Biomechanics of the Spine

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Abstract: The field of modern biomechanics has deep historical roots from the ancient Egyptians, who documented the earliest accounts of spinal injury (2600–2200 BC), as well as from the ancient Hindus, who were noted for their treatment of spinal deformities (3500–1800 BC). Building on this foundation, the modern explosion of spinal instrumentation introduced the concept of internal fixation for spinal stabilization, further advancing the understanding of the mechanics of musculoskeletal motion. The spine is a complex mechanical structure complete with levers (vertebrae), pivots (facets and disks), passive restraints (ligaments), and actuators (muscles). Each of these elements merits special consideration and thus is addressed individually. Understanding actions and reactions, force vectors, related component vectors, and the movements and/or deformation that they cause allows the spine surgeon to apply fundamental physical principles to clinical practice. Kinematics is the application of these physical principles toward the study of the motion of rigid bodies. Knowledge of the principles and laws that are clinically relevant regarding spinal instrumentation is crucial to success. Clinical biomechanics requires the assessment of 3 key questions: 1) how do the components of the implant connect together, 2) how does the implant connect to the spine, and 3) how does the construct function biomechanically? The ability to apply basic biomechanical principles in the clinical arena provides a framework that the surgeon can use in the clinical decision-making process.

Key Words: spine, biomechanics, vertebral column

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HISTORY

Spinal biomechanics originated during the Egyptian Old Kingdom at the time of the construction of the pyramids (2600–2200 BC). The Edwin Smith surgical papyrus documented the earliest accounts of spinal injury, where an extensive amount of knowledge related to spinal anatomy and injury was acquired through the mummification process. This Egyptian document described the differences between cervical sprains, fractures, and fracture dislocations.[1] Likewise, it was the ancient Hindus who were noted for their treatment of spinal deformities (3500–1800 BC). Lord Krishna received notoriety for the application of traction to a severely kyphotic devotee. This early passage predates Hippocrates by more than a millennium; however, it is the earliest known reference referring to spinal correction.

Hippocrates II (460–361 BC), from the island of Cos, recorded his work in the *Corpus Hippocraticum*,[1] where applied localized pressure and traction were used to treat spinal deformity. His knowledge of spinal anatomy was limited, and the treatments applied for correction were rudimentary, with only a preliminary understanding of spinal biomechanics. Hippocrates' approach to understanding spinal disease was based on humoral pathology rather than structural pathology. His writings discussed the anatomy by which the bony vertebrae were connected by intervertebral disks, ligaments, and musculature and that damage to the spinous process did not cause serious medical implications, yet injury to the spinal cord (termed *spinal marrow*) would have serious ramifications.

During the 15th century, Leonardo da Vinci represented the true Renaissance man. His writings in *De Figura Humana* detailed human anatomy.[2] He was the first to describe the spine accurately, denoting the curvatures, articulations, and vertebral levels. He also suggested that stability of the spine was provided by the cervical musculature. It was Giovanni Borelli who published the first comprehensive treatise on biomechanics in *De Motu Animalium*, however. Borelli was considered the "father of biomechanics" because of his enthusiastic integration of mechanics with biology.[2] He accurately assessed the mechanical action of the musculature and the force transmission through joints. Borelli changed the earlier concepts of muscle motion, stating that long lever arms allowed weak muscles to move heavy objects. The concept that human motion could be explained mechanically changed the current state of medicine during the 16th and 17th centuries and remains the foundation of modern biomechanics.

Judicial hangings during the 20th century further advanced the understanding of biomechanics. Original hanging fixtures were investigated for knot position to avoid decapitation and suffocation of the prisoners. This motivated Paterson in 1890 to describe the pathologic outcomes of hangings of this sort, where the "hangman's fracture" was classified as indicating a fracture at the pars interarticularis of C2 and rupture of the anterior and posterior ligaments.[3] It was Wood-Jones who hypothesized the mechanical nature of the knot position and its biomechanical contribution to the hangman's fracture. He suggested the use of a submental knot during hangings to ensure a rapid and efficient death and coined the biomechanical terms *hyperextension* and *distraction injury* as the basis for the hangman's fracture.[4]

Modern biomechanics have further advanced the understanding of the mechanics of musculoskeletal motion. The modern explosion of spinal instrumentation introduced the

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concept of internal fixation for spinal stabilization. Ongoing biomechanical research continues to improve spinal instrumentation performance to achieve a state whereby optimal biomechanics of the spine are restored.

VERTEBRAL COLUMN

The spine is a mechanical structure complete with levers (vertebrae), pivots (facets and discs), passive restraints (ligaments), and actuators (muscles).[5] The vertebral column complex consists of the primary axial load-bearing structures of the vertebral bodies and the intervening intervertebral disks. The ventral column is subsequently tied to the dorsal column at each segment through the pedicles. The laminae function as the roof to complete the bony spinal canal. The facet joints limit rotation, flexion, extension, lateral bending, and translation. The muscles and ligaments also function to limit torso movement while contributing to the axial load-bearing capacity at the same time. Each of these elements merits special consideration and thus is addressed individually.

Vertebral Body

There is a general trend toward increasing width and depth as well as height of the vertebral bodies as one descends rostral to caudal.[6–9] The only significant deviations are a tendency for the height of C6 to be less than that of C5 and C7 and the dorsal heights of the lower lumbar vertebral bodies to decrease below L2. The progressive increase in size directly correlates with strength and the axial load-resisting ability of the spine.[9–14] This phenomenon is also at least partially responsible for the decreased incidence of fractures of the lower lumbar spine.

In the cervical spine, the vertebral bodies of C3 through C7 have a rostral-dorsal-lateral projection which, when articulated with the caudal-dorsal-lateral aspects of the vertebral body above, forms the uncovertebral joint. This functions as an extension of the intervertebral disk, has an important role in rotation resistance, and is associated with the phenomenon of coupling (Fig. 1).[15]

Although there are some regional variations, the general shape of the vertebral body is that of a cylinder. Notably, the dorsal aspect is slightly concave where it forms the ventral aspect of the spinal canal. This must be kept in mind during ventral instrumentation with vertebral body screws, where lateral projection radiographs may overrepresent the extent of bone leading to neural impingement (Fig. 2).

Facet Joints

The facet joints are apophyseal joints consisting of a loose capsule and a synovial lining. Although they play a minor role in axial load bearing unless in an extension posture, their varied orientations have important implications for permissible segmental motion.

The facets of the cervical spine adopt a coronal orientation facing the instantaneous axis of rotation (IAR). The result is a high degree of mobility secondary to a diminished ability to resist flexion and extension, lateral bending, and rotation.[9,15,16]

The facets of the thoracic region take an intermediate position as the coronally oriented cervical facets transition to sagittally oriented facets in the lumbar spine.[9,17,18] This results

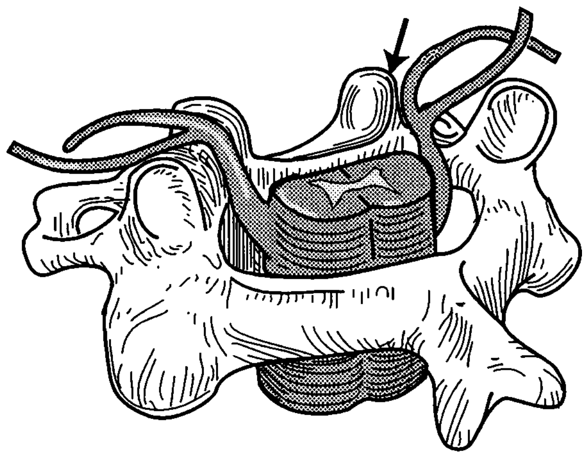

FIGURE 1. Uncinate process (arrow) and its relation to the rostral-dorsal-lateral aspect of the vertebral body and exiting nerve root. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;2.

in the lumbar spine having minimal resistance to flexion or translation but substantial resistance to rotation. The facet joint angle (with respect to midline) increases as one descends down the lumbar spine. As a result, despite the near-vertical orientation of the L5–S1 disk interspace, the sagittal orientation of lumbar facet joints has become nearly coronal at L5–S1, providing protection from subluxation. Thus, subluxation from degenerative spondylosis is more common at L4–5.

Lamina, Spinal Canal, and Spinal Canal Contents

The lamina functions primarily as the dorsal protection for the dural sac and as the foundation for the spinous processes. Forces may be applied to produce motion through

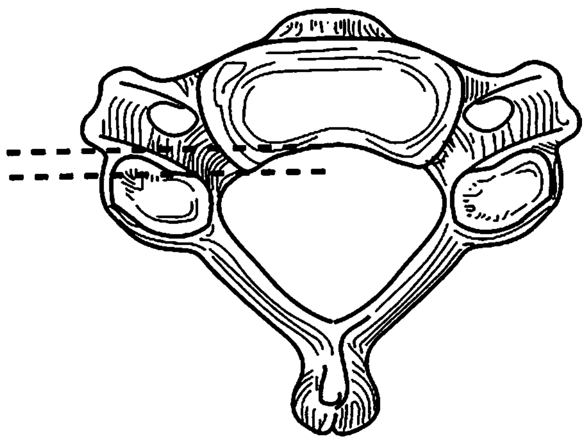

FIGURE 2. Vertebral body shape. Note the dorsally directed concavity. This may present problems in the interpretation of lateral radiographs. A lateral radiograph "sees" the dorsal aspect of the vertebral body at the level of the lower dashed line. In the midsagittal plane, however, the dorsal aspect of the vertebral body (and dural sac) is at the level of the upper dashed line. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;2.

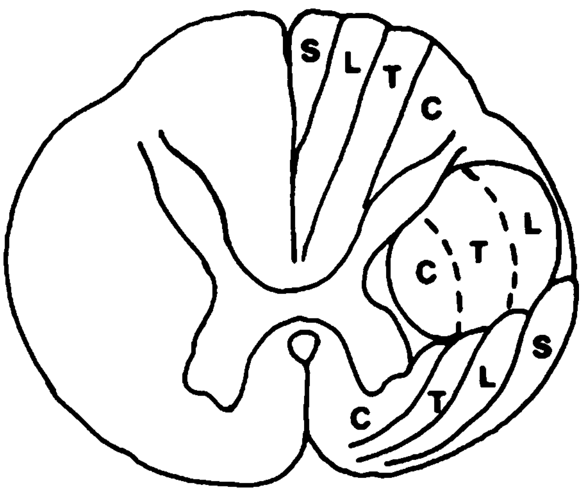

FIGURE 3. Diagrammatic axial section of the spinal cord demonstrating the somatotopic orientation of spinal tracts. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;4.

paraspinal musculature and its subsequent attachment to the spinous processes.

The spinal canal dimensions vary from region to region, resulting in a generous extramedullary space in the upper cervical spine and little extramedullary space in the upper thoracic spine.[6–8,19] In the setting of preexisting spinal stenosis, margins of safety may be significantly compromised. The implications for certain types of instrumentation, such as sublaminar hooks or wires, are obvious. Likewise, subluxation has a predictable effect on spinal canal encroachment.[20]

Tracts within the spinal cord are somatotopically oriented (Fig. 3). The descending fibers of the corticospinal tracts are arranged with hand fibers medially and foot fibers laterally. This alignment allows the earlier exiting fibers to peel off medially as they course toward their respective anterior horns. Similarly, the ascending dorsal columns and spinothalamic tracts are arranged so as to allow entering fibers an unhindered path to layer on top of more caudally originating fibers. The dorsal column fibers enter the dorsal horn and layer on more caudally originating fibers from a medial to lateral arrangement in the ipsilateral spinal cord. The spinothalamic fibers, however, cross over the midline and are thus layered caudal to rostral in a lateral to medial arrangement in the contralateral half of the spinal cord.

Pedicle

With the advent of pedicle screw fixation, knowledge of pedicle anatomy has become increasingly more clinically significant.[15,21–23] In the cervical spine, the pedicles are shorter and have a proportionally larger diameter than in the remainder of the spine.[15,21] Aside from a decrease in the upper thoracic spine, pedicle transverse width increases as one descends (Fig. 4A).[7,24,25] Although the sagittal pedicle width decreases in the lumbar region (see Fig. 4B),[7,24,25] this tends to be clinically insignificant, because transverse width is generally the limiting factor.[24,25]

The increasing transverse pedicle angle of the lumbar spine necessitates a wider angle of approach in the lower lumbar spine (see Fig. 4C).[7,24,25] This is offset by the greater margin of safety, however. The smaller margin of safety, the increased sagittal pedicle angle (see Fig. 4D),[7,24,25] and the relation of adjacent neural structures[26,27] of the thoracic spine make pedicle screw fixation there a riskier proposition than in the lumbar spine.

Intervertebral Disk

The nucleus pulposus and the annulus fibrosus of the intervertebral disk provide substantial resistance to axial stress,[28] but this ability decreases with age. In the thoracic spine, this strength is buoyed by the costovertebral joint.[29] Unfortunately, the intervertebral disk is also subjected to flexion, extension, and lateral bending forces, which can result is significant bulging and herniation of the disk.

The angled orientation of the fibers of the annulus fibrosus at 30° with respect to the end plate makes it effective at resisting rotation but poor at resisting compression. Disk bulging occurs on the concave side of a bending spine (Fig. 5A). This corresponds with osteophyte formation. The nucleus pulposus, however, moves in the opposite direction (see Fig. 5B).[30]

Transverse Processes

The transverse processes function as attachment sites for paraspinal musculature. Because of their projection and orientation, they allow leverage for lateral bending. They are often the site of fractures in trauma because of their relatively small size and the substantial loads applied in this region. The ability to use hooks in the lower thoracic spine is diminished by the atretic nature of the transverse processes in this region. The relatively larger transverse processes of the lumbar spine make them potential sites for bony fusion, but this is limited somewhat by their poor blood supply.

Spinous Processes

The spinous processes are usually directed dorsally and caudally and are bifid from C3 through C6. The spinous processes of the cervical and upper midthoracic spine tend to project in a more caudal fashion. This often necessitates their removal to access the interlaminar space.

Ligaments

The strength characteristics of the various ligaments differ from ligament to ligament as well as from region to region.[9,31] The effectiveness of a ligament is determined by its intrinsic morphology and the length of the moment arm through which it acts.[32] The moment arm is the perpendicular distance between the force vector (the force and its direction as applied by the ligament) and the IAR. Thus, a weaker ligament with a longer moment arm may be just as effective as a stronger one acting through a shorter moment arm. The interspinous ligament is not substantial and is often absent at the L5–S1 level and deficient at the L4–L5 level, but its significant moment arm length provides a substantial mechanical advantage (Fig. 6). The ligamentum flavum is a complex strong ligament. Its more ventral site of attachment and resultant shorter moment arm provide less flexion

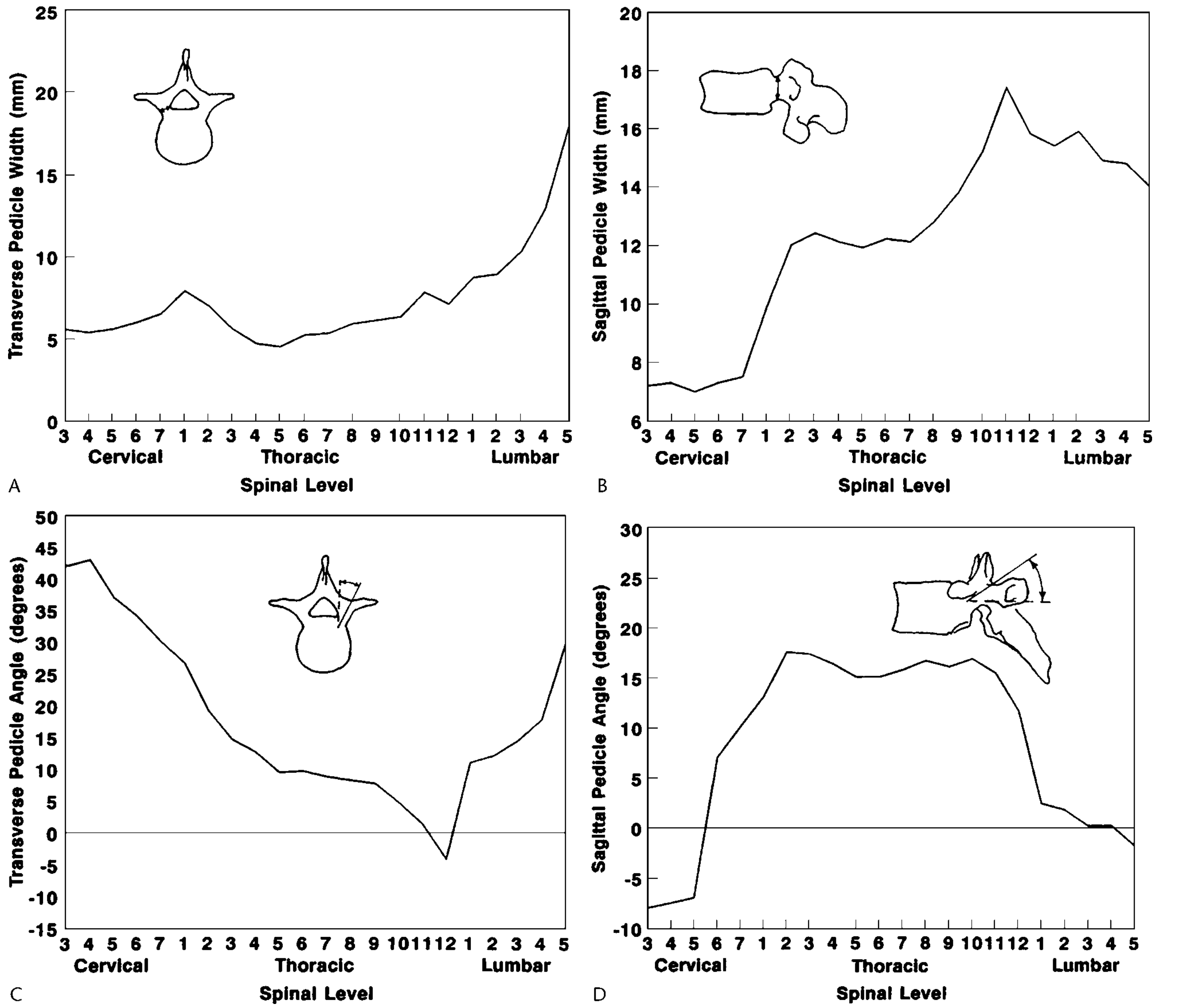

FIGURE 4. A, Transverse pedicle width versus spinal level. B, Sagittal pedicle width versus spinal level. C, Transverse pedicle angle versus spinal level. D, Sagittal pedicle angle versus spinal level. (From Panjabi et al,[7] Krag et al,[24] and Zindrick et al[25]; with permission).

resistance.[33] It is deficient in the midline, facilitating surgical exposure of the epidural space. The anterior longitudinal ligament (ALL) is a relatively strong ligament that, by virtue of its attachment ventral to the IAR, provides resistance to extension. The posterior longitudinal ligament generally has the shortest moment arm. Further weakening its ability to resist flexion is its relatively narrow width, except in the region of the disk interspace, where it widens to find its primary attachment point to the annulus fibrosus. Although capsular ligaments have a short moment arm, their strength in relation to the stresses that they resist is high.

Muscles

Muscles move the torso by directly or indirectly affecting the spine. The erector spinae muscles contribute to spinal extension and lateral bending via their bony attachments. The psoas muscle causes flexion. The rectus abdominous muscle is an indirect spinal flexor and is particularly effective because of its long moment arm.[34] Its role in spinal support should be emphasized in every rehabilitation program. The complex role of muscles is evidenced by the common chronic pain syndromes that can result from muscle imbalance as well as the difficulty in modeling its significance in biomechanical testing.

Bone

The vertebral bodies bear most of the axial load, and their dimensions are proportional to the subsequent loads. Weight-bearing potential is directly affected by the ratio of cortical to cancellous bone, which is the highest in small pedicles. Small pedicles also have a high bone density, which correlates with screw-pullout resistance. The relatively low bone density of the sacrum makes this region prone to pullout failure. Osteoporosis severely influences bone integrity. A 50% decrease in osseous mass can result in a 75% decrease in strength.[9]

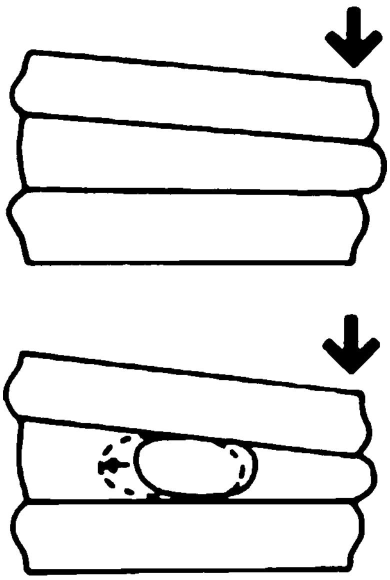

FIGURE 5. Eccentrically borne load results in annulus fibrosus bulging on the concave side of the resultant spinal curve, and annulus fibrosus tension is present on the convex side of the curve. Nucleus pulposus, however, tends to move in the opposite direction as the annulus fibrosus bulge when an eccentric load is borne (solid to dashed outline). Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization.* AANS Publications; 2001;7.

FUNDAMENTAL BIOMECHANICS

Physics is the science of matter and energy and their interaction. Understanding actions and reactions, force vectors, related component vectors, and the movements and/or deformation that they cause allows the spine surgeon to apply fundamental physical principles to clinical practice.[35,36] Kinematics is the application of these physical principles toward the study of the motion of rigid bodies. The disciplines of physics and kinematics are hopelessly intertwined. What follows is a distillation of the principles and laws that are clinically relevant regarding spinal instrumentation.

Vectors, Moment Arms, Bending Moments, and Axes of Rotation

A vector is an entity that has magnitude and direction. Applying a reference coordinate system allows a vector to be modeled as a set of component vectors. This simplification often allows complex interactions to be broken down into more digestible parts. The Cartesian coordinate system is a commonly used system. There are 3 axes: the x, y, and z axes. These have been assigned many different and more descriptive terms, but rostral, caudal, ventral, dorsal, right, and left are used here.

Forces acting on the spine may be modeled with such vectors (Fig. 7A). When a force vector acts on a lever (moment arm), it results in a bending moment (see Fig. 7B). A bending moment applied to a point in space causes rotation. The IAR is the axis about which the rotation occurs.

Each axis may be subjected to translational and rotational forces. These forces can occur in either of 2 directions that are the opposite of each other. Thus, 2 translational and 2 rotation movements around each of the 3 axes result in a total of 12 potential movements. One set of opposite movements constitutes a degree of freedom; therefore, potential movements may be considered as 6 degrees of freedom about each IAR (Fig. 8).

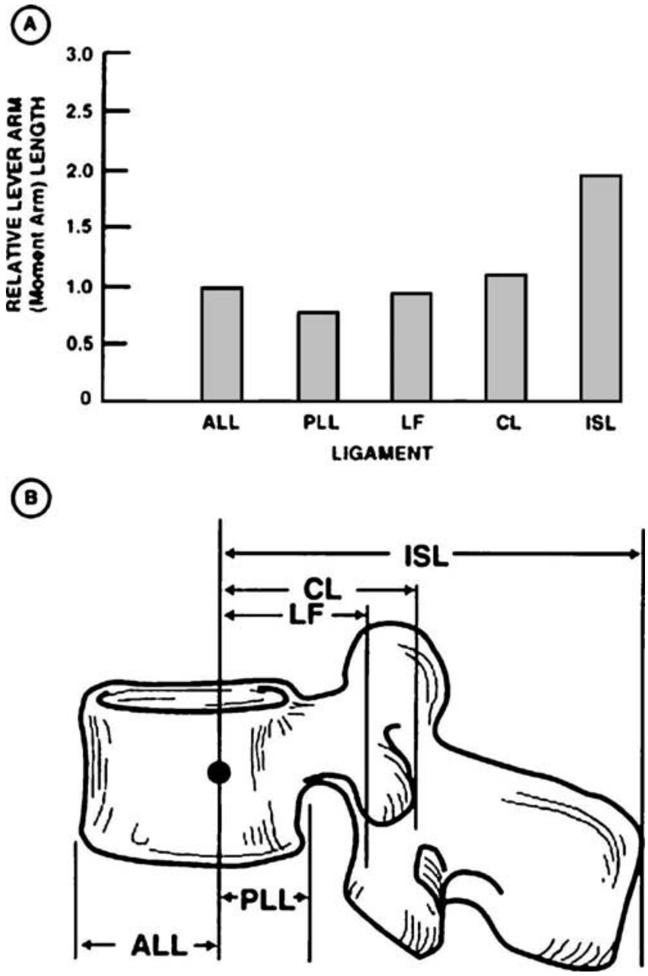

FIGURE 6. A, Relative lever arm (moment arm) length of ligaments causing flexion (or resisting extension). B, Ligaments and their effective moment arms. Note that this length depends on the location of the instantaneous axis of rotation (●). An "average" location is used in this illustration. ALL indicates anterior longitudinal ligament; PLL, posterior longitudinal ligament; LF, ligamentum flavum; CL, capsular ligament; ISL, interspinous ligament. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization.* AANS Publications; 2001;9.

When a spinal segment rotates, there is an axis that does not move, the IAR. The IAR usually passes through or at least close to the vertebral body. The exact location varies depending on the intrinsic curvature of the spine as well as other factors, such as degenerative disease, fractures, ligamentous injury, and instrumentation (Fig. 9A, B).[37,38]

The IAR may be viewed as fulcrum or transition point. On flexion, all points ventral to the IAR are brought closer together, whereas those dorsal to the IAR are further separated (see Fig. 9C). The IAR is dynamic because it moves with

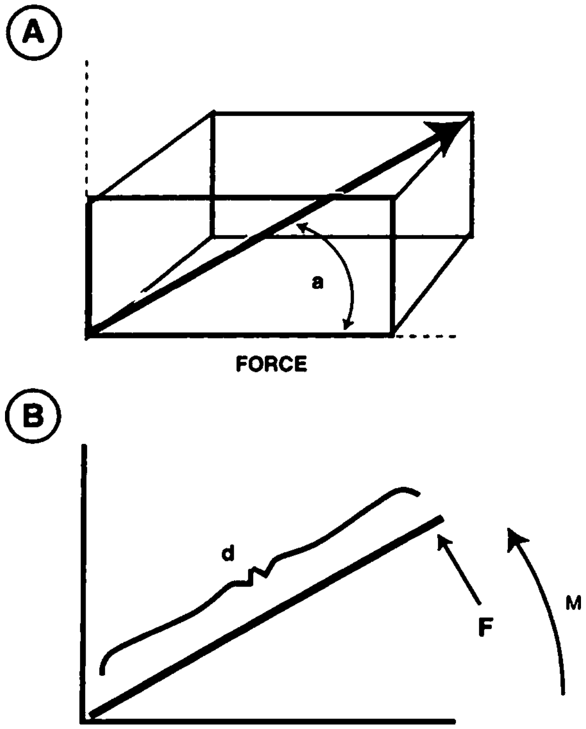

FIGURE 7. A, Force vector in 3-dimensional space. If a force (F) is applied at a distance (d) from a fulcrum (instantaneous axis of rotation), a bending moment (M) is created (B). Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;19.

movement of the involved spinal segment. When translation is superimposed on rotation, there is a corresponding helical axis of motion (HAM). This concept is helpful in modeling screw biomechanics.

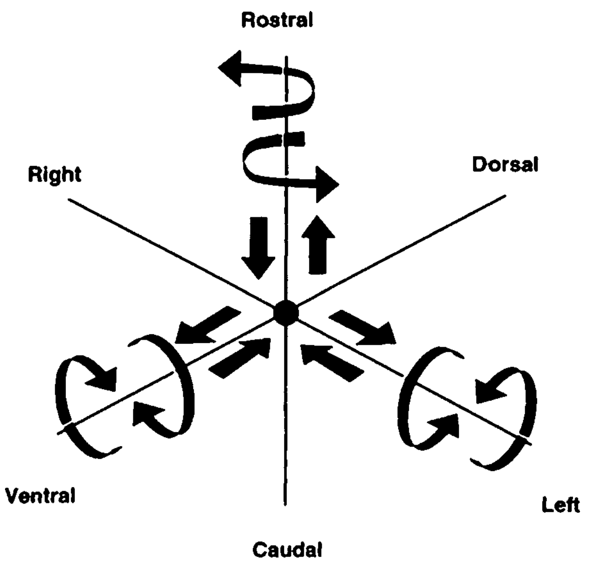

FIGURE 8. Cartesian coordinate system with the instantaneous axis of rotation as the center. Translation and rotation can occur in both of their respective directions about each axis. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;20.

When a force acts on a spinal segment, it does so through a moment arm. This may be thought of as an "imaginary lever" that extends from the IAR to intersect perpendicularly with the applied force. Applied forces may be intrinsic (eg, ligaments) or extrinsic (eg, instrumentation). The bending moment, M, is defined by the product of the force and the moment arm through which it is applied:

$$M = F \times D,$$

where M is the bending moment, F is the applied force, and D is the length of the moment arm (Fig. 10). The bending moment is effectively a torque.

Momentum and Newton's Laws of Motion

Momentum is the product of mass and velocity. It has direction and is thus expressed as a vector. It is a key concept in understanding how objects interact with other objects and their external environment. The fundamental rules governing action–reaction phenomena were elucidated by Sir Isaac Newton.

Newton's first law of motion is referred to as the law of inertia: "If a body is subjected to no net external influence, it has a constant velocity, either zero or nonzero." Thus, an object's speed and direction do not change as long as there is no force acting on it.

Newton's second law of motion is referred to as the law of superimposition of forces: "The time rate of momentum of a body is equal in magnitude and direction to the vector sum of the forces acting upon it." The forces acting on an object may be summed into a single resultant vector.

Newton's third law of motion is referred to as the law of conservation of momentum: "Interactions between objects result in no net change in momentum." When objects collide, the overall momentum of the bodies remains constant, such that any momentum lost by a single body is gained by another. In other words: "For every action there is an equal (in magnitude) but opposite (in direction) reaction."

Hooke's Law and the Load/Deformation Curve

When a load (stress) is applied to a solid at rest, there is some degree of deformation (strain) of the solid. Exactly how much deformation there is and whether any of it is permanent can be predicted by the solid's stress/strain curve (Fig. 11). The initial part of the curve is the neutral zone. This may also be referred to as the zone of nonengagement. As a biologic example, consider the bony elements of the spine. For small external forces, some of the energy is dissipated in the adjoining ligaments, tendons, and soft tissues.

As the load increases, the solid becomes "fully engaged" and enters the elastic zone. This portion of the curve is governed by Hooke's law: for small displacements, the size of the deformation is proportional to the deforming force. The relation is linear, and when the stress is removed, the strain totally recovers. Larger stresses may exceed the limits of the elastic zone and the yield point to enter the plastic zone. Hooke's law no longer applies, and the solid develops a permanent set or deformation that does not change when the stress is removed. If greater loads are applied, the solid may

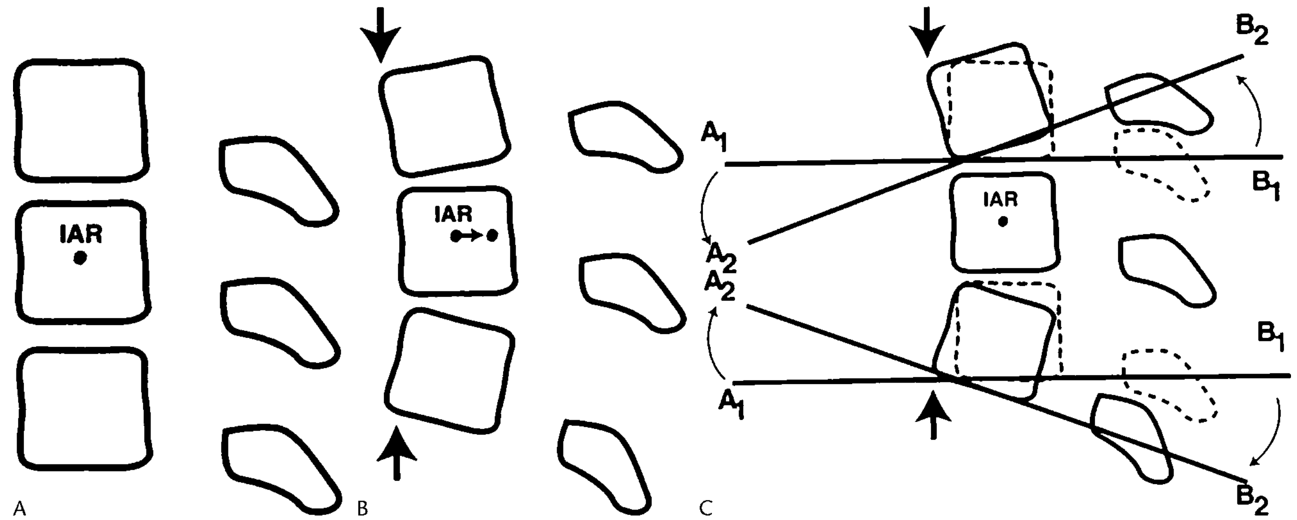

FIGURE 9. Depiction of an applied bending moment altering the location of the instantaneous axis of rotation (IAR) from the preload situation (A) to the postload situation (B). Because a ventral bending moment was applied, the IAR, as is often the case, moved dorsally. C, Depiction of the fulcrum-like nature of the IAR. If spinal flexion occurs, as depicted, all points ventral to the IAR come closer to each other and all points dorsal to the IAR become farther apart, as depicted by the curved arrows. A_1 and B_1 designate ventral and dorsal points aligned with the vertebral end plates in the neutral position. A_2 and B_2 represent ventral and dorsal points aligned with the vertebral end plates after flexion. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;20.

reach the point of failure (ultimate strength). The area under the curve is proportional to the energy absorbed before failure and is a measure of strength. The removal of stress before the yield point results in recovered energy and is a measure of the solid's resilience.

Elastic Modulus

For small deformations in the elastic zone (where Hooke's law applies), an elastic modulus may be defined as follows:

$$\text{Elastic Modulus} = \text{Stress}/\text{Strain},$$

where the elastic modulus (modulus of elasticity) is constant for a given material. Strain is the change in length or angle of a material and may be normal (linear) or shear (angular) in nature. Normal strain reflects the ability of a solid to resist tensile and compressive forces. Shear strain reflects the ability of a solid to resist angular deformation.

There are actually 3 types of elastic moduli. Young's modulus is a measure of the elastic properties of a body that is stretched or compressed. The shear modulus is a measure of the shear deformation experienced by a body that is subjected to transverse forces of equal and opposite direction applied at opposite faces of the body. The bulk modulus represents the elastic deformation of a solid when it is squeezed.

Section Modulus, Stress, and Moment of Inertia

The section modulus (Z) is an indication of the strength of an object based on its intrinsic geometry. The section modulus of a simple rod of diameter D is given by the equation:

$$Z = \pi \times D^3/32$$

It may be intuitively obvious that the strength of a rod or a screw would be substantially affected by its diameter; ultimately, when and how a rod or screw fails depends on several other factors as well, however. Stress (θ) helps to account for these other factors, such as the load and the nature of its application (eg, moment arm length). Specifically, it is

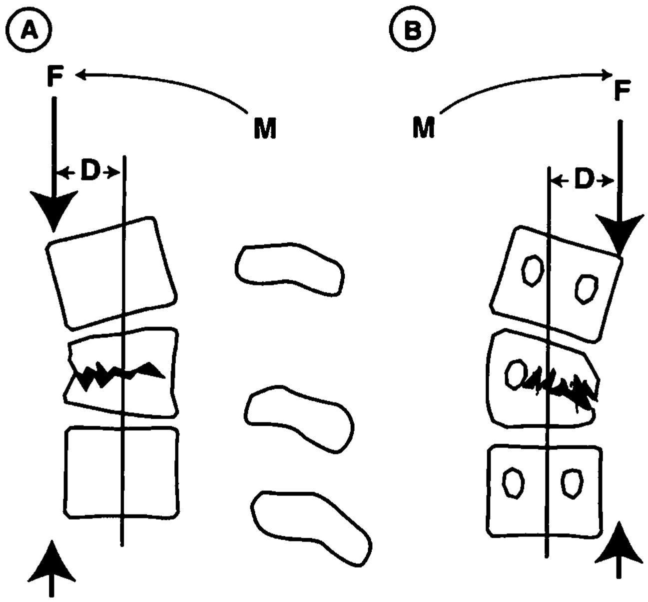

FIGURE 10. Bending moment (M, curved arrow) is the product of the force (F) and the length of the moment arm (D). The maximum bending moment is located at the center of the circle defined by the radius of the bending moment's arc (ie, instantaneous axis of rotation). A, Lateral view. B, Anteroposterior view. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;22.

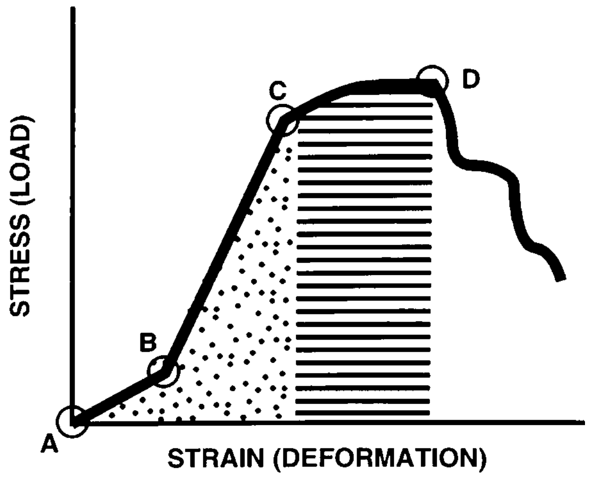

FIGURE 11. Typical stress/strain curve for a biologic tissue, such as a ligament. AB, Neutral zone. BC, Elastic zone. C, When the elastic limit (yield point) is reached, permanent deformation can occur (permanent set). CD, Plastic zone where a permanent set occurs. Past D, failure occurs and the load diminishes. The hashed plus dotted area represents strength, whereas the dotted area represents resilience. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;24.

a measurement of the force per unit area applied to a structure and is defined by the equation:

$$\theta = M/Z,$$

where M is the bending moment and Z is the section modulus. With the vital role of rods and screws in present-day constructs, it is vitally important to understand the concepts of strength and stress and their relevance in failure. Fortunately, a simple cylinder is associated with minimal mathematic complexity. As discussed previously, the strength of a rod is proportional to the third power of its diameter. The same holds true for a screw, using its inner diameter. Stress is a function of the bending moment, which is a function of the applied force and the moment arm length. Putting this all together, it is possible to predict not only instrumentation failure but the location of such failure. When an implant fails, it always fails at the point of maximum stress application (Fig. 12).

The stiffness of a cylinder is defined by its moment of inertia (I):

$$I = \pi \times D^4/16$$

Thus, increasing the diameter increases the stiffness by the power of 4, whereas the strength only increases by the power of 3.

Coupling

Coupling is a phenomenon in which a movement of the spine along or about a first axis obligates movement along or about a second axis. The mechanism and result vary among different regions of the spine. In the cervical spine, the orientation of the facet joints and the presence of the uncovertebral joints are the driving factors. Thus, lateral bending results in rotation of the spinous processes away from the concave side of the curvature. Conversely, in the lumbar spine, the spinous processes rotate toward the concave side of the curve. This explains the obligatory rotatory component associated with degenerative scoliosis of the lumbar spine (Fig. 13).

CLINICAL BIOMECHANICS

The development of a complex construct requires the assessment of 3 key questions: 1) how do the components of the implant connect together, 2) how does the implant connect to the spine, and 3) how does the construct function biomechanically?

Component–Component

Interfaces

It is helpful to start with a few definitions. An implant is a device that stabilizes the spine through connections with the spine at 2 or more points. The construct is the sum total of the implant and the spinal segments affected by the implant. Anchors (eg, screws, wires, hooks) are used to affix longitudinal members (eg, rods, plates) to the bone. A cross-fixator may be used to connect 2 longitudinal members. This section examines the interfaces between longitudinal members, anchors, and cross-fixators.

The locking mechanism used to secure 2 components is the foundation of implant integrity. There are 7 commonly used fundamental types of locking mechanisms that may be used alone or in combinations: 1) 3-point shear clamps, 2) lock screw connectors, 3) circumferential grip connectors, 4) constrained bolt–plate connectors, 5) constrained screw–plate connectors, 6) semiconstrained screw–plate connectors, and 7) semiconstrained component–rod connectors (Fig. 14). Interfaces may be enhanced through the use of knurled surfaces allowing better seating of locking screws (eg, Cotrel-Dubousset) or by providing a grid-on-grid surface.

Implant Surface Characteristics

To achieve maximum friction, 2 surfaces must "match"; stated more scientifically, they must have a high coefficient of friction (μ). The analogy of an automobile tire and the terrain it travels on provides an illustrative example.

With the rise in popularity of sport utility vehicles (SUVs), you may have noticed that off-road tires tend to have deep treads with a knobby surface. Similarly, with the rise in popularity of auto racing, you may have noticed that racing tires tend to be wide and smooth. The tires are intentionally designed to match the surface that they travel on—rocky and irregular paths contrasted with smooth and flat racing tracks. In the first example, a rough surface is matched to a rough surface, whereas in the second, a smooth surface is paired to a smooth surface. This achieves maximum surface-to-surface contact.

The biomechanical equivalents in spinal instrumentation are the grid-on-grid interface and a circumferential grip connection. Inappropriately matching 2 surfaces can result in slippage and, ultimately, mechanical failure, just as trying to use a drag slick on your SUV would lead to poor performance (Fig. 15).

The force applied to 2 surfaces also plays a role in the friction force developed. The friction force is directly proportional to the perpendicular force applied to the 2 surfaces modified with a constant, the coefficient of friction (μ), which represents the surface characteristics. This may be described mathematically by the equation:

$$f = \mu \times N,$$

where f is the friction force, μ is the coefficient of friction, and N is the perpendicular force applied to the 2 surfaces (Fig. 16).

In applying this concept to the design and use of fixed rigid spinal instrumentation connectors, the goal is to achieve the highest friction force. The friction force represents what must be overcome so as to move 2 touching objects in relation to each other. A secure connection thus requires a sufficiently high coefficient of friction and applied force to develop a friction force that resists any applied external forces.

Implant–Bone

Abutting Implant–Bone Interfaces

Abutting implants are typically used in the interbody region. This is because of the fact that an abutting implant must bear a load to be effective and most of the axial load is transmitted through the neutral axis. Interbody implants may be composed of bone, nonbone (eg, acrylic), or a combination of both (eg, interbody cages). Because most long-term interbody strategies depend at least partly on the achievement of bony fusion, this section focuses on bone as an implant.

Good "carpentry" of bone components is critical to optimizing bony fusion outcome. The creation and shaping of a mortise in the vertebral body are just as important as the precise fitting and shaping of an interbody bone graft in minimizing the chance of dislodgment and other forms of

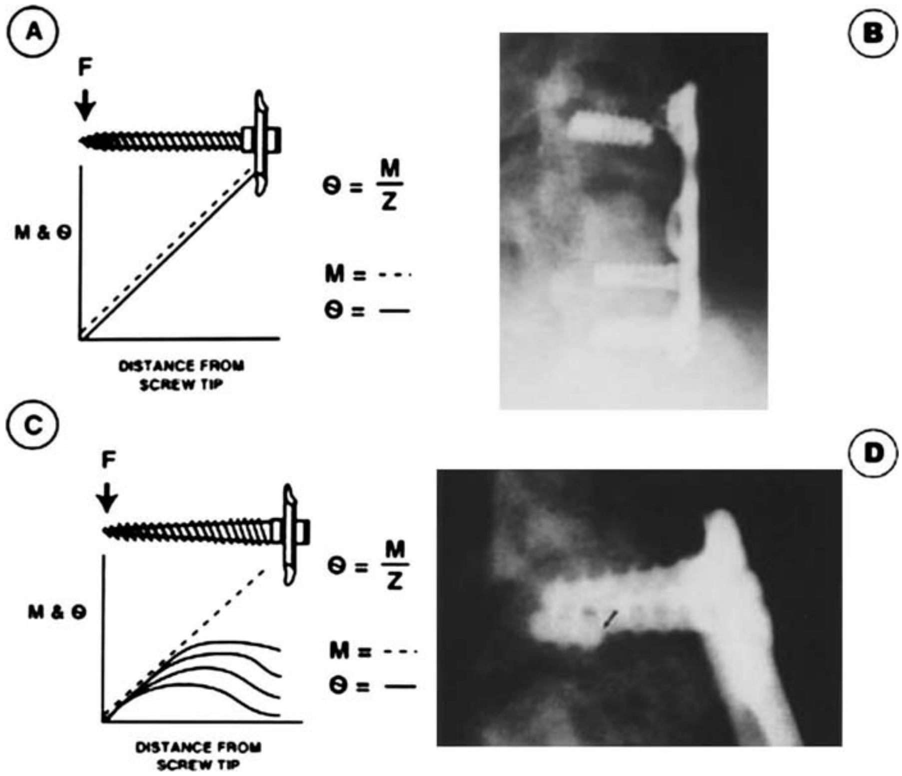

FIGURE 12. Example of the relation between stress and the strength of an object (eg, screw). A, Screw with a constant inner diameter attached to a plate in a fixed moment arm cantilevered manner is exposed to a load. This is associated with a bending moment that linearly increases along the screw from its tip (point of force application) to the plate (dotted line). Because stress (θ) is defined as $\frac{M}{Z}$ ($\theta = \frac{M}{Z}$) and the inner diameter of the screw (denominator of the stress equation) is a constant, the stress also increases linearly as one passes along the screw toward the plate (solid line). The stress is thus maximal at its junction with the plate. B, If the screw were to fracture, it would fracture at this juncture. C, This scenario is altered if the inner diameter of the screw is ramped (conically shaped). The bending moment still increases linearly (dotted line); however, because stress (θ) equals $\frac{M}{Z}$ and Z (ie, strength) is proportional to the third power of the inner diameter of the screw, the denominator of the equation increases "exponentially" as one passes along the screw. A relatively complex relation is therefore established between the resultant stress and the location along the screw. This depends on the "rate of change" of the screw's inner diameter. A family of curves (solid lines) dependent on the extent of the ramp of the inner diameter is thus generated. D, Clinical example of screw failure of a ramped inner-diameter screw is depicted (arrow designates fracture site). Note that failure occurs between the tip of the screw and its attachment site to the plate. If failure occurs, it occurs at the point of maximum stress application (apex of the curve describing the stress/location relation for the particular screw; solid line). M indicates bending moment; Z, section modulus (strength); θ, stress. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;27.

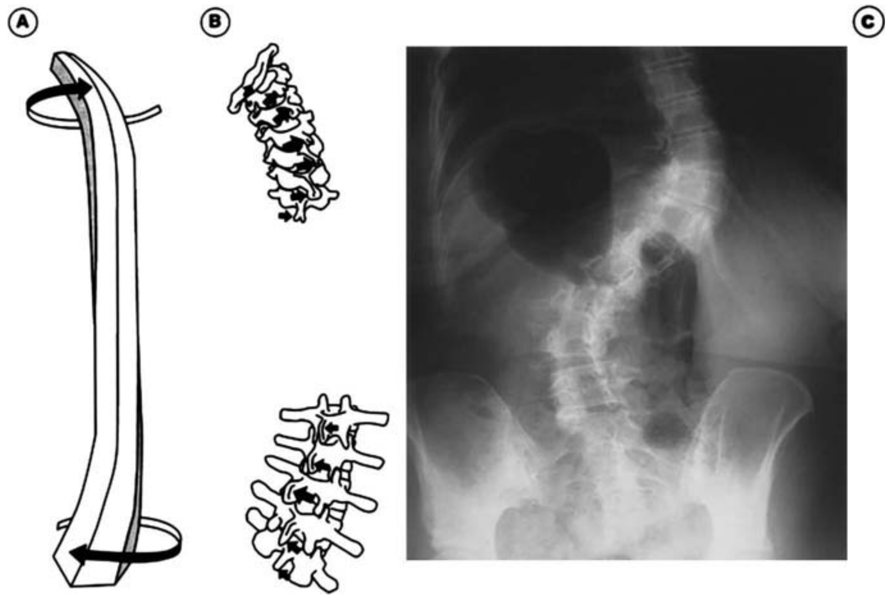

FIGURE 13. Perhaps the most important manifestation of the coupling phenomenon is the relation between lateral bending and rotation in the cervical and lumbar regions. This is depicted diagrammatically (A) and anatomically (B). Note that the coupling phenomenon results in spinal rotation in opposite directions in these 2 regions. C, Biconcave thoracic and lumbar curve, as depicted in an anteroposterior view, illustrates this phenomenon. Note that the lumbar spinous processes are rotated toward the concave side of the curve. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;28.

failure. Three factors directly affect the incidence and extent of subsidence in this regard: 1) the closeness of fit of the bone graft in the vertebral body mortise, 2) the surface area of contact between the bone graft and vertebral body, and 3) the character or quality of the contact surfaces.

Closeness of Fit

Just as we learn in childhood that square pegs do not set firmly in round holes, squared-off bone grafts do not fit well in a round mortise, or vice versa. A poor fit increases the likelihood of 2 types of adverse outcomes: nonunion because of an inadequate surface area of contact (Fig. 17A) and excessive subsidence because of the concentration of stresses and loads at the points of contact between the vertebral body and the bone graft (see Fig. 17B). Conversely, maximizing the surface area of contact and optimizing the closeness of fit between the bone graft and the vertebral body minimize stress concentration and thus minimize the chance of nonunion or excessive subsidence (see Fig. 17C).

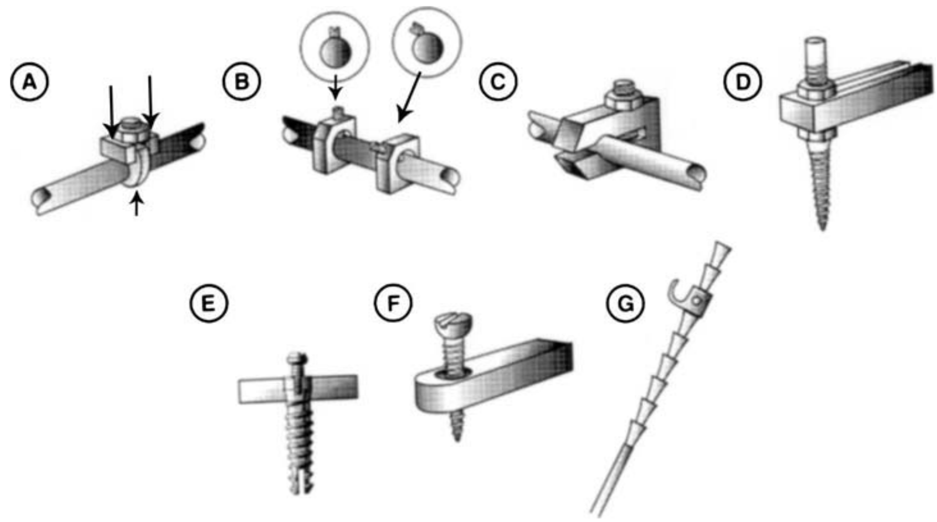

FIGURE 14. Seven fundamental component–component locking mechanisms. A, Three-point shear clamp. B, Lock screw: end on (left) and tangential (right). C, Circumferential grip. D, Constrained bolt–plate. E, Constrained screw–plate. F, Semiconstrained screw–plate. G, Semiconstrained component–rod. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;143.

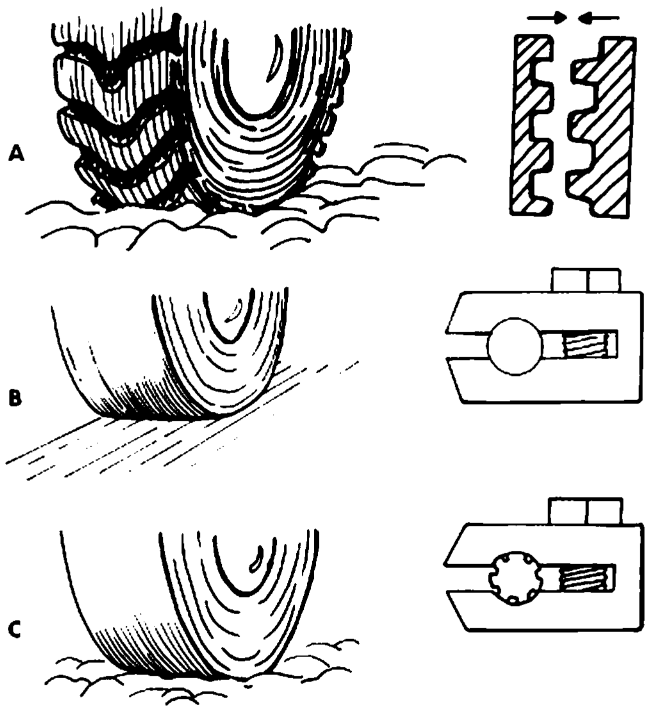

FIGURE 15. Two opposing surfaces of a component–component interface must match if the security of fixation is to be optimized. In each portion of this figure, a tire versus terrain analogy is depicted on the left and the component–component relation is depicted on the right (see text). A, Mud tire on off-road terrain and grid-grid interface. Note the meshing of the 2 surfaces. B, Racing slick on an asphalt road and a circumferential grip connector on a smooth rod. In A and B, the surfaces are matched and contact between the surfaces is optimal. If a mismatch of surfaces exists, the surface area of contact is diminished. C, This is depicted with a racing slick interfacing with an off-road terrain and a knurled rod interfacing with a smooth component. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;149.

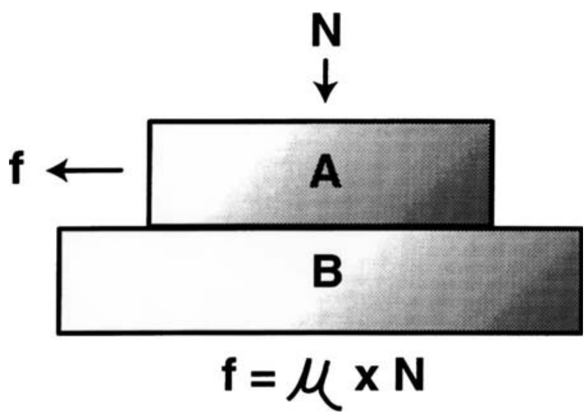

FIGURE 16. Force (N) applied to 2 surfaces in a perpendicular manner causes the friction between the 2 surfaces (f) to be increased proportionally. The only other factor involved is the coefficient of friction (μ). This describes the frictional relation between the 2 surfaces. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;149.

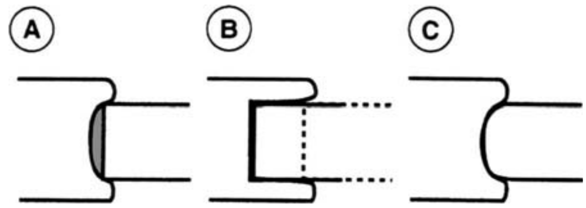

FIGURE 17. Square (flat)-ended bone graft does not "fit well" into a rounded vertebral body mortise. A, Nonunion may thus occur because of an inadequate surface area of contact (shaded region). B, Alternatively, subsidence, which may not be desirable, may occur because of the inability of the vertebral body mortise to prevent "penetration" by the ill-fitted strut graft. C, Maximizing the surface area of contact and the closeness of fit between the bone graft and the vertebral body minimize concentration of stress and the chance of nonunion or excessive subsidence. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;447.

Surface Area of Contact

The extent of subsidence is inversely proportional to the surface area of contact between the bone graft and the vertebral body (Fig. 18). The larger the surface area of contact, the less is the subsidence, and vice versa. An illustrative example is that the force required to penetrate a Styrofoam block with the eraser end of a pencil is much greater than that required for the sharpened end.

Quality of the Contact Surfaces

Two qualities determine the efficacy of the contact surfaces: the extent of end plate preservation and the proximity of the point of contact to the edge of the vertebral body. Some studies have shown that simply burring the end plate results in higher fusion rates without increasing the degree of clinically significant settling.[39,40] Anyone who has stood on an aluminum soda can has realized the power of the "boundary effect." The vertebral body cortex is superior to the softer inner cancellous bone at bearing axial loads. The size of the load that may be borne when all or a portion of the load is placed over the vertebral body margin is dramatically increased by the buttressing effect of the cortical boundary. Thus, the greatest biomechanical advantage would be achieved with a bone graft

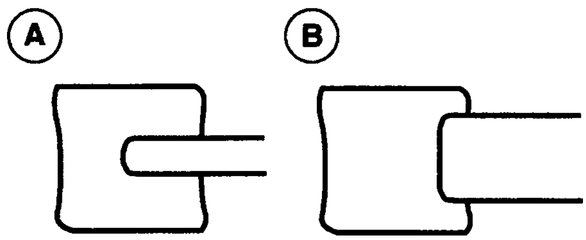

FIGURE 18. Surface area of contact between the bone graft and the accepting bone (vertebral body) is inversely proportional to the extent of penetration (subsidence) of the bone graft. A smaller surface area of contact (A) resists penetration less well than a large surface area of contact (B). Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;227.

that contacts the entire vertebral body surface and is the same size, contacting the entire cortical margin.

Penetrating Implant–Bone Interfaces

Penetrating implant bone interfaces can be broken down into 2 general categories: those with pullout resistance (eg, screws) and those without pullout resistance (eg, nails). The former may be further subdivided into those whose shape is constant and those whose configuration is altered on placement (eg, expanding tip screws). The constant-shape screw is not only the most frequently used clinically but has the most biomechanical data available, and is thus the focus of this section.

Screw Anatomy

Every screw has 4 basic components that can be altered alone or in combination to achieve different clinical results: the head, the core, the thread, and the tip (Fig. 19).

Head

The screw head is designed to resist translational forces as it engages the underlying surface and is terminally tightened. The type of underlying surface dictates the optimal size of the head. Soft cancellous bone requires a larger head than hard cortical bone. Metal surfaces allow the use of even smaller heads. The shape of the undersurface of the head often dictates how the screw/plate system functions. A rounded undersurface permits toggling and thus creates a dynamic or semiconstrained implant. A flat undersurface has the opposite effect. Once it has engaged the underlying surface, overtightening of the screw can result in either of 2 failures: stripping or pullout and deformation of the underlying surface.

Core

Under clinical conditions, screws are often required to bear substantial cantilevered loads (loads oriented perpendicular to the long axis of the screw). Bending strength is defined by the equation:

$$Z = \pi \times D^3/32,$$

where D is the core diameter. Fracture resistance is then proportional to the cube of the core diameter. For the commonly used diameters of 5.0 mm and 6.0 mm, this translates to nearly a 2-fold increase in relative strength: 125 and 216, respectively. Thus, all other things being equal, it is important to use the largest diameter screw allowed by the bony anatomy.[41]

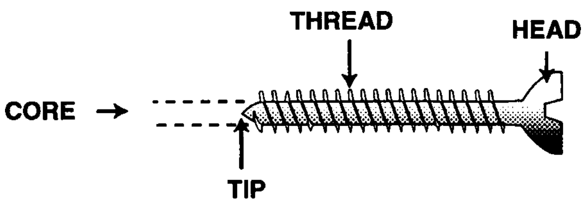

FIGURE 19. Important anatomic aspects and characteristics of a screw: the head, the core, the thread, and the tip. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;158.

Thread and Tip

The outer diameter and, more importantly, the thread depth are the main determinants of pullout resistance. There are 3 basic types of screw thread configurations: cortical, self-tapping, and cancellous.

Cortical bone is relatively incompressible and is thus subject to pathologic compression during screw insertion. Cortical screws have a shallow thread to minimize any compression. An added measure is pretapping the hole. A screw tap has cutting edges that carve threads into the wall of the bone and a full-length flute to help gather and remove bone debris.

Self-tapping screws have a leading edge flute and accomplish the aforementioned in a single step. Pretapped cortical and self-tapping screws have similar pullout strengths when used properly. It is of note that when used in cortical bone, both screw types may be inserted and removed several times without an adverse impact on their pullout strength.[42]

Cancellous bone is softer, and cancellous screws have deep threads that compress the bone during insertion. Pretapping cancellous bone only weakens the implant–bone interface.

Pullout Resistance

The 2 most important determinants of screw pullout resistance are the major screw diameter[43] and thread depth. Cortical purchase, depth of screw penetration, and thread design also play a significant role. Most of the pullout load is transmitted through the cortical bone to the first few threads adjacent to the screw head. Although this underscores the importance of proximal cortical purchase, opposite cortical purchase has been shown to be less of a factor.[44,45]

Fundamental to the concept of thread design is that pullout resistance is, as a general rule, proportional to the volume of bone between the screw threads. The pitch and shape of the thread determine this volume (Fig. 20). The thread pitch is the distance from a particular point on a thread to the corresponding point on the next thread. This is also known as the lead, or the distance the screw travels in a single turn.

The manner in which the screw hole is prepared can have a significant impact on pullout resistance. Using an awl to compress and compact the cancellous bone as opposed to simple drilling increases pullout resistance. Expandable tip screws are another possibility. Conversely, cortex and pilot hole overdrilling has a negative impact on pullout resistance.

Triangulation

This is the effect achieved through the use of rigidly connected diverging or converging screws. Pullout strength is optimal when the screws are oriented at 90° to each other[41]; however, the local geometry may dictate limitations. The triangulation effect is proportional to the triangular area below the screw. This area also represents the area of bone that would be "extracted" in pullout failure (Fig. 21).

Mechanical Attributes of Spinal Implants

Successful construct design mandates an understanding of the biomechanical forces applied to the spine by the implant. In reality, the resultant force vectors are often extremely complex. Fortunately, these forces may be broken down into

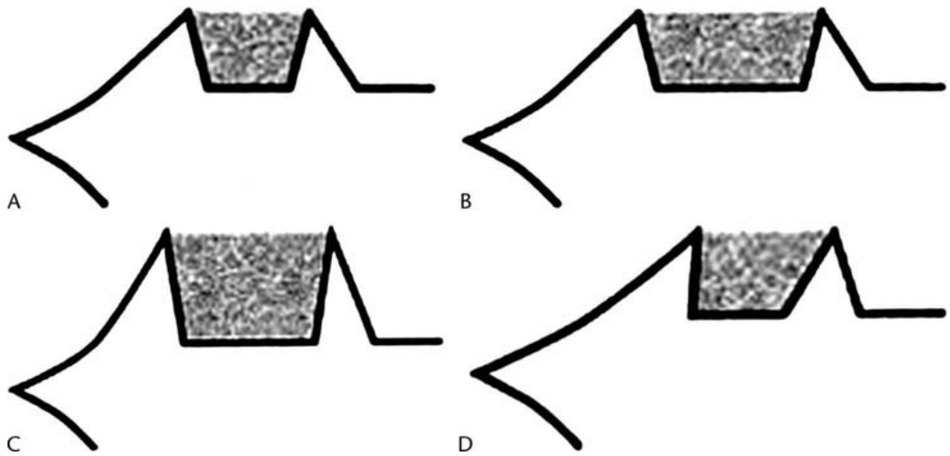

FIGURE 20. Methods of minimizing screw pullout. A, Screw pullout resistance is mainly a function of the volume of bone (shaded area) between screw threads. B, Thread pitch affects this by altering interthread distance. C, Thread depth affects this by altering thread penetration into bone. Thread shape affects this by altering the amount of bone volume directly. If the pitch and depth are unchanged, the only factor that can affect bone volume is screw thread volume (metal volume). D, Decreasing screw thread volume (metal volume) increases bone volume. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization.* AANS Publications; 2001;161.

component force vectors, facilitating an understanding while maintaining an accurate first-order approximation.

When designing a construct, 2 fundamental biomechanical concepts form the framework for understanding how implant forces interact with the spine: IAR and the neutral axis.[9] As previously stated, the IAR is the axis about which each vertebral segment rotates at any given instant in time. When a force is applied at a finite distance (known as the moment arm) from the IAR and perpendicular to the long axis of the spine, a bending moment is created (Fig. 22A). The power of the bending moment may be used to correct deformity and restore stability. Conversely, in the case in which there is minimal deformation and axial loading is the paramount concern, a graft should be placed in line with the neutral axis (see Fig. 22B).

Spinal instrumentation applies forces to the spine via 1 or more of the following basic mechanisms: 1) simple distraction, 2) 3-point bending, 3) tension-band fixation (TBF), 4) fixed moment arm cantilever beam, 5) nonfixed moment arm cantilever beam, and 6) applied moment arm cantilever beam. All these may be implemented through ventral, dorsal, or lateral approaches.

Simple Distraction

As previously mentioned, when a force is applied at a distance that is perpendicular to the IAR, a bending moment is

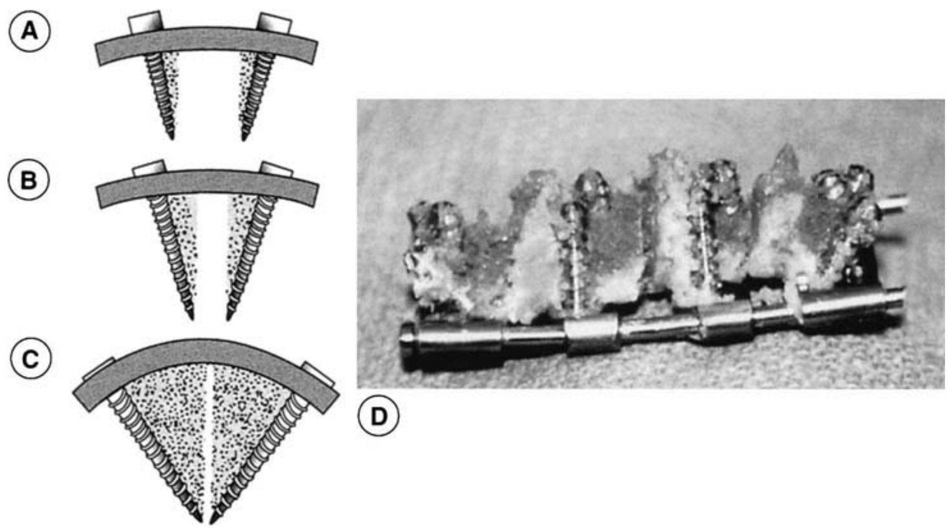

FIGURE 21. A, Triangulation effect is proportional to the shaded area subtended by the screw. The shaded area can be increased by lengthening the screws (B) or by altering the trajectory (C). D, When a triangulated screw implant is pulled out, a significant quantity of bone is extracted with the implant. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization.* AANS Publications; 2001;162.

created. Simple distraction is usually applied through ventral interbody or dorsal strategies. If a ventral implant that is in line with the neutral axis is in place, it is positioned to resist axial loads without producing a bending moment. If it is placed ventral to the neutral axis, however, extension results. A dorsally placed distraction force would tend to promote kyphosis and thus has little clinical application by itself. It may be combined with 3-point bending instrumentation in which the central ventrally directed force acts as a fulcrum.

Three-Point Bending

Three-point bending constructs usually involve the placement of dorsal instrumentation, typically over 5 or more segments. The 2 ends of the implant impart dorsally directed forces that are balanced by a ventrally directed force equal in magnitude to the sum of the 2 terminal forces. A common example is the springboard (Fig. 23A). The end support and the person at the opposite end represent downwardly directed forces countered by the upwardly directed force of the fulcrum (see Fig. 23B). Clinically, this strategy has been combined with distraction or compression using Harrington rods or universal spinal instrumentation techniques (see Fig. 23C).

Mathematically, the bending moment for a 3-point bending construct is defined by the equation:

$$M_{3PB} = D_1 \times D_2 \times F_{3PB}/D_{3PB},$$

where M_{3PB} is the 3-point bending moment, D_1 and D_2 are the distances from the fulcrum to the respective terminal fixation points, D_{3PB} is the sum of D_1 and D_2, and F_{3PB} is the ventrally directed force at the fulcrum. In most clinical scenarios, D_1 and D_2 are of relatively comparable lengths. A terminal 3-point bending construct, however, may be created by placing the fulcrum closer to either of the ends. This may be helpful in the clinical setting of sagittal deformation, where the rostral segment is translated in a ventral direction with respect to the more caudal segment (Fig. 24).[9,46]

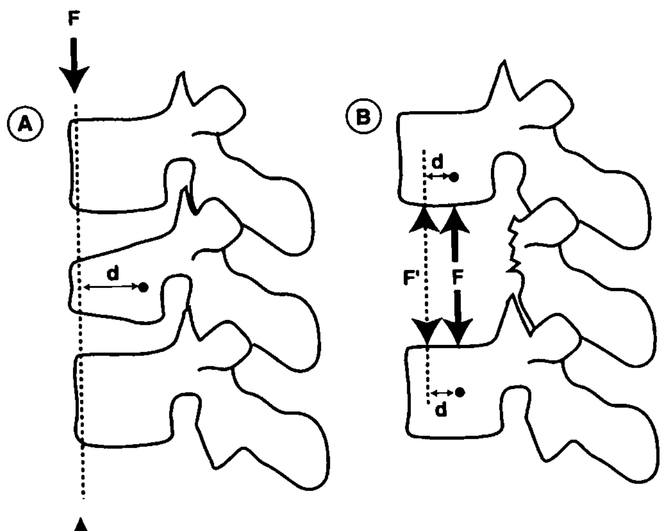

FIGURE 22. A, Compressive force (F) that is applied at a finite distance (d) from the instantaneous axis of rotation (IAR) (●). B, A distraction force (F) that is applied "in line" with the IAR (in the neutral axis) does not result in bending moment application. Distraction force (F′) that is applied at some distance (d) from the neutral axis causes a bending moment, the magnitude of which is dictated by the perpendicular distance (d) from the IAR. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization.* AANS Publications; 2001;189.

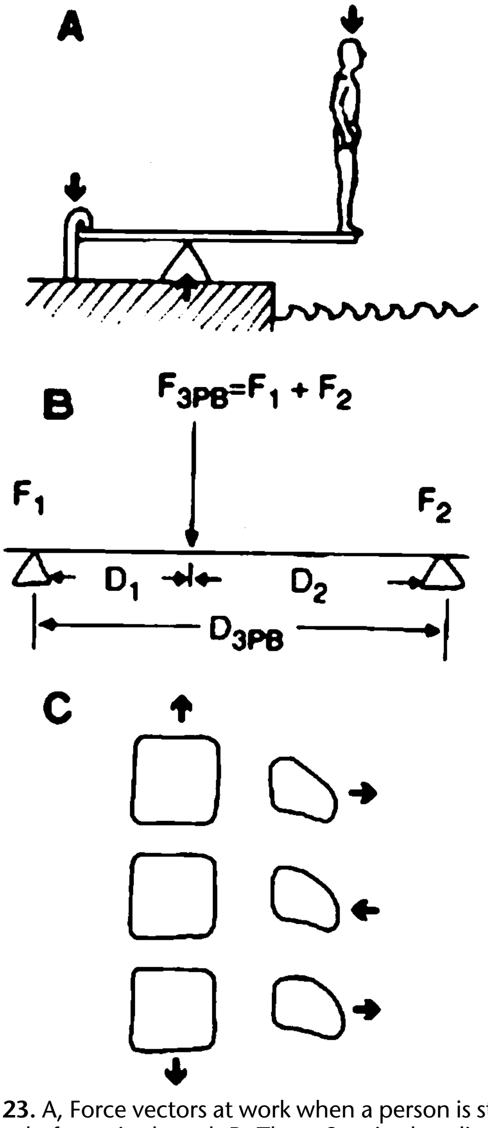

FIGURE 23. A, Force vectors at work when a person is standing on the end of a springboard. B, These 3-point bending forces are defined by the equation:

$$M = D_1 D_2 F_{3PB}/D_{3PB},$$

where D_1 and D_2 are the distances from the fulcrum to the terminal hook–bone interfaces, D_{3PB} is the sum of D_1 and D_2, and F_{3PB} is the ventrally directed force applied at the fulcrum. C, Spinal 3-point bending constructs (horizontal arrows) are usually applied in combination with another force vector complex, commonly distraction (vertical arrows). Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization.* AANS Publications; 2001;191.

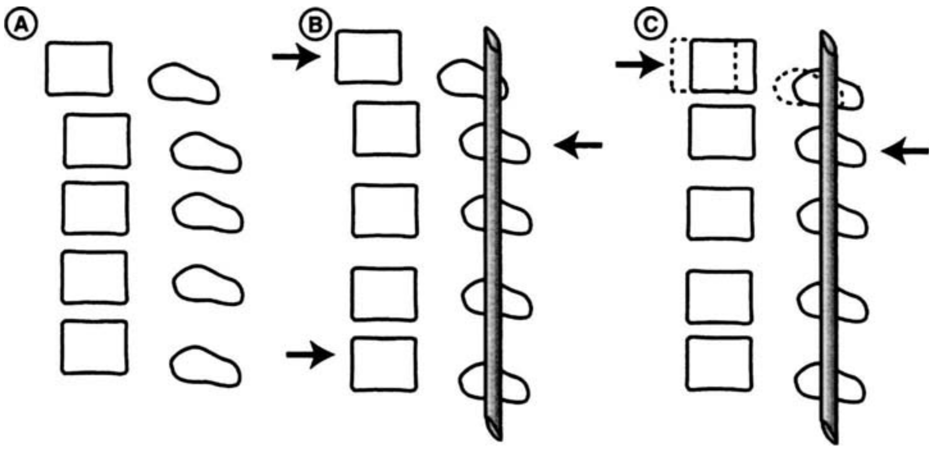

FIGURE 24. Terminal 3-point bending. A ventral translational deformation relative to the next most caudal segment (A) can be corrected by the application of 3-point bending forces to each of the 3 segments depicted (B). C, This results in translational deformation reduction. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;192.

Tension-Band Fixation

TBF (compression) may be applied via dorsal or ventral strategies using wires, clamps, springs, or rigid constructs. In addition to the compression of the spinal elements, a bending moment is created that produces extension (dorsal application) or flexion (ventral application) (Fig. 25). The resulting bending moment of a TBF construct may be mathematically defined by the following equation:

$$M_{TBF} = F_{TBF} \times D_{IAR \cdot TBF},$$

where M_{TBF} is the 3-point bending moment, F_{TBF} is the compressive force applied at the upper and lower limits of the instrument–bone interface, and $D_{IAR \cdot TBF}$ is the perpendicular distance from the IAR to the applied force, F_{TBF} (Fig. 26).[46]

Comparing 3-Point Bending and Tension-Band Fixation

The bending moment generated by a 3-point bending fixation construct is proportional to the length of the construct. As a result, this technique typically requires a long construct to

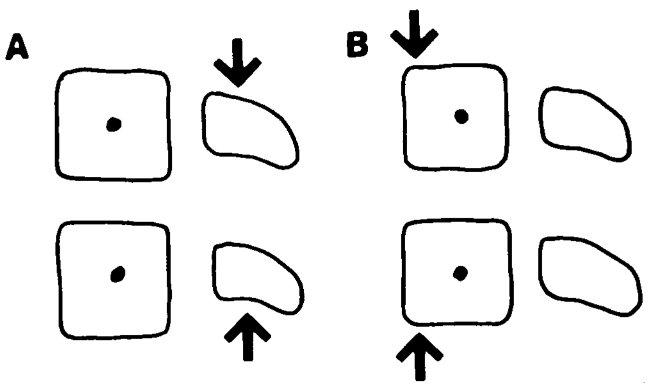

FIGURE 25. A, Dorsal spinal tension-band fixation (TBF). B, Ventral spinal TBF. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001; 192.

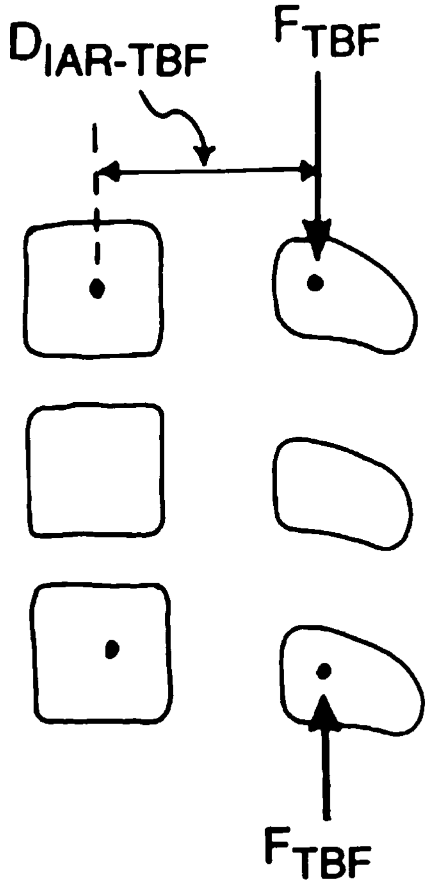

FIGURE 26. Forces applied by a tension-band fixation (TBF) construct are described by the equation:

$$M_{TBF} = F_{TBF} D_{IAR-TBF}$$

where M_{TBF} is the bending moment, F_{TBF} is the compression force applied at the upper and lower termini of the construct at the instrument–bone interface, and $D_{IAR-TBF}$ is the perpendicular distance from the instantaneous axis of rotation to the TBF-applied force vector. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;193.

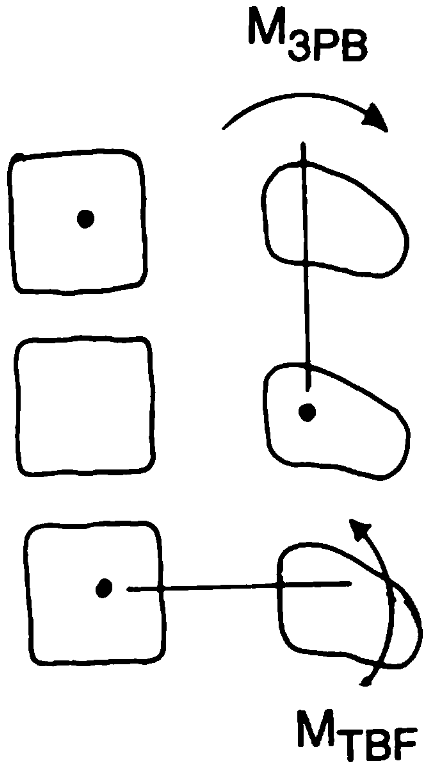

FIGURE 27. Moment arm applied by 3-point bending constructs (M_{3PB}) is parallel to the long axis of the spine, whereas that applied by tension-band fixation constructs (M_{TBF}) is perpendicular to the long axis of the spine. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization.* AANS Publications; 2001;195.

optimize its efficacy. Conversely, the bending moment generated by a TBF construct is independent of the construct length. This technique may be used over as few as 2 segments. It can be shown mathematically by simultaneously solving the bending moment equations for each technique: to achieve an equal bending moment, the moment arm of a 3-point bending fixation construct must be 4 times longer.[47] In fairness, it should be clarified that the moment arms are not equivalent with regard to their orientation. The moment arm in a 3-point bending construct is parallel to the long axis of the spine, whereas in a TBF fixation construct, the moment arm is perpendicular to the long axis of the spine (Fig. 27).

Fixed Moment Arm Cantilever Beam

A cantilever beam is a structure designed to bear a load while spanning a space. The beam is supported only at a single end. The supported end may be attached in 1 of 3 ways: fixed moment arm, nonfixed moment arm, or an applied moment arm. A fixed moment arm cantilever beam, the simplest configuration, is rigidly attached at a single end (Fig. 28A). A clinical example is a rigid pedicle screw construct (see Fig. 28B). This application provides rigid fixation with a relatively short moment arm. Although typically applied in a neutral mode at the time of surgery, the axial loading resulting from the assumption of an upright posture produces a bending moment that is maximal at the screw–rod/plate interface. The resulting stress may be sufficient to cause screw fracture, especially in a screw with a constant inner diameter (see Fig. 28C).[48]

Nonfixed Moment Arm Cantilever Beam

In this application, the cantilever beam is allowed to toggle at its attachment point (Fig. 29A). This has 2 important

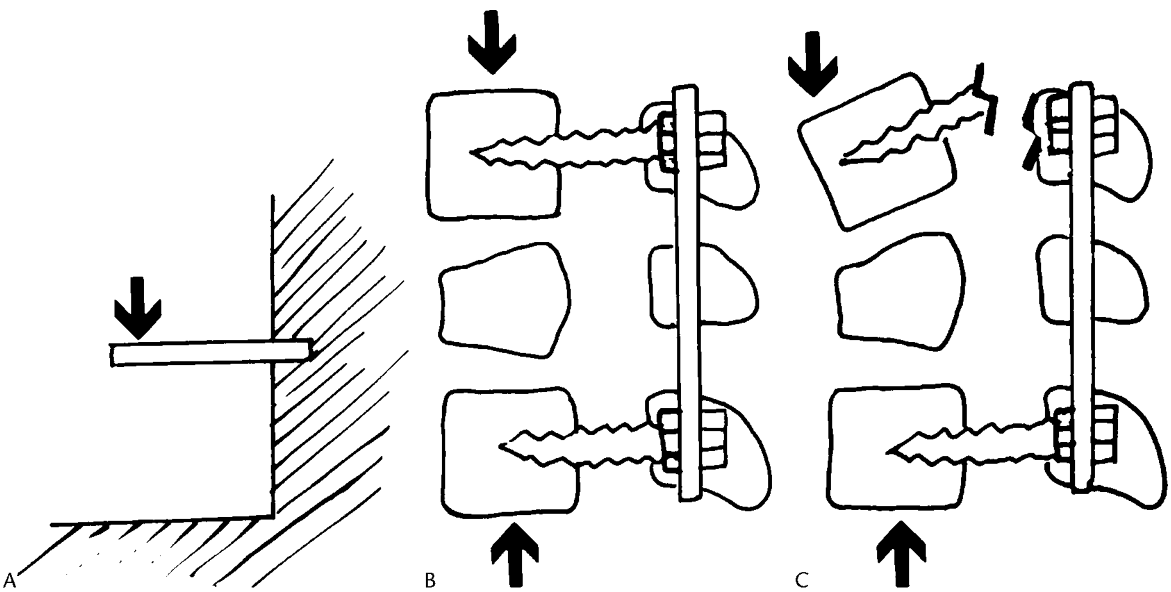

FIGURE 28. Fixed moment arm cantilever beam. In this case, the cantilever beam is rigidly affixed to the wall. A, Note the lack of need for an accompanying applied force vector during the bearing of a load (arrow). B, Stress realized by a fixed moment arm cantilever beam during load bearing is often maximal at the screw–plate or screw–rod interface. C, This may result in construct failure at this location after the bearing of an axial load (arrows). Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization.* AANS Publications; 2001;196.

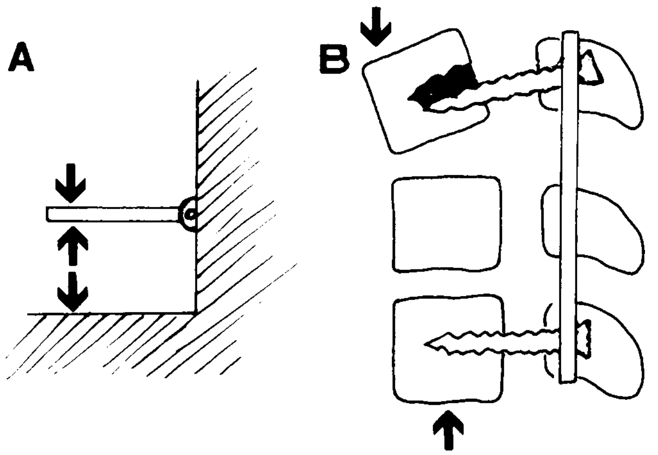

FIGURE 29. Nonfixed moment arm cantilever beam. A, In this case, the cantilever beam is fixed by a hinge to the wall. Note the requirement for an accompanying applied force vector (opposed arrows) during the bearing of a load (single arrow). B, Nonfixed moment arm cantilever beam constructs may fail by screw pullout, as depicted. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;197.

consequences. First, little if any bending moment is applied to the spine. Second, this instrumentation has little intrinsic ability to bear an axial load and thus must be used when the vertebral column is intact or in conjunction with another structure that has load-bearing capability (eg, cage, interbody

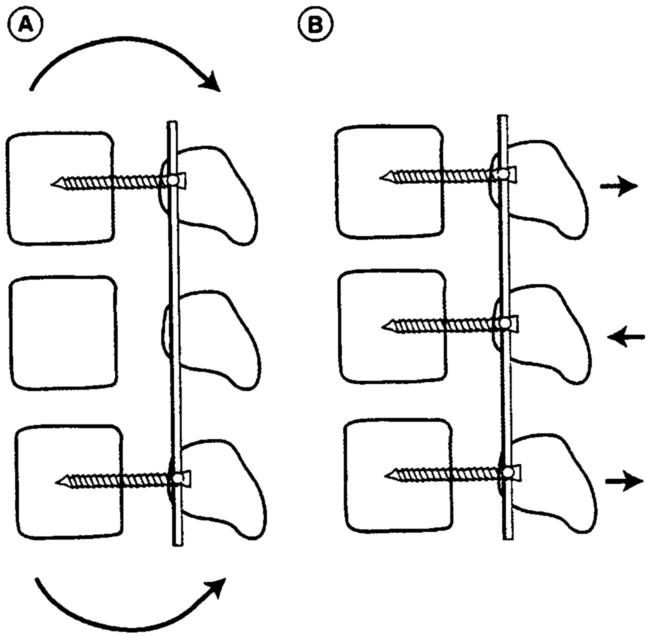

FIGURE 30. Nonfixed moment arm cantilever beam constructs can function in a tension-band fixation mode (A) by resisting flexion via the application of a bending moment (curved arrows) or in a 3-point bending mode (B) (straight arrows). Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;197.

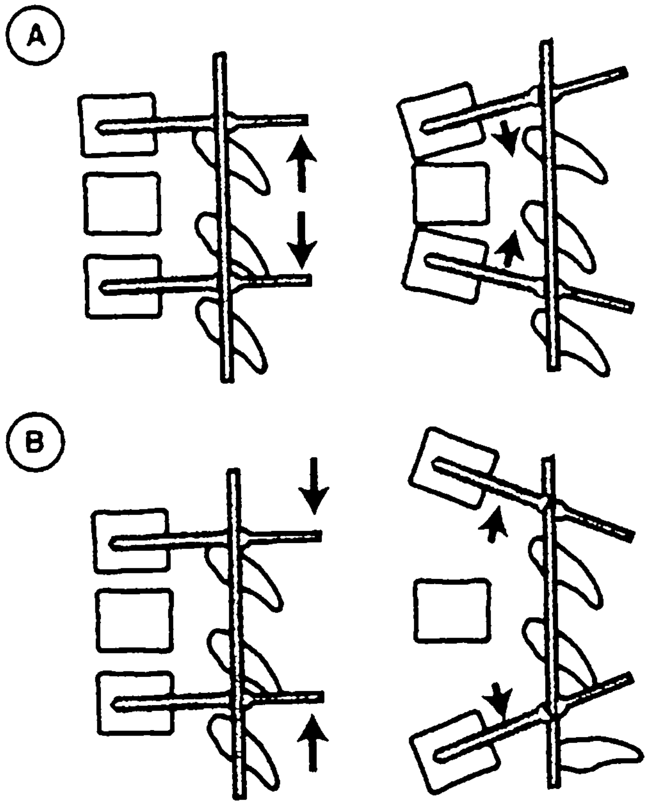

FIGURE 31. A, Applied moment arm cantilever beam construct using a flexion moment. B, Applied moment arm cantilever beam construct using an extension moment. Reprinted with permission from Benzel EC. *Biomechanics of Spine Stabilization*. AANS Publications; 2001;197.

bone graft). Permitting the screw to toggle results in this construct having a weak screw pullout resistance (see Fig. 29B). This is most pronounced in soft bone and may be offset somewhat by maximizing the amount of cortical bone purchase.

In the clinical setting, nonfixed moment arm cantilever beams may function much like TBF constructs when the spine is flexed (Fig. 30A). With good screw purchase, they may also function as a 3-point bending construct (see Fig. 30B).

Applied Moment Arm Cantilever Beam

The ability of a cantilever beam to impart a bending moment to the spine provides great utility in the correction of deformity. A flexion (Fig. 31A) or extension (see Fig. 31B) moment may be delivered to the spine, with extension being the most common clinical scenario.

CONCLUSION

Although the last 25 years have seen tremendous advances in the field of spine surgery, one must remember that its roots go back as far as 4 centuries. Having gained some historical perspective, we tried to impart a working knowledge of the pertinent anatomy of the vertebral column. Most construct failures are the result of poor implant and/or patient selection rather than actual device failures. This underscores the importance of being able to apply basic biomechanical principles to the clinical arena. We have tried to emphasize this by introducing some of the fundamental biomechanical principles and concluding with some relevant clinical applications in construct design. We hope this information provides a framework that the surgeon can use in the clinical decision-making process.

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