Lumbosacral and Spinopelvic Fixation

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CHAPTER 18

Instrumented Fusions of the Lumbosacral Spine: A Technical Overview

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THE ROLE OF FUSION

In order to assess the role of instrumentation at the lumbosacral junction, it is necessary to examine the indications for surgery at this transitional region. There are numerous techniques and implants available to decrease the unacceptably high pseudoarthrosis rates of noninstrumented fusions.[12,16,57] The purpose of this chapter is to review the biomechanics of the various spinal implants available for stabilization and reconstructive procedures.

Fusion is the definitive treatment for instability. A "fracture milieu" is created by decortication of bony surfaces, and the pathologic motion between adjacent segments is eliminated.

THE ROLE OF INSTRUMENTATION

Instrumentation provides stable fixation in a predetermined position until fusion occurs. Fusion of the spine to the pelvis is a considerable challenge because it involves large lever arms. The bone–implant interface must be strong enough to withstand the physiologic loads multiplied by these lever arms. Wilkins[66] first used spinal instrumentation in 1887. Boucher[8] improved on the work of King[34] in 1959, giving us the predecessor of today's pedicle screw. However, it was not until in the late 1950s, when Harrington[28] began developing his hook–rod system, that the concept of spinal instrumentation came into vogue. To the present day, the Harrington rod system still represents the basic principle of rod–hook systems.[29] The original facet screw fixation system evolved into the pedicular screws that later were engaged into the rod–hook systems.

Instrumentation is used in conjunction with biologic substrates (i.e., bone graft), and, as Hibbs stated, "There is no substitute for a solid bony fusion." Likewise, spinal constructs are only as strong as the bone to which they are applied.

PRINCIPLES OF CORRECTION AND STABILIZATION

Spinal Stability

The stability of the spine with respect to yielding and probable fracture can be predicted if the mechanical behavior is known. The determinants of mechanical behavior vary among individuals; therefore, it is impossible to predict the response to abnormal loading.[38] A universal definition of instability that includes all variables of mechanical behavior does not exist. The following is a review of current definitions.

Pope and Panjabi[49] defined an unstable structure as one that is not in an optimal state of equilibrium. The stability of the spine is affected by restraining structures that, if damaged or lax, alter the equilibrium and therefore increase instability.[49] A spinal column that is able to maintain alignment when subjected to physiologic loads is considered stable. If displacement of the spinal column, either acutely or during the healing phase of a fracture, is likely to occur, then the spinal column is considered unstable.[50] The American Academy of Orthopedic Surgeons defined segmental instability as an abnormal response to applied loads, char-

acterized by motion in the motor segment beyond normal constraints.[22,46]

If the spine under physiologic loads is able to limit patterns of displacement so as not to damage or irritate the spinal cord and nerve roots, then, according to White and Panjabi,[67] the spine is stable. Holdsworth and Hardy,[31] in 1953, classified osseous and neurologic injuries, clarifying the need for operative and nonoperative treatment based on stability. Whitesides and Shah[65] defined stability as the ability to withstand stress without progressive deformity or further neurologic damage. Most definitions of stability include the effect of dynamic loading as well as the presence of deformation over time. The radiograph taken in the supine position does not reflect the position of the vertebrae under a physiologic load, nor does it reveal a deformity that may exist in the future.[25]

Mechanical stability is a function of the integrity of the bony and soft tissue structures that compose the spinal column. These structures include the cortical and cancellous bone of the vertebral body, pedicles, facet joints and posterior arch of the lamina, and spinous processes. The soft tissue elements include the anterior and posterior longitudinal ligaments, the annulus and nucleus of the disc, the facet capsules, the interspinous and supraspinous ligaments, and the supporting musculature of the spine.[30] It is difficult to restore stability at the level of the motion segment. The usual solution is to fuse the unhealthy motion segment, sacrificing its mobility, so as to restore function to the spine as a whole.

Principles of Stabilization

Haher and colleagues[27] demonstrated that mechanical damage to one of the columns of the spine causes migration of the instantaneous axis of rotation (IAR) to another position within the motion segment. A displaced IAR is the major sign of instability and is therefore an indication for treatment. Regardless of how one defines the spinal columns, it is important to augment the deranged column or columns and to restore the mechanical stability of the spine.

Load-deflection curves have been generated by subjecting the normal spine to physiologic loads in compression. These curves were established as a norm for the intact lumbar spine. The same curves were generated in compromised lumbar spines in which a specific, predetermined column was destroyed. The overlay graphs were plotted, and the deflection for a specific load was compared. The ability of the spine to support a physiologic load is called the load carrying capacity of the lumbar spine.[19]

The columns of the spine support the weight of the head, upper extremities, and trunk. If columns are destroyed, the ability to support that load is decreased. Mechanical instability results, and the need for surgical stabilization is established.[25]

The spine must resist torsion when the trunk and upper extremities are subjected to twisting while the lower extremities remain in a fixed position. To study the ability of the spine to resist torsion, torsion–rotation curves have been generated for lumbar spines within physiologic limits. Multiple motion segments were used to preserve the normal sagittal curves of the spine, thereby allowing the lumbar spine to establish its physiologic axis of rotation. The data generated for the intact spines were used as a control. These data were then compared with those obtained from compromised specimens with single- or multiple-column destruction. The data reflect the loss of torsional rigidity. With destruction of the anterior column, the spine is unable to resist torsion, with 95% loss of rigidity. Destruction of the facet joint does not significantly alter the torsional resistance. The ability of the lumbar spine to resist rotation is severely compromised if the anterior column is destroyed. The ability of the annulus to resist rotation is greater than that of the facet joint alone or in combination with any other structures.

In normal spines, the IAR is located posterior to the annulus in the vicinity of the facet joints.[26] With destruction of the annulus, the IAR migrates posteriorly. Destruction of the posterior elements causes the IAR to migrate anteriorly. Those structures farther from the axis of rotation in the lumbar spine have the greatest effect on resisting rotation. The anterior column is the structure responsible for controlling rotation in the intact spine.[27] Loss of the anterior column renders the lumbar spine incapable of resisting torsion. Mechanical instability results, and the need for surgical stabilization is established. It has also been shown that the IAR centers on the spine–instrumentation interface. For example, in a posterior pedicle screw and rod construct, the IAR migrates to the screw–rod junction.

Principles of Correction

In spite of the many revolutionary ideas set forth by Harrington, the system addressed deformities only in the coronal plane. This was accomplished by distracting the scoliotic spine between two hooks. Kyphosis at the distal hook often resulted. In an attempt to provide correction in the coronal and sagittal planes, Luque developed a sublaminar wiring technique. This technique provided segmental control with multiple points of fixation. Although the technique was safe in the hands of most spinal surgeons, fear of potential neurologic injury precluded its use without proper scientific evidence. A combination of Harrington and Luque principles was widely practiced in the form of the Wisconsin

construct. This construct eliminated the use of sublaminar wires yet provided segmental fixation with spinous process wiring.

Since its introduction in 1984, Cotrel and Dubousset instrumentation[14] has become the standard segmental instrumentation system. It provides for multiple points of fixation and less risk of neurologic injury than sublaminar wiring. It allows for correction in the coronal and sagittal planes; however, the transverse correction that is achieved by "derotation" is more of a global type than a segmental type. This system allows for application of distractive and compressive forces along the same rod as a means of control of coronal and sagittal plane deformities. The principle proposed by Cotrel and Dubousset—rotation not translation—is not uniformly accepted, and other approaches to curve correction and implant development have been tried.

Control over each segment in the construct is the crucial issue in correction. The techniques for segmental control evolved through a combination of segmental hooks and wires, claws, and, ultimately, pedicular screws, which provide control in six degrees of freedom. The objective is to achieve a physiologic spinal contour. Each segment is translated and prompted to previously bent rods. This philosophy is best illustrated by the ISOLA[4] instrumentation system and the Universal spine systems.[1] The use of stiff but ductile rods may achieve a physiologic spinal contour by in situ bending to bring the spine to the "normal" contour. This technique accentuates placing the rods in such a way that at the end of the correction maneuver, the center of rotation of the whole assembly, which is now in the rods, is as close as possible to the normal center of rotation. The ductility of the rod is crucial to the success, as Jackson has stated.[33] Most of the posterior systems use the implants themselves or the ancillary instruments as tools for translating the spine.

The anterior approach for correction of spinal deformities was introduced in the 1930s to correct spondylolisthesis by Capener.[9a] Control over spinal deformity may be gained by resection of discs or bone. The primary principle is that by changing the relative lengths of the spinal columns, one can change sagittal alignment. Shortening of the anterior column is kyphosing, and lengthening is lordosing. Resection or insertion of a wedge can align scoliosis, and structural grafts or implants can restore height and the normal relations between vertebrae. Luque and Heinig improved on the anterior approach by "limberizing" the spine. A similar principle of release is used posteriorly and is called the "eggshell" procedure.

Yet another philosophy of correction, presented by Harms (see Chap. 41), is based on his posterior tension-band and anterior load-sharing principles. Relatively thin posterior compression implants, combined with anterior solid spacers, may lead to a solid fusion in the desired position, with 80% of the load applied to the anterior column and 20% to the posterior column. The main logical application of this concept is to kyphotic deformities.

Another important principle of correction is the restoration of disc height. If done properly, it allows indirect opening and decompression of the foramina. This can be achieved by posterior or anterior lumbar interbody fusion or posteriorly via destruction and in situ bending of posterior-screw-based construct. This maneuver depends on the integrity of the ligaments across the motion segment, and the lack of structural anterior support may be a disadvantage.

QUALITIES OF COMBINED BONE AND IMPLANT CONSTRUCTS

Instrumentation devices that are used to fuse the spine to the pelvis are composed of three basic elements: modalities of fixation to the spine, modalities of fixation to the pelvis, and longitudinal connecting members. (Facet screws posteriorly or spacers anteriorly would be exceptions.)

There are three basic options to gain purchase in the posterior elements of the spine: hooks and claws, wires and cables, and screws. Large forces can be applied through a hook. Once hooks are engaged in a construct and locked, forces can be applied through them in other directions, acting on the entire instrumented segment. Wires may produce transverse traction to a straight rod or may achieve segmental fixation, enabling control of additional degrees of freedom. Pedicular screws are the means that allow control of the spine to all degrees of freedom with substantial application of forces. Claws, either between two hooks or between a hook and a screw, and combinations of screws and sublaminar or sub-pars wires, are an augmentation of those means.

Fixation to the pelvis may include hooks and wires, as in the spine. Pedicular screws may be inserted into the sacral body or the ala. Probably the best place to insert a screw into the sacrum is to the body, toward the promontorium, reaching the upper end-plate of the sacrum for purchase. Iliosacral screws go from the ilium to the body of the sacrum. They engage at least three cortices, and they provide a strong foundation for a spinopelvic construct. The Luque-Galveston method engages the ilium in the construct. Double screws in the sacrum are used, as are combinations of multidirectional screws to the body of the sacrum, to the ala of the sacrum, and to the ilium. The Jackson "Iliac Buttress," which is based on rods that are inserted into the sacrum between its two cortices, under the wings of the ilium, provides a strong means of purchase. The rods are in-

serted through screws that are located, classically, at the upper part of the sacrum.

Rods or plates may be used to connect the screws. The various sizes, and the locking nature of the connectors, correspond to the philosophy behind the specific system being applied.

The transverse devices, such as the DDTs, DLTs, Crosslinks, or even Chopin plates, give the construct quadrilateral strength, which is fundamental in many systems.

Devices for the Front of the Spine

Insertion of anterior rods over L5 and S1 is difficult. An anterior plate placed over L5-S1 is occasionally used.

Surgical Techniques

The classic anterior and posterior surgical techniques for spine fusion are being challenged by arthroscopic minimal invasive surgery, either posteriorly, based on Kambin's (see Chap. 44) development of discoscopies, or anteriorly, as demonstrated by the BAK system (see Chap. 43). The advantage of this concept is clear, and the techniques of instrumentation and the implanted devices themselves will probably be modified and specially designed to be handled easily through an arthroscopic approach.

Bone Graft and Supplements

The role of bone graft in posterior fusions is clearly established and well known, and there is no difference between the lumbosacral junction and other places in the spine. However, the role of bone graft and spacers in the anterior part of the spine has been evaluated and appreciated only in recent years, after the importance of sagittal balance became apparent.

A variety of materials can be used as anterior struts, all of them accompanied by bone graft to achieve bony fusion. They can be made of metal or ceramic, allograft, or blocks of hydroxyapatite, new cements, or other materials. Likewise, a number of implants have been designed for anterior spinal instrumentation and fusion. Common to all these techniques is the placement of an anterior strut in compression. With time, bone graft incorporation leads to a shift from load sharing of the bone-implant construct to load-bearing by the bone.

Screws and Purchase of Implants in Bone

The advent of pedicle screws has revolutionized spinal surgery. They have increased the strength of spinal constructs without sacrificing motion segments. Pedicle screws allow for segmental control and stabilization of individual motion segments, with equal or fewer complications and improved fusion rates over previously available spinal instrumentation systems.[71]

The biomechanical behavior of screws comes primarily from animal studies in long tubular bones.[26,55,56,62,64] The tensile strength of screws is proportional to the square of their diameter, whereas their torsional strength is proportional to the cube of their diameter.[62] These two mechanical properties are limited by the mediolateral width of the pedicle. Several authors have examined this situation, with pedicular diameters in the lumbar spine ranging from 7 to 18 mm.[37,54,74] Diameters increased caudally, as would be expected. As a result, commercially available screws range from 4.5 to 8.5 mm and more in diameter. Likewise, various angles of insertion into the pedicle have been described.[37,43,53] The work of Saillant,[54] in agreement with that of other researchers,[37] demonstrates that an increased screw path length with an angled approach theoretically provides improved fixation through increased length of purchase.

The sacrum itself provides poor bony purchase. The sacral ala are two quadrangular structures that have a dearth of cancellous bone within them, especially in elderly persons. Screw purchase can be achieved through the pedicles of the sacrum, across the sacroiliac joint, or intrasacrally. Edwards and others[20] have studied this situation extensively, demonstrating both biomechanically and clinically that the medial (25°) sacroiliac and lateral (35°) S1 screw positions provide greater proximal sacral fixation over the standard 10° to 15° medial sacroiliac screw. Further, he showed that a converging S2 orientation provides superior midsacral fixation to that of the standard S2 position. DePeretti and Argenson[17] advised insertion of the screw through the pedicle and body of S1, with the point of insertion below and lateral to the articular process of S1 and an oblique course forward and inward at an angle of 10°. The work by Zindrick and colleagues[74] demonstrated S2 to be the weakest sacral insertion site if standard insertion techniques are used. They were able to quadruple mean pullout strength by angling the S2 screw 45° laterally. Similarly, they improved S1 pullout with lateral angulation.

Pullout strength is proportional to the shear strength of the material in which the screw is embedded multiplied by the outer diameter of the screw and by the length of the screw.[11,21] Zindrick and associates[74] believed that the bone quality of the specimen tested was the most important factor determining the ability of a screw to resist loosening. An and colleagues[2] demonstrated a statistically significant difference in screw loosening secondary to cyclic loading below a critical bone mineral density (BMD) value of 0.45 g/cm^2 and suggested that DEXA scanning may be of benefit in

predicting anterior vertebral screw fixation failure. The applicability to transpedicular screws is an area of ongoing research.

Current recommendations call for 80% depth of screw insertion. Biomechanical data show the 30% increase in grip strength to be statistically significant. Krag and coworkers[37] demonstrated 32.5% higher loads to cut-out with 6-mm screws at 80% insertion depth. In comparing various insertion depths for 4.5-mm cortical screws and 6.5-mm, fully-threaded screws, Zindrick and associates[74] showed no difference in pullout strength between 50% insertion and insertion to cortex for both screw types. They attributed this result to poor specimen quality. However, through-cortex insertion was significantly stronger than to-cortex insertion for both designs. This is contrary to work by Lavaste.[39] Zindrick and colleagues also examined depth of insertion versus screw loosening in regard to medial-lateral and cephalad-caudad cyclic loading, using 6.5-mm, fully-threaded cancellous screws. In all tests, increased depth of insertion significantly improved the number of cyclic loads necessary to achieve loosening.

Self-tapping screws provide somewhat greater holding power than screws that are inserted into tapped holes[3,56]; however, this advantage is offset by the loss of ability to generate compression. These data were obtained in long tubular bones and may not be applicable to vertebrae. Likewise, these authors demonstrated a linear relation between holding power and cortical thickness, as would be expected. Zdeblick and coworkers[73] demonstrated a linear relation between the insertional torque, during either tapping or inserting of a screw, and the number of cycles to screw pullout. Additionally, they found that an insertional torque of less than 4.0 in-lb led to earlier screw pullout.

Krag and associates[37] failed to demonstrate any significant biomechanical advantage between two thread types (V and B), in agreement with previous biomechanical studies.[35,56] However, they chose a buttress thread for the Vermont spinal fixator, citing ease of insertion and surgeon familiarity as the reasons.

THE RESPONSE OF BONE TO IMPLANTS

Infection

Is there an optimal amount of metal that can be implanted in vivo? Some researchers and clinicians have suggested that there may be a threshold, beyond which a patient may be predisposed to infection. Initially, infection rates with spinal instrumentation appeared higher than with noninstrumented procedures. This probably occurred because unfamiliarity with the newer systems led to longer operative times. Similarly, these procedures tended to be performed in patients who had failed previous, noninstrumented procedures. As has been discussed, more recent studies[71] have found no increased rate of infection over previous studies. Reynolds and colleagues,[51] in a prospective study over 3 years including 1000 patients, reported a deep tissue infection rate of 2.2%. All of the infections occurred from the posterior approach. Additionally, 21 of the 22 patients with infection had wound drainage, compared with only one patient in the control group. Richards[52] presented 10 cases of delayed deep infection in patients who had undergone posterior Texas Scottish Rite Hospital (TSRH) instrumentation for idiopathic scoliosis. These patients presented at an average of 25 months after surgery. Richards attributed this fact to intraoperative seeding followed by a period quiescence. Thalgott and associates[61] proposed a classification system to guide therapy and to identify patients at risk for development of infection. Their study also showed that cigarette smoking may be a significant risk factor.

Stress Shielding

Several investigators have examined the effect of load-bearing constructs on the underlying bone, specifically the possibility of stress shielding. However, there is no consensus from the research available. Bone loss is known to occur under plates in long bones.[44] Two reasons have been proposed: a regional necrosis secondary to disruption in blood supply, and a new equilibrium that is established between the bone and the load-sharing or load-bearing implant.[63] Lipscomb[40] demonstrated a 16% decrease in bone mineral content initially in patients who underwent a lumbosacral fusion. This was subsequently followed by an increase in BMC in 60% of patients to a level above that of the preoperative state. Using a canine model, McAfee and colleagues[41] showed an increased chance of fusion in instrumented versus noninstrumented spines. However, this was at the expense of bone density, which was significantly lower for the instrumented spines, a phenomenon they termed "device-related osteoporosis." Their results are in agreement with those of other researchers using various animal models.[23,58] Smith and coworkers,[59] using an animal model, similarly demonstrated a decrease in bone mineral content at 3 months. However, between 3 to 6 months, the BMC began to return to a baseline level. In addition, the amount of decrease in BMC was much less than expected by these researchers. Loosening at either the implant–construct or implant–bone interface may have accounted for this result, because there was a 37% decrease in screw torque from insertion to removal. There was also an unexpected decrease in BMC at the immediate adjacent caudad segment; the authors could not account for this.

Stress shielding is an accepted phenomenon in animal models. Whether it truly occurs in humans and

what effect it has on the fusion mass and adjacent bone is still to be determined.

Biology with Instrumentation

Spinal constructs are only as strong as the bone to which they are applied. Nowhere is this more important than in the elderly population, who have decreased BMD secondary to age or disease. Coe[13] and Yamagata[70] and their colleagues have demonstrated a linear relation between BMD and maximum pullout strength, using various spinal instrumentations. In the former study, laminar hooks were significantly more resistant to posterior pullout than were transpedicular screws and spinous process wires. There appeared to be no threshold level below which a significant decrease in pullout strength could be noted. However, other authors[2] have demonstrated a statistically significant difference in screw loosening secondary to cyclic loading below a critical BMD value of 0.45 g/cm^2. This difference in results may be accounted for by specimen variability or by a more physiologic mode of testing applied by the latter authors. Wittenberg and associates[68] evaluated the effect of BMC on screw pullout using human cadaveric vertebra. Among the five specimens with a mean density of 88 ± 11mg/cc^2, one screw loosened. This increased to more than one screw in 6 specimens with a mean density of 63 ± 12mg/cc^2, and no loosening with a mean density of 114 ± 14mg/cc^2.

QUALITIES OF IMPLANTS

A stable static system is a system in which the equations for static equilibrium ($\Sigma F_{xyz} = 0$ and $\Sigma M_{xyz} = 0$) are satisfied. The loading experienced by the spine is a complex system of dynamic and static loads and therefore cannot accurately be evaluated in this manner (by a free body diagram, for example). A dynamic system is one in which the application of forces or moments varies with time in magnitude, direction, or both. In the spine, these dynamic forces result from many factors, including muscle forces, body weight, internal reaction forces, and externally applied loads. Although the segmental architecture of the spine limits the motion generated by these dynamic loads, it is the incorporation of muscle forces, ligamentous restraints, and intervertebral discs that provides the spine with dynamic stability by counteracting body weight and external loading. A collection of vertebrae stacked on top of one another (as isolated bones) could not form a dynamically stable structure or even achieve a stable static equilibrium. Current surgical techniques do not give us the opportunity to properly restore the dynamic stability of the spine. The best option that exists to stabilize the spine is to bring the damaged segments to a state of reduced dynamics and increased static equilibrium. This is achieved by fusion of adjacent levels and by sacrifice of the motion at the damaged levels altogether. In order to execute a fusion, a solid construct of bone, bone graft, and implants is built to perform mechanically until the bone healing that results in solid bony fusion occurs.

When choosing or comparing spinal constructs, it is convenient to speak in terms of stiffness. Structural stiffness is the ability of a structural member to resist elastic deformation under an applied load. It is divided into three types: axial stiffness, torsional stiffness, and flexural stiffness. Axial stiffness is defined as the magnitude of the applied axial force divided by the linear

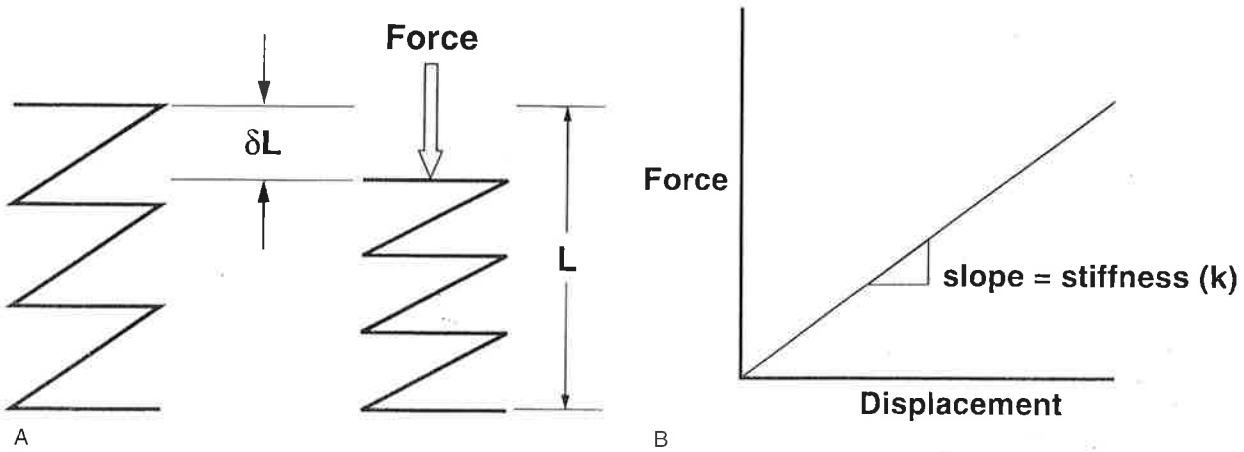

FIG. 18-1. (**A**) A linear spring element in compression, illustrating axial stiffness. Axial stiffness is equal to the applied axial force (F) divided by the resulting linear deformation (δL). (**B**) Graphical representation of axial stiffness (k), which is the slope of the linear, elastic region of the curve of force versus displacement.

deformation realized as a result of that force (Fig. 18-1*A*). Graphically, this translates to the magnitude of the slope of the elastic (linear) region of a curve of force versus deflection for an axially loaded element (see Fig. 18-1*B*). Axial stiffness is analogous to the modulus of elasticity (E) in this respect, because E is the slope of the elastic region of a curve of stress versus strain (Fig. 18-2). When conducting mechanical tests using spinal instrumentation, it is essential that we compare stiffness (i.e., load versus deflection) and not E, because E is an intrinsic material property that is constant for a given material rather than for a given construct.

In order to demonstrate this concept more clearly, consider the separate but identical mechanical testing of a carpenter's nail and an I-beam from a large office building, both constructed of the same steel. Plots of stress versus strain for axial compressive loading will be identical in slope (same E) for both members, yet the slope of the force-versus-deflection curve for the I-beam will be much larger than that of the carpenter's nail (indicating greater axial stiffness). Most spinal implants are constructed of either titanium or stainless steel. Factoring in the amount of material, in the form of the effective cross-sectional area, isolates the mechanical properties of the material itself rather than the entire implant or construct. The use of stiffness as a tool for quantitative comparison avoids the difficulties that may arise when attempting to determine the magnitude of the actual effective cross-sectional area needed to calculate the effective stress.

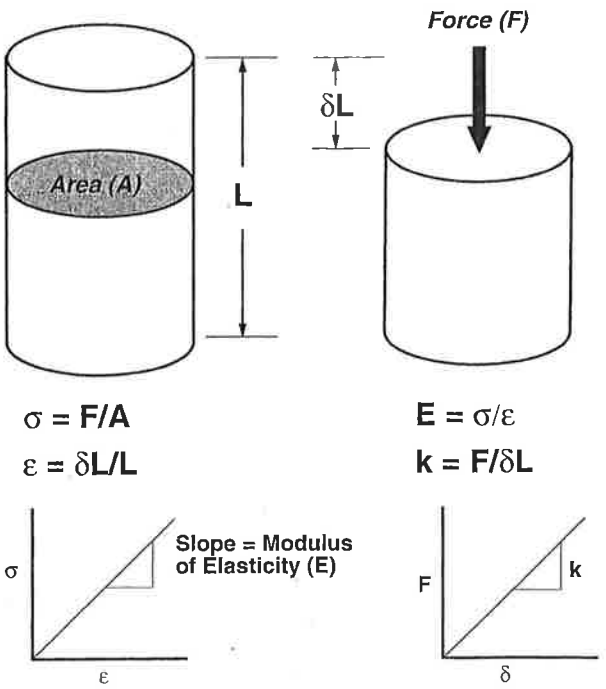

FIG. 18-2. Normal stress is equal to the applied force (F) divided by the contact area (A) to which the force is orthogonal. The slope of the linear portion of a curve of stress versus strain is the modulus of elasticity (E), which is an intrinsic material property. Plotting force versus displacement does not factor in the initial geometric properties of the specimen. The resulting slope of the linear portion of the graph defines stiffness (k).

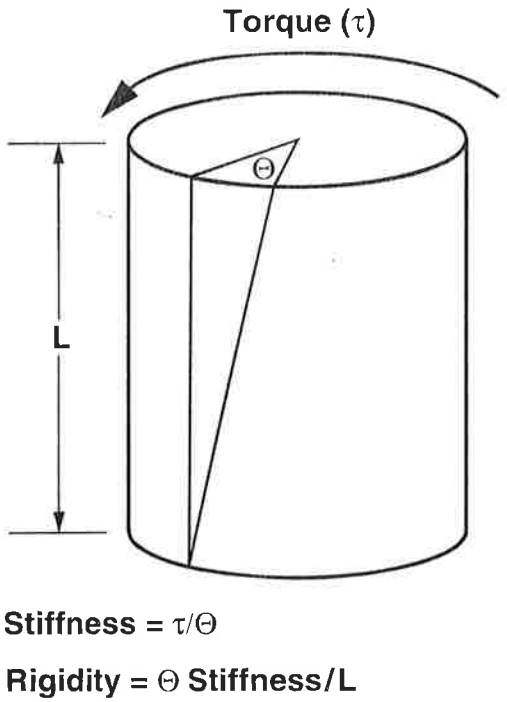

FIG. 18-3. Torsional stiffness is defined as the applied torque (τ) divided by the angular displacement (Θ) achieved as a result of that torque. Dividing the torsional stiffness by the length of the column defines torsional rigidity, which normalizes the angular stiffness as a function of the specimen's geometry.

Torsional stiffness is defined as the magnitude of the applied torque divided by the resulting angular deformation (Fig. 18-3). Because a goal of quantifying an implant's mechanical properties is comparison, it is important to remove as many testing variables as possible, thereby isolating the response of the implant. Dividing the calculated torsional stiffness by the length of the torqued shaft yields torsional rigidity, thereby normalizing the data for length.

Flexural stiffness can be defined two ways. Under the application of a pure bending moment, the stiffness is equal to the moment necessary to create a unit angular displacement at an end of a flexural member; therefore, dividing the applied moment by the resulting angular displacement will suffice (Fig. 18-4*A*). During three-point bending of a flexural member, the flexural

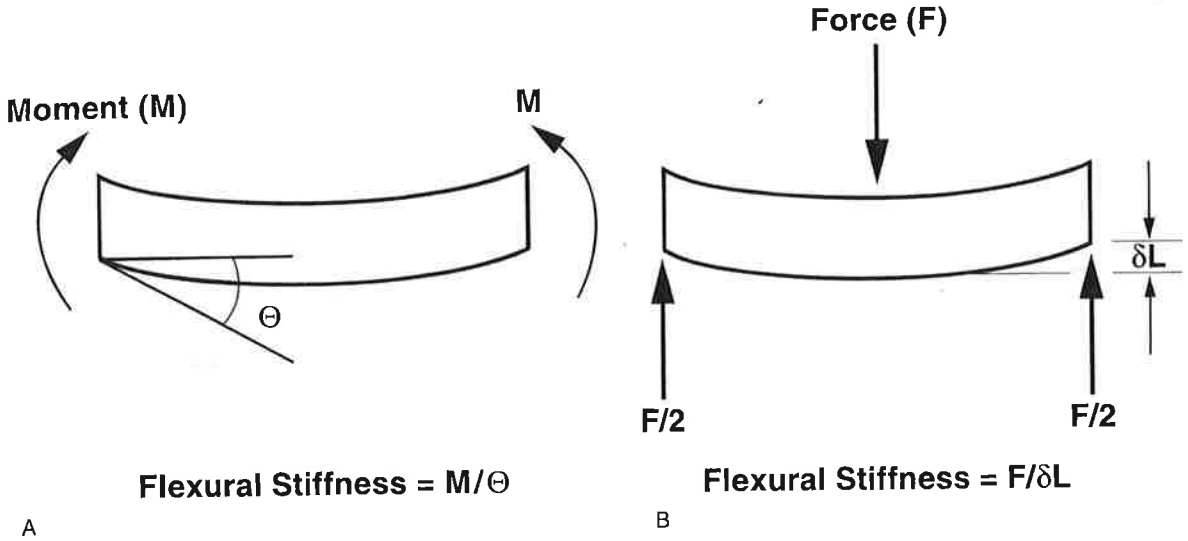

FIG. 18-4. Flexural stiffness can be defined in two ways: (**A**) The moment applied to a member divided by the resulting maximum angular deflection (Θ), or (**B**) The force (F) applied divided by the linear midspan deflection (δ).

stiffness can be calculated by dividing the force on the concave side of the member by the resulting linear displacement of the member at the point of loading (see Fig. 18-4*B*). This quantity can also be normalized by considering the geometry of the member, particularly the length of the segment.

Two terms often used incorrectly, in place of one another, are flexibility and ductility. Flexibility is closely related to stiffness (Fig. 18-5*A*) in that it is the achieved axial (or angular) deflection divided by the applied load (or torque). This quantity can readily be recognized as the inverse of the slope of the load-versus-deflection curve. Conversely, ductility is the inherent ability of a material to deform permanently. This is

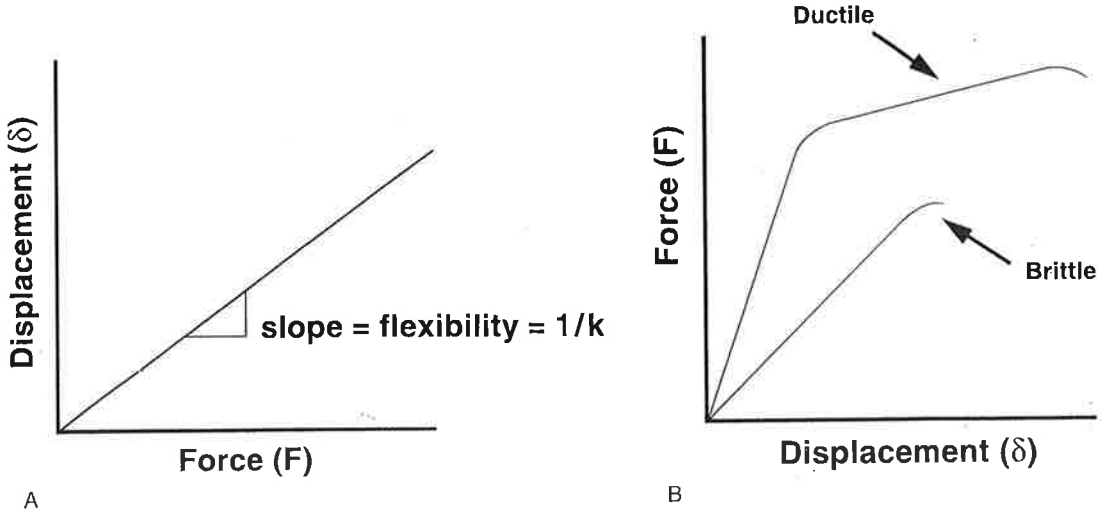

FIG. 18-5. (**A**) Flexibility is the inverse of stiffness; therefore, it is defined as the slope of the linear, elastic region of the curve of displacement (δ) versus force (F). (**B**) A ductile material is one that possesses the ability to permanently deform. This is graphically represented by a large plastic, postyield region on a curve of stress versus strain or force versus deformation. A brittle material is one that lacks the ability to permanently deform. A small or nonexistent postyield, plastic region on a curve of stress versus strain or force versus deformation graphically represents this material property.

characterized graphically by a large, plastic (nonlinear, postyield strength) range on a curve of stress versus strain or load versus deflection (see Fig. 18-5*B*). Therefore, ductility is an intrinsic material property; it depends not on the amount of material, but on the nature of the material. A nonductile material, one whose plot does not have a significant plastic region, is denoted a brittle material. Flexibility is dependent on the amount of material, just like its inverse, stiffness.

There are three different types of mechanical experiments that may be conducted: strength tests, fatigue tests, and stability tests.[48] A strength test is a destructive experiment that involves determining the ultimate load or moment a structure can sustain (i.e., the point of failure). A fatigue test, which is also a destructive experiment, allows the researcher to determine the number of cycles necessary to fatigue a structure to failure. A stability test is a nondestructive experiment that can be subdivided into two techniques: stiffness testing and flexibility testing. A stiffness test is a stability test in which the displacement (either axial or angular) is the independent experimental variable. The universal testing apparatus, such as a material testing system or Instron, a servohydraulic, or a screw machine, is used to control the displacement rate and direction, and the load necessary to accomplish this is recorded as the dependent experimental variable. In contrast, a flexibility test is a stability test in which the applied load is the independent variable and is governed by the programmed testing apparatus. The resulting displacement is sensed and recorded as the dependent experimental variable. Panjabi[48] suggested that the flexibility method is a natural method and thus superior to the stiffness method. This is because the flexibility method allows complete freedom of movement of all the vertebrae of the spine, thereby allowing the natural behavior of the spine to take place. In the stiffness method, the "free" end is not truly free because it must be attached to the crosshead or hydraulic actuator. This constraint of the specimen to follow the movements of the testing apparatus results in unphysiologic motion.

Some materials, either naturally or because of the mode of manufacturing, have physical, mechanical, and structural surface properties that differ from those of the core of the structure. The ability of a material to resist permanent deformation at its surface is a measure of that structure's hardness. This property is important when designing implants, because a difference in hardness along articular surfaces hastens wear.

The optimal combination of structural properties, for a spinal instrumentation system, is strength (both ultimate and fatigue), ductility, and moderate stiffness. This would allow for in situ bending in the plastic range of the stress-strain curve; meanwhile, service loads would remain in the elastic range, and adequate load sharing would be possible.[33]

Mechanics of Implants and Failure Analysis

There are six factors that determine the behavior of structures (implants) under load:

the mechanical property of a material
the amount of a material
the distribution of a material
the type of loading
the configuration of loading
orientation in relation to load.

A hypothesis is developed from the available evidence and the possible chains of events that could have caused the failure. If the failure can be duplicated under biomechanically controlled laboratory conditions, much can be learned about how the failure occurred. If this is not possible, there are factors about the failure that are not understood. The purpose of biomechanical testing, therefore, is to recreate failures under controlled in vitro conditions, and to relate these mechanisms to the actual in vivo situation.

Haher and colleagues[24] showed nondestructive cyclic load testing to be the most relevant way to examine a construct's susceptibility to failure. Several researchers[5,23] have employed this mode of testing to evaluate the stability of various spinal implants. However, these studies were limited to far fewer cycles than a patient could accomplish in vivo. The primary limiting factor in studies of this nature is the biologic deterioration of specimens over time. Many researchers have begun using ultra-high-molecular-weight polyethylene (UHMWPE) blocks to eliminate this variable. Additionally, in vitro testing is unable to account for the healing fusion mass and load relief to spinal implants in vivo. Zdeblick and coworkers,[72] using an animal model, demonstrated that stiffer constructs were more likely to produce a solid fusion mass.

The sequence of steps in the investigation of construct failure are as follows:

1. Collection of background history and selection of level
2. Preliminary physical examination and radiographic analysis
3. Surgical revision–excision of failed part and reconstruction
4. Macroscopic examination and analysis with photographic documentation
5. Microscopic examination and analysis (electron microscopy possible)
6. Nondestructive mechanical testing, including hardness and toughness
7. Selection and preparation of metalographic sections
8. Examination and analysis of metalographic sections

9. Determination of failure mechanism
10. Biochemical analysis (bulk, local, surface corrosion products, deposits or coatings, microprobe analysis)
11. Determination of fracture mechanics
12. Mechanical retesting under simulated failure conditions
13. Analysis of evidence, formulation of conclusions
14. Statistical analysis and report writing.

When performing failure analysis, three important principles must be adhered to:

1. Locate the origin of the failure,
2. Do not put fracture mating pieces back together; if parts must be repositioned to determine deformation of the total part, fracture surfaces must be protected.
3. Do no destructive testing without considerable thought; if performed at all, destructive testing must be performed only after all possible information has been extracted from the part in the original condition and after all original features have been carefully documented with photography.

If there are several failures from one mechanism, one should determine whether any of the failures was caused by fatigue fracture. If this is found, it is usually the source of the primary failure. Fatigue fracture is the expected type of failure of an element after long service; however, there are many possible reasons for fatigue fractures. Fatigue failures are common at the lumbosacral junction, unless specific actions have been taken to prevent them during design, insertion, bone grafting, and service.[69]

In general, failure occurs on two levels: local and global. Localized compression failure occurs continuously in trabecular bone. In areas of high stress, where increased strength is needed, these localized failures are accompanied over time by rebuilding and remodeling (according to Wolff's law of trabecular architecture). Global failure can be defined as the scenario in which the rate and magnitude of local trabecular failures exceeds that of rebuilding and remodeling, so that the overall ultimate compressive strength of the body is compromised below the level necessary to withstand service loading.

Materials

Metals are the chief implantable materials used in spinal surgery. They provide strength, can be contoured to match bony contours, and are biocompatible. The most commonly used metals are stainless steel and titanium alloys. The production of alloys allows for the use of strategic ratios of different elements in order to engineer the most stable and economical instrument by harnessing the optimal benefits from each material within the composite construct.

Stainless steels are alloys comprised of many elements, most notably iron (>58%), chromium (17% to 20%), and nickel (13% to 16%). Stainless steel alloys are favored because of their low cost and the wide range of structural and mechanic properties that can be achieved by varying the ratios of the constituents. Chromium offers the alloy the ability to resist corrosion.[7]

Pure titanium is a good material for orthopaedic uses because of its low density, high ductility, and compatibility with CT and MRI scanning. Its chief drawback is its low E. As a result, titanium is commercially combined with aluminum and vanadium to produce the alloy Ti-6Al-4V, which has a greater tensile and greater fatigue strength than stainless steel. This alloy, which is commonly used in orthopaedic instrumentation, is composed chiefly of pure titanium (approximately 90%), aluminum (5.5% to 6.5%), and vanadium (3.4% to 4.5%), but trace amounts of iron (<0.25%) and oxygen (<0.2%) are also present. Although the alloy's E is greater than that of pure titanium, it is still not as great as that of stainless steel, and therefore larger hardware must be used to provide comparable rigidity. The structural properties of titanium alloys are usually about half those of stainless steel alloys.[7]

Polymethylmethacrylate (PMMA) is sometimes used as an adjunct to internal fixation. PMMA is stiffer than cancellous bone but less stiff than cortical bone. It is used primarily in cases in which bone quality can compromise fixation, leading to failure of fixation. Several authors have demonstrated the efficacy of PMMA in enhancing screw fixation.[9,21] Zindrick and colleagues[74,75] showed a 96% increase in mean pullout force with the use of pressurized cement but no significant difference with nonpressurized cement. They recommended 4 cc of cement if the screw hole has not been compromised and 6 cc in cases of screw loosening or stripping.[75] Spivak and coworkers[60] studied the effect of a hydroxyapatite (HA) coating on transpedicular screw pullout strength using a dog model. They found the HA-coated screws to be less resistant to pullout than standard screws in both recommended and poor initial fit holes. They cited two possible reasons: the HA coating dulled the self-tapping portion and outer margins of the screw, and the HA coating decreased the available space for bony fill between the threads. Other researchers, using long bones, have shown that pullout strength increases with time after insertion in vivo, as a result of bone ingrowth into the screw threads.[32,62]

Corrosion

Corrosion in implants is a serious and costly matter. Mechanically, corrosion is not catastrophic unless it results in failure (i.e., fracture). If two heterogeneous

metals are in contact with each other in an electrolyte solution, an electric current is produced. Positively charged ions leave the anode and are attracted to the cathode. The corrosion is greatest when the cathode is large and the anode is small. This is why we do not mix dissimilar metals. An oxide layer acts to protect an implant by blocking the passage of ions, thereby preventing current flow. Corrosion of implants in vivo can occur by one or a combination of several processes: crevice corrosion, which is secondary to oxygen depletion in a confined space, leading to a decrease in local pH; fretting corrosion, which is the result of abrasion of metallic surfaces that removes the oxide layer; and stress corrosion, which is secondary to environmental conditions.

Very few studies have examined corrosion in spinal instrumentation. Bidez and colleagues[6] demonstrated corrosion wear phenomena at the Luque rod–wire interface in both retrieved specimens and those tested in vitro. The authors point out that this study was limited to low cycling conditions and that corrosion wear should be of concern with more physiologic loading. These findings were corroborated by Nasca and associates,[47] who used nondestructive cyclic testing of swine spines.

Rods Versus Plates

The stiffness of rods and plates depends primarily on the type of material (stainless steel is stiffer than titanium) and the thickness of the rod or plate [½ bh^3 (Area Moment of Inertia for a Plate) vs. ¼.πr^4(Area Moment of Inertia for a Rod)]. Cunningham and colleagues,[15] in biomechanical tests using UHMWPE blocks, demonstrated that implants with rod constructs had higher stiffness values than plate-based systems. In the same study, fatigue testing at 600 N led to failures at between 600,000 and 800,000 cycles for the rod-based pedicle screw systems. All the systems failed at the superior screw collar junction, except the Compact Cotrel-Dubousset (CCD) system, in which the rod fractured. This was probably due to the decreased size of the CCD rod.

Carson and associates,[10] using an in vitro and finite element model, described the "4R-4Bar" concept, in which, below approximately 30° of pedicle-to-pedicle angle, the flexibility of a construct increases, leading to inability to resist an applied load. Eventually, load sharing by the spine or construct failure ensues. The authors were able to eliminate this "bilevel mechanism" by crosslinking the longitudinal members of the implant. Dick and colleagues[18] compared zero, one, or two crosslinks in various modes of loading to determine their effect on construct stability. They found no significant increase in axial, flexion, extension, or lateral stiffness. However, torsional stiffness was increased 44% with one crosslink and an additional 26% with two crosslinks.

With older instrumentation systems, the surgeon was obligated to go several levels above and several below the level of instability to achieve fixation. More recently, pedicle screw systems have allowed for an approach using one level above and one below, with the preservation of motion segments. This is especially important in the lower lumbar spine. All systems must be load-sharing and not load-bearing; otherwise, hardware failure is more likely to ensue. McCormack and associates[42] retrospectively reviewed 28 patients with three-column failure who underwent short-segment instrumentation posteriorly. Using a load-sharing classification scheme that primarily evaluates the anterior column, they found screw breakages in 10 of 28 patients with scores greater than 7 points, indicating involvement of the most anterior column. These authors now recommend some type of anterior support in three-column failure.

CONCLUSION

Restoration of spinal balance depends on appropriate selection and implementation of spinal implants, with fusion as the final common pathway to achieve this goal. The future of spinal surgery may lead to the development of new implants using modern metallic alloys and biologic substrates to augment instrumentation in the form of bone morphogenic protein and bone matrix analogs. New, minimally invasive techniques for fixation will lead to alternatives to traditional methods of spinal fusion. Fusion itself, as a conceptually inferior solution, may disappear altogether, leaving the stage to arthroplastic types of reconstruction.

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