumbar spine, their use in the mid and upper thoracic spine is more complicated, due to the smaller overall dimensions of the thoracic pedicles, and greater variation in pedicle morphology. An alternative to standard intrapedicular screw placement is the extrapedicular screw trajectory (**Fig 4.2-3**), first described by Dvorak et al [10]. Greater pullout strength has been measured for extrapedicular screws, likely due to the greater angulation possible with this technique, the longer screw length, and the perforation of up to four strong cortices. The overall 3-D stability of thoracic spinal segments stabilized with extrapedicular screws has been shown to be equivalent to that of the conventional intrapedicular technique, and with no additional risk of loosening through fatigue [11]. Thus, this technique offers an alternative, which provides greater safety due to the increased distance between the screws and the spinal canal, while avoiding any compromise in the rigidity and strength of the construction.

The use of simple lamina hooks in the thoracic spine is safe with respect to damage of neural structures. However, hook disengagement has been reported in scoliosis correction surgery [12]. To achieve a higher resistance to the complex 3-D

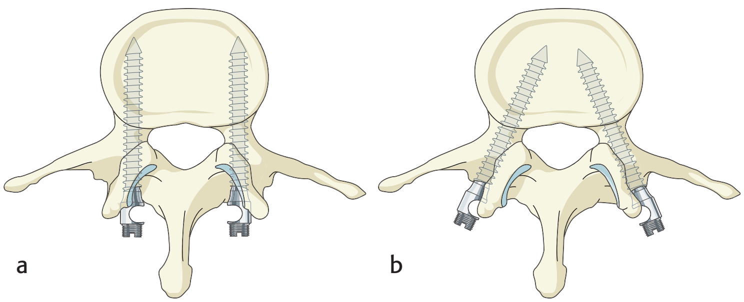

Fig 4.2-1a–b
The use of convergent screw trajectories (right) increases the pullout strength and overall stability of pedicle screw constructs, in comparison with parallel screw insertion (a).

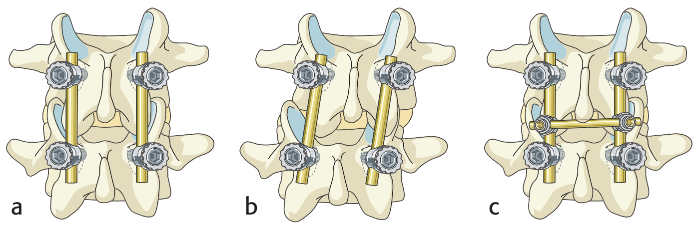

Fig 4.2-2a–c
a–b The use of conventional parallel pedicle screws and rods (a) for spinal segments with diminished anterior integrity may be inadequate. Displacement of the stabilized segment by rotation of the pedicle screws—a so-called “four-bar” mechanism—may result (b).
c Further stability can be achieved by the use of convergent screw trajectories or the addition of a cross-link.

forces, pedicle hooks with additional supporting screws have been developed [13, 14]. Biomechanical pullout tests have shown that a significant increase of failure load can be achieved with the use of screw-augmented hooks [15].

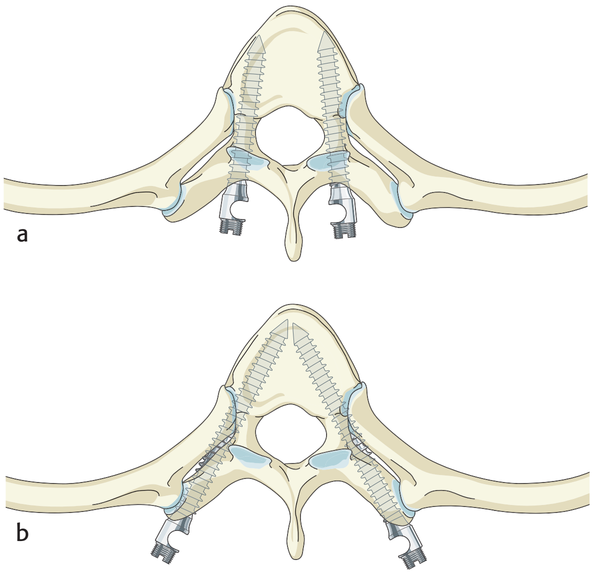

Fig 4.2-3

In contrast to the standard intrapedicular screw insertion (a), an extrapedicular screw insertion (b) allows a greater margin of safety with respect to the spinal canal in the thoracic spine, and may offer greater pullout strength and stability.

3 LOAD-SHARING CHARACTERISTICS OF STABILIZED SPINAL SEGMENTS

Spinal implant constructs and the stabilized spinal segment together form a mechanical system. Loads and moments are shared between the natural anatomy and the stabilizing hardware. Recent in vivo measurements, using a telemetric device, have provided valuable insight into the 3-D loading of an internal (posterior) fixator during daily physiological loading [16]. However, measurements of fixator loads alone provide no information about the overall force flow, ie, how much of the total load is transferred by the implant and how much by the spine. Cripton et al [5], in a combined experimental and analytical study, have quantified the load sharing in stabilized spinal segments. By simultaneously measuring intradiscal pressure and the forces in a modified internal fixator during physiological loading, analysis of the load distribution within the instrumented spinal construct was possible by applying principles of force and moment equilibrium. The results of this study provide valuable insight for the design of spinal implants, and also for the evaluation of surgical indications.

In the intact stabilized spine, it has been demonstrated that, for flexion and extension, spinal loads are carried predominantly by equal and opposite forces in the disc and the fixator, a force couple. Only a small portion of the total loading is transferred directly by bending of the implant or through the posterior elements. For side bending, the majority of loading is transferred through equal and opposite forces in the fixator rods. For torsional loading, loading is distributed more or less evenly between implant forces, torsional resistance of the disc, and forces acting on the posterior elements. Therefore, the anterior structures play a crucial role in the overall load-bearing function of the stabilized spine. The load-sharing in instrumented spinal segments is summarized in **Fig 4.2-4**.

In the case of severe anterior column injury, all loads must be carried by the implant itself. Based on the in vivo measurements of implant loading by Rohlmann et al, and the force flow analysis in the study of Cripton et al, global moments of up to 30 Nm may act through the spine [5]. This may exceed the safe limit for many implants. Therefore, in the case of very unstable anterior column injuries, additional support of the anterior column is critical to prevent failure of the instrumentation. The importance of effective load sharing between the anterior and posterior spinal columns is further reinforced by the work of Polly et al [17], in which it was shown that the overall stiffness of the stabilized spine increases by a factor of three, as an interbody graft is moved within the disc space from the posterior toward a more mechanically advantageous anterior position.

Further work is required to characterize the force flow through instrumented spinal segments, including the load transfer through intervertebral devices and anterior stabilization constructs.

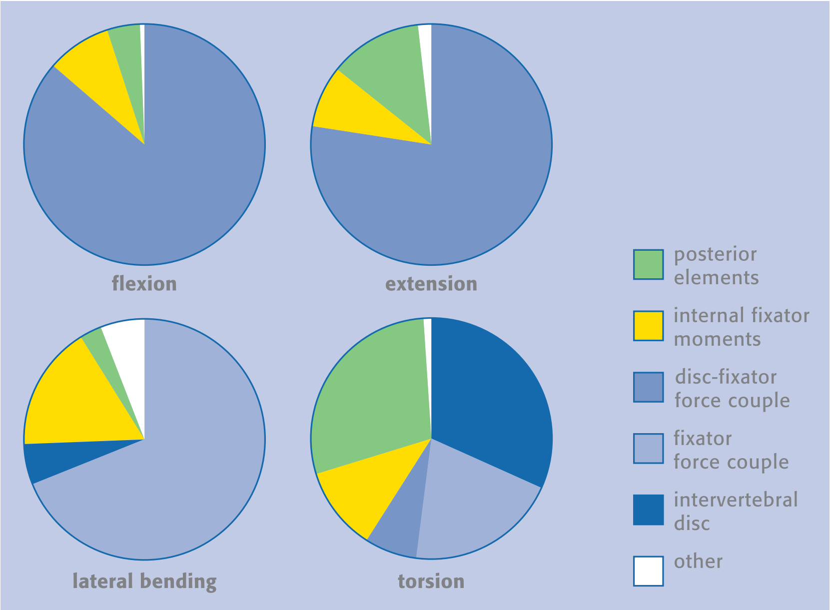

Fig 4.2-4
Predicted load sharing between a standard posterior stabilizing implant and the anatomical structures of the spine. The integrity of the anterior column is crucial for successful load bearing in the spine, even with rigid metallic implants. Adapted from Cripton et al [13].

4 ANTERIOR STABILIZATION

The importance of an intact anterior column for the load-bearing function of the spine has been demonstrated. In certain cases, spinal stabilization can be effectively achieved using only anterior implants. Anterior cervical plating offers several advantages for stabilizing spinal constructs, including good visibility of the fusion site, decreased rate of graft expulsion, and increased fusion rate in multilevel constructs. Anterior cervical plates act as a tension band during spinal extension and as a buttress plate during flexion. Constrained cervical systems have a rigid, angle-stable connection between the plate and screws (eg, cervical spine locking plate, CSLP), whereas unconstrained systems rely on friction generated by compression of the plate against the anterior cortex for stability (eg, H-plate) (**Fig 4.2-5**). In extended biomechanical testing, constrained systems have shown a greater rigidity, whereas unconstrained plates can lose a significant amount of stability over time [18]. However, it has also been shown that the capability of the CSLP to stabilize the spine after a three-level corpectomy is significantly reduced after fatigue loading [19], whereas no difference in stability was noted for a single-level stabilization. Therefore, the demands of the surgical indication heavily influence the performance of the implant. The surgeon has the option of selecting systems with monocortical or bicortical screw fixation, often with the same plate. In general, no significant differences in stability have been found between monocortical and bicortical fixation [20], however, further improvements in stabilization have been shown using monocortical locking expansion screws [21]. Bicortical screw fixation still has specific indications, eg, for multilevel stabilization, poor bone quality, or realignment of kyphotic deformities, but it also has the potential to abut on the spinal cord. Another concern in the cervical spine, with its inherent mobility and relatively low compressive forces, is delayed or nonunion (pseudarthrosis) due to possible stress shielding of the graft. This is particularly true for the latest generation of constrained (locking) plates with which it is more difficult to set the graft under compression. For this reason dynamic (semiconstrained) anterior plates may be preferable. Reidy et al have shown in a cadaver corpectomy model that axial load is transmitted preferentially to the graft with the dynamic cervical plate, in comparison to a static plate, especially when the graft is undersized [22].

Several systems have been developed for anterior stabilization of the thoracolumbar spine, including the Ventrofix and the Kaneda SR, used mostly for reconstruction in trauma, tumor, and posttraumatic kyphosis. The advantages of anterior fixation are better decompression of the spinal canal and reconstruction of the anterior column, combined with excellent visibility.

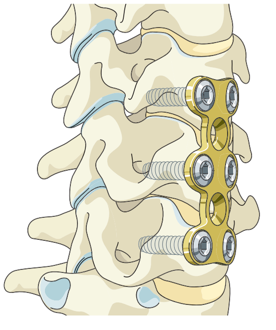

Fig 4.2-5
Constrained cervical fixation systems rely on an angle-stable connection between the plate and screws for the efficient transfer of load; monocortical screw fixation is possible. Conventional plates rely on friction to transfer load, and usually require bicortical screw insertion.

Anterior stabilization devices transfer load through a combination of compressive or tensile loading along the length of the implant and bending or torsion of the implant. Due to the low profile of these implants, and their position directly on the anterior column, bending forces are much lower than for posterior pedicle screw systems, but their stabilizing potential is also lower, due to a shorter effective lever arm. The relative effectiveness of anterior, posterior, and combined anteroposterior fixation in a corpectomy model has been addressed in a study by Wilke et al [23]. It was shown that typical posterior pedicle screw systems provide excellent stability in flexion and lateral bending, but not in extension or axial rotation; however, this stability is dependent on adequate support of the anterior column, with an interbody graft if necessary. In extension, motion is restricted only by the stiffness of the posterior implant itself. Likewise, anterior fixation provides stabilization in flexion and lateral bending, but not in extension or axial rotation. In lateral bending, the implants provide better stabilization when the spine is bending away from the implant side, as the devices act as a tension band. Anterior double-rod systems provide better stabilization than single-rod systems, and systems which use locking head screws are stiffer than those without (**Fig 4.2-6**). The addition of a transverse element further increases the stability of a double-rod construction. In all loading directions, combined anteroposterior fixation provides better stabilization than posterior or anterior stabilization alone. Therefore, in cases of severe destruction of the vertebral body or gross fracture dislocation, combined anteroposterior fixation would be warranted.

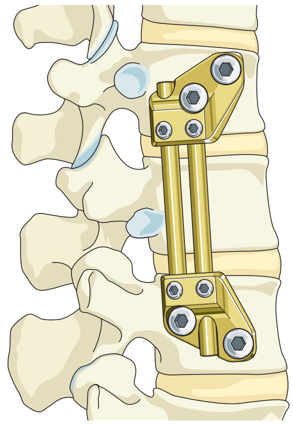

Fig 4.2-6
Anterior double-rod fixation systems provide increased resistance to torsional loading (eg, Ventrofix). The addition of transverse elements or locking head screws further increases the stiffness of such implants. In cases of severe destruction of the vertebral body or gross fracture dislocation, combined anteroposterior fixation would be warranted.

5 INTERVERTEBRAL CAGES

Intervertebral cages have been developed to augment spinal arthrodesis by reconstructing the anterior column to restore the height of the intervertebral space, thereby stabilizing the affected segment and providing containment for cancellous bone graft. A variety of cage designs are available for insertion using an anterior or posterior approach [24, 25]. These include: bilateral threaded, cylindrical implants (eg, BAK, Ray TFC), bilateral box-shaped implants (eg, contact fusion cage), and single, open-box or ring-shaped implants (eg, Syncage). Intervertebral cages were originally proposed as stand-alone devices for anterior lumbar interbody fusion (ALIF) or posterior lumbar interbody fusion (PLIF).

The clinical success of stand-alone intervertebral cages is beyond the scope of this brief review; however, the mechanical requirements for successful stand-alone devices are substantial. Axial compressive loads in the spine range from 400 N to more than 7000 N during heavy lifting [2]. Intervertebral cages must be strong enough to bear these loads without failure of the implant itself, however, the bone graft around and within the cage must be stressed and strained sufficiently to evoke the biological signals (release of cytokines) for bone formation [26, 27]. In this context it is proposed that extensive stress-shielding may lead to delayed union or nonunion. This conflict is reflected in most current cage geometries and materials, but further work is required to fully understand the underlying mechanobiology [28]. These devices must also resist penetration or subsidence into the underlying cancellous bone of the vertebral body. The subchondral bone of the vertebral end plate provides the necessary strength for cage support. Removal of the end plate to provide a bleeding cancellous bone bed may compromise this support, especially for devices with a limited contact area, as the resistance to implant subsidence then depends on the quality of underlying trabecular bone [29]. However, the strength of the end plate has been shown to be greatest at its periphery in the posterolateral corners [30–32], and removal of only the central end plate only does not compromise the strength of the cage-bone interface, especially for implants with a large, peripheral contact area, such as the Syncage [33]. Thus, an effective compromise between the biological and biomechanical requirements for fusion can be achieved.

The 3-D stabilizing potential of anterior stand-alone cages has been critically evaluated in several biomechanical studies [24, 34]. While most ALIF cage designs improve stability of the instrumented spinal segment in flexion and lateral bending, the stability in extension and axial rotation may not be adequate [34]. Comparison of anterior implantation and lateral implantation has shown that resection of the anterior annulus is not responsible for this lack of stability [35], which has led to the conclusion that the lack of rigidity may be associated with distraction of the facet joints during cage insertion (**Fig 4.2-7**). Although this contradicts the original concept of “distraction compression” by Bagby [36], whereby the distracted annulus imparts a compressive force on the interbody cage, stabilization of the intervertebral space due to pretensioning of the annular fibers is likely only temporary, due to the viscoelastic nature of this tissue [37].

More significantly, biomechanical testing of PLIF devices has shown that, as a stand-alone device, cages inserted from a posterior approach do not provide adequate stabilization [25, 38]. Instability in all three principal motion directions has been demonstrated for PLIF devices with varying designs, most likely as a result of the necessary destruction of the facet joints and posterior annulus [25]. In this case, box-sectioned cages may be preferable in order to achieve maximum height for distraction, while minimizing the width of the required approach.

The use of additional posterior instrumentation improves the stability of ALIF devices, and would appear to be critical in ensuring success with PLIF devices. Supplemental translaminar or pedicle screw fixation significantly reduces the critical extension and axial rotation motion of spinal segments stabilized with intervertebral cages [25, 34]. The use of translaminar screws may be preferred, as these can be placed in a minimally invasive fashion. Without posterior stabilization, instability in extension could lead to nonunion, loosening or migration of the cage.

A potential alternative for the above-mentioned combined instrumentation is the recent development of a novel "stand-alone" device which merges the principle of the interbody cage with the anterior tension band instrumentation (Synfix). Cain et al have compared the stabilizing properties of this screw-cage construct with conventional 360° instrumentation using cage and pedicle screws or translaminar screws. Motion analysis demonstrated a significant increase of segmental stiffness with the Synfix compared to cage/translaminar screw instrumentation in flexion/extension and rotation [39]. However, testing was nondestructive and included only a few cycles. For a definite judgment the comparative biomechanical behavior under repetitive loading (fatigue) as well as clinical results and fusion rates have to be evaluated.

In the cervical spine, in contrast to the lumbar spine, stand-alone interbody cages (or structural bone grafts) are used routinely after one-level discectomy and have demonstrated near 100% fusion rates. In a comparative biomechanical in vitro study, cervical segmental stability has been assessed after implantation of interbody cages and structural bone grafts. After single-level discectomy, physiological segmental stability could be reestablished with both techniques, with the cage tending to result in slightly higher stiffness [40].

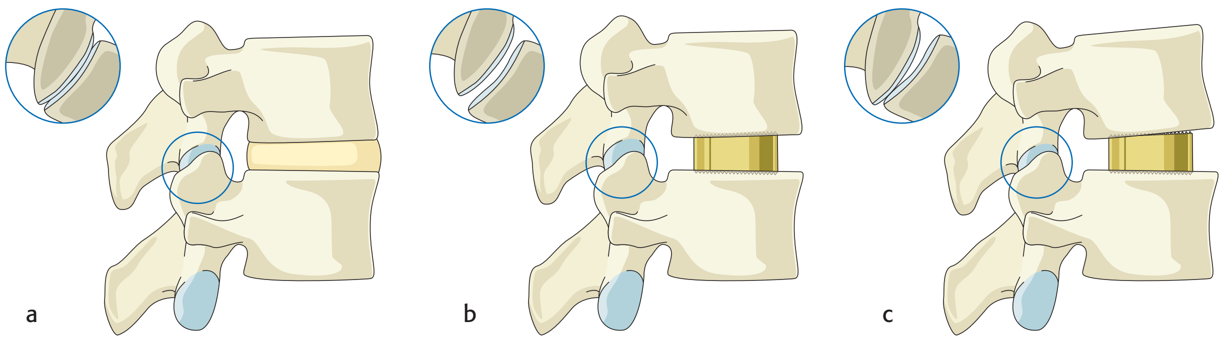

Fig 4.2-7a–c
The use of intervertebral cages as a stand-alone implant for spinal fusion may be limited by the poor stabilization in extension. Extension is limited in the normal spine partly by the interaction of the facet joints (a). Following the insertion of a stand-alone cage, the facet joints are distracted (b) and the spinal segment is more mobile (c).

6 ADJACENT SEGMENT EFFECTS

Due to the proven rigidity of current spinal internal fixators, it has often been suggested that degeneration of the disc adjacent to a fused spinal segment is the result of increased biomechanical stress on this motion segment. Shono et al [41] have shown in a biomechanical in vitro study that the displacement of the adjacent motion segment is increased following fusion. In these experiments, a fixed displacement was applied to the entire spinal specimen, and it is therefore logical that, as the motion of the fused segment decreases due to its increased stiffness, motion at the adjacent segment must increase to produce the total displacement. On the other hand, Rohlmann et al [9] have demonstrated with a simplified analytical model that the influence of rigid instrumentation on the adjacent discs is minor. In their analysis, controlled loads were applied to a spinal model. This seemingly contradictory result may also be reasonable, as the response of the mobile disc to a given load is determined only by its own inherent stiffness, which is not altered by the adjacent fusion. Nevertheless, small but significant increases in adjacent segment mobility have been shown in vitro when controlled loads were applied to spinal segments [42]. Is "adjacent segment disease", therefore, the result of altered biomechanical stresses? This depends on whether adjacent segment motion in vivo is increased following fusion. The animal study of Dekutowski et al [43] provides some support for increased adjacent segment motion, however, the overall incidence of adjacent segment degeneration would likely be much higher if its cause were purely mechanical. It is well accepted that disc degeneration is a multifactorial disease with genetic and environmental factors [44]. To which extent mechanical factors contribute to the disease likely also determines whether or not disc degeneration is initiated or aggravated adjacent to a fused segment.

7 DYNAMIC STABILIZATION

Nonrigid posterior stabilization of the spine is a relatively new concept for the treatment of spinal pathologies. Ligamentoplasty was introduced in 1992 by Graf. This posterior dynamic stabilization system consists of pedicle screws connected via elastic polymer elements [45]. The underlying philosophy is to maintain physiological lordosis while restricting flexion/extension motion to unload and "protect" the respective disc. In vitro studies have demonstrated that flexibility is reduced in all principal directions with the Graf ligamentoplasty [46], however, the clinical success of this device has been controversial [47, 48].

Currently, the most advanced and frequently used device is the dynamic neutralization system (Dynesys) for the spine. Dynesys is a nonfusion pedicle screw system composed of titanium pedicle screws joined by polycarbonate urethane spacers containing pretensioned polyester cords. With such a system, affected segments can be restored to their proper anatomical position, and motion in all planes can be effectively controlled. However, by their design motion is not absolutely prevented, in contrast to solid fusion implants. It has been shown, in a cadaveric model of the destabilized spine, that Dynesys is able to improve stability in all principal anatomical directions, however, axial rotation was poorly controlled while in flexion and bending the system is potentially as stiff as a conventional internal fixator [49]. Freudiger et al [50] have demonstrated that the Dynesys limits shear translation of unstable spinal segments under much higher levels of physiological loading, and reduces bulging of the posterior annulus, which may relieve pain. Due to the compliance of the construction, overloading of adjacent segments may be prevented. Furthermore, as the development of these devices continues, the compliance of the connecting elements may be optimized to partially restore the normal kinematics of the stabilized segment. However, the efficacy of such a system depends heavily on the condition of the anterior column. Furthermore, the long-term performance of such a device may be limited by material fatigue or screw loosening, as the instrumentation must continue to bear load throughout the whole life of the patient.

8 INTERSPINOUS PROCESS DISTRACTION

The principle of implanting a spacer between adjacent spinous processes was already used by Knowles in the late 1950s to unload the posterior annulus in patients with disc herniation, thereby achieving pain relief [51]. In recent years various systems have entered the market, such as the Interspinous "U", the Diam, the Wallis, and the X-Stop. All devices aim to limit motion in extension. Biomechanical testing has shown that extension motion is diminished while flexion, axial rotation, and lateral bending are maintained [52]. Restricting extension is thought to reduce narrowing of the spinal canal and buckling of the yellow ligament [53]. Furthermore, an unloading of the facet joint has been demonstrated in an in vitro cadaver study [54].

The resulting increase of segmental kyphosis is likely compensated by the adjacent segments, and how this may affect the sagittal profile and balance in the long term needs to be evaluated in the future. However, although there is limited clinical follow-up data available, for patients with spinal stenosis which improves in flexion, the interspinous device is a feasible option which causes limited trauma with implantation.

9 ARTHROPLASTY OF THE SPINE

Functional disc replacement is a logical progression in the treatment of degenerative disorders of the intervertebral joint. Arthroplasty in the spine has several possible advantages: preservation of function, decrease of adjacent level degeneration, and no requirement for the harvesting of bone graft. An excellent historical review of arthroplasty of the spine by Szpalski et al [55] highlights the many design concepts to date. Such a device must not only possess adequate strength to withstand the considerable compressive and shear loads transmitted through the spinal column, but must also respect the complex kinematics of intervertebral motion.

The evolution of total joint prostheses in diarthrodial joints has been toward devices which emulate physiological motion patterns. Mobile bearing knee prostheses, for example, employ a large, conforming polyethylene plate, which is not fixed to the tibia as in the conventional total knee joint, but rather moves on the surface of a highly polished metallic tray which is affixed to the tibia. Theoretically, this design should allow a more natural motion pattern and a larger range of movement. Due to its conformity throughout the full range of motion, stresses transmitted through the polyethylene and into the bone should be lower, reducing polymer wear and prosthesis loosening. A similar design philosophy is apparent in many current disc prostheses.

Motion of the natural intervertebral joint cannot be compared to a simple ball-and-socket joint. The major motions of an intervertebral segment in flexion and extension are a combination of sagittal rotation plus translation. The center of rotation constantly changes throughout the full range of motion (**Fig 4.2-8**). The Bryan cervical disc system is comprised of a low-friction elastic nucleus located between titanium end plates, which allow free rotation in all directions. A flexible membrane surrounds the articulating nucleus. Using a sliding polyethylene core between two fixed metallic end plates, the Charité artificial disc allows a stable articulation with a physiological motion pattern determined by the interaction of the prosthesis, surrounding soft tissue, and facet joints. In contrast, the Prodisc and Maverick artificial disc are

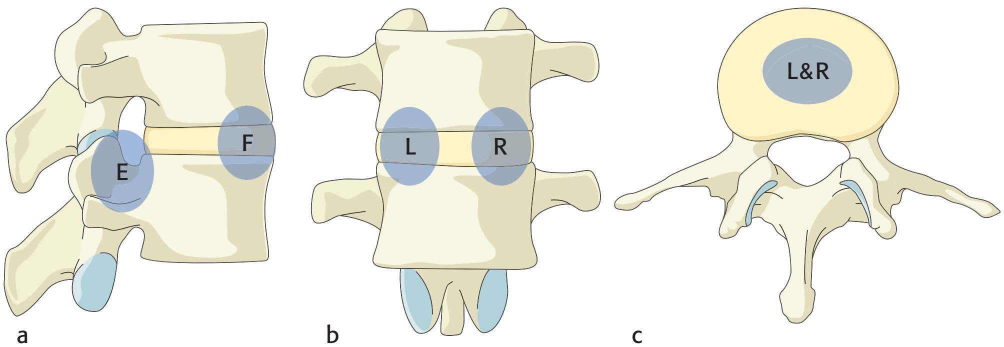

Fig 4.2-8a–c
The kinematics of the intervertebral joint are complex. The center of rotation moves during flexion/extension (a), left and right side bending (b), and left and right torsion (c). Future designs for intervertebral prostheses or dynamic stabilization systems must respect this unique characteristic of spinal motion.
F center of rotation in flexion
E center of rotation in extension
L/R center of rotation in left and right bending/rotation

constrained devices with a single articulation, allowing free rotation in all directions around a fixed center of rotation. Unconstrained devices allow a greater range of motion and theoretically prevent excessive facet loads in extreme motion. In contrast, constrained disc arthroplasties may reduce shear force on the posterior elements [56], however, anteroposterior misplacement of the device may lead to motion restriction and component liftoff [57]. Only comparative prospective clinical trials can conclusively show if any of these concepts is advantageous for the patient.

As with other total joint prostheses, the stability of the prosthesis and the motion segment likely depends on well balanced ligaments and surrounding soft tissues. Therefore, precise operation technique with retention of stabilizing tissue is essential for a good outcome. Wear of prosthesis components, as in other arthroplasties likely occurs, however, the histocompatibility for titanium and polyethylene particles has been tested in animal models and an absence of strong inflammatory host responses was shown [58, 59]. Finally, the kinematics of each new device must be verified against representative motion patterns of the normal spine [60]. In one study, spinal kinematics before and after implantation of a cervical disc prosthesis (Prodisc) was compared with spondylodesis. Using a displacement-controlled protocol, with the prosthesis in place almost no alteration in motion patterns could be recorded compared to the intact state, unlike in the fusion case where the adjacent segments compensated for the fused level to achieve full motion [61]. This is in agreement with Puttlitz et al who demonstrated the establishment of an approximately physiological kinematics in all 6° of freedom with cervical disc arthroplasty [62]. In another biomechanical in vitro study, Cunningham et al compared the Charité disc prosthesis with an interbody fusion device (BAK) with and without posterior instrumentation. Unlike interbody fusion, also in the lumbar spine, the disc prosthesis exhibited a near physiological segmental motion pattern in all axes except rotation which was increased [63]. Long-term data are still scarce for the life time of disc prostheses, preservation of motion, and long-term patient satisfaction. Therefore, total disc replacement still has to establish its advantages compared to conventional spondylodesis.

In contrast to total disc arthroplasty, replacement of only the degenerated or excised nucleus pulposus is an option offered by the Prosthetic Disc Nucleus (PDN). The PDN is a hydroactive implant which mimics the natural fluid exchange of the nucleus by swelling when unloaded and expressing water when under a compressive load. Wilke et al [64] have shown that the PDN implant can restore disc height and range of motion to normal values after nucleotomy. There is, however, little data on the long-term biomechanical and biological behavior of such implants in the intervertebral disc space, and the overall effectiveness of replacing only the nucleus pulposus in a degenerated disc.

10 SUMMARY

Current spinal stabilization techniques are designed to limit segmental motion in order to promote bone formation and achieve solid fusion. Even rigid metallic implants rely on the principle of load sharing between the anatomical structures of the spine and the implant itself to maintain stability. The integrity of the anterior column of the spine is especially critical for a successful outcome. In the case of severe anterior injury, support of the anterior column is essential to prevent failure of the instrumentation. Combined anteroposterior spinal fixation is more stable than either a single anterior or posterior procedure. Intervertebral cages are especially effective for restoring disc height, but may not offer adequate stability as stand-alone devices. Cages with supplemental posterior fixation provide full 3-D spinal stabilization. New concepts in less-rigid and fully-dynamic stabilization of the spine, which are intended to restore the function of degenerated spinal segments, are being introduced and merit further study.

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