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Rockwood and Green's

Fractures in Adults

EIGHTH EDITION

EDITORS
Charles M. Court-Brown
James D. Heckman
Margaret M. McQueen
William M. Ricci
Paul Tornetta III

Vol. 1

ASSOCIATE EDITOR
Michael D. McKee

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Volume 1

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We dedicate this Eighth Edition of Rockwood and Green's: Fractures in Adults to Charles A. Rockwood, Jr, MD, and David P. Green, MD, who served as our inspiration and mentors for carrying on the revision and update of this textbook.

To Susan for her patience and understanding during my 30-year tenure on the editorial board.

JDH

To the future: Emily, Jessica, and Rosie.

CCB

To my children Sacha, Tyler, Robbin, and Everett for enriching my life every day, and my partner Niloofar for her love and support.

MMcK

To Caroline, Elizabeth, and William without whom life would be easier but much less fun.

MMcQ

To Ann, Michael, and Luke, my reasons for being, for their patience, love, and support.

WMR

To my mother Phyllis, who found the best in people, had compassion for all, and whose insight, guidance, and love have always made me believe that anything is possible.

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Preface

The eighth edition of *Rockwood and Green's: Fractures in Adults* continues with the changes that were instituted in the seventh edition. In this edition there are two more chapters and 61 new authors drawn from three continents and 11 different countries. In addition, many of the new authors represent the next generation of orthopedic trauma surgeons who will be determining the direction of trauma management over the next two or three decades.

Orthopedic trauma continues to be an expanding discipline, with change occurring more quickly than is often realized. When Drs. Rockwood and Green published the first edition in 1975, there were virtually no orthopedic trauma specialists in most countries, fractures were usually treated nonoperatively, and mortality following severe trauma was considerable. In one generation the changes in orthopedic surgery, as in the rest of medicine, have been formidable. We have worked to incorporate these changes in this edition. There is expanded coverage in this edition of the inevitable complications that all orthopedic surgeons have to deal with, and we have included chapters on geriatric trauma and the psychological aspects of trauma. The other area of orthopedic trauma that is expanding quickly, particularly in the developed countries, is the treatment of osteoporotic (or fragility) fractures. These fractures are assuming a greater medical and political importance, and orthopedic implants are now being designed specifically to treat elderly patients. It is likely that this trend will continue over the next few decades; many of the chapters in this edition reflect this change in emphasis.

The changes in the eighth edition include major changes in its chapter structure. Each of the clinical chapters now follows a specific template beginning with the physical examination, classification, and additional studies used in the diagnosis of each problem. This is followed by a description of the outcome measures used to evaluate patients for the specific injury they sustained. The indications and contraindications for each treatment method, including nonoperative and operative methods are highlighted in tables, as are the technical aspects of the surgeries. Old favorites such as pitfalls and problems are also listed in tables with solutions. Finally, the author's preferred treatment is now presented in the form of an algorithm, allowing the reader to understand the thought process of the expert writer in deciding on the treatment for the multiple subtypes of injuries described in each chapter. We believe that this will make it easy to get the most out of each chapter.

Finally, we are proud to introduce a new electronic format that should allow for easier access across platforms, a change that is overdue! Video supplementation is also available for the majority of the clinical problems.

We are indebted to the efforts of the experts who have taken the time to share their knowledge and experience with our broad readership and hope that this new edition will contribute to the care of patients.

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SECTION ONE

General Principles

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BIOMECHANICS OF FRACTURES AND FRACTURE FIXATION

Mark J. Jo, Allan F. Tencer, and Michael J. Gardner

INTRODUCTION TO BIOMECHANICS OF FRACTURES AND FRACTURE FIXATION

"Biomechanics" is a complex and encompassing term that applies to many aspects related to orthopedic surgery, and specifically to fractures and fracture fixation. The application of biomechanical principles and concepts is essential to understand how the fracture occurred, how to best treat the injury, and how to avoid mechanical failures of the fixation construct. One must first understand the fundamental terms and concepts related to mechanical physics. This establishes the foundation that will be used to apply these concepts to the field of orthopedic surgery. The biomechanical properties of bone as well as the biomechanics of fracture healing are also essential to understand how bone is injured and how to best restore its function. Finally, understanding the biomechanical properties of common implants and failures seen with their application helps the clinician to a thorough understanding that aids in patient care.

In the study of biomechanics as it relates to fracture fixation, the fundamental mechanical question remains: Is the fixation system stable and strong enough to allow the patient early mobility before bony union is complete? This must occur without delaying healing, creating bone deformity, or damaging the implant, and yet be flexible enough to allow transmission of force to the healing fracture to stimulate union. The common adage in orthopedics is that, "Fracture healing is a race between bony union and implant failure." A thorough understanding of the biomechanical concepts as they relate to bone, fracture, and implants is essential for the proper treatment of patients with fractures.

BASIC CONCEPTS

Before describing the performance of fracture fixation systems, some basic concepts used in biomechanics must be understood. A *force* causes an object to either accelerate or decelerate. It has *magnitude* (strength) and acts in a specific direction, which is termed a *vector*. However complex the system of forces acting on a bone, each force may be separated into its vector components (which form a 90-degree triangle with the force). Any of several components, acting in the same or different directions, can be added to yield the net or *resultant force*. As seen in Figure 1-1, a simplified example of the hip joint shows that the forces acting about the hip include the body weight, joint reactive force, and the hip abductors. As the hip in this example is at rest, the net force must be zero; therefore, if the body weight and hip abductor forces are known, the joint reactive

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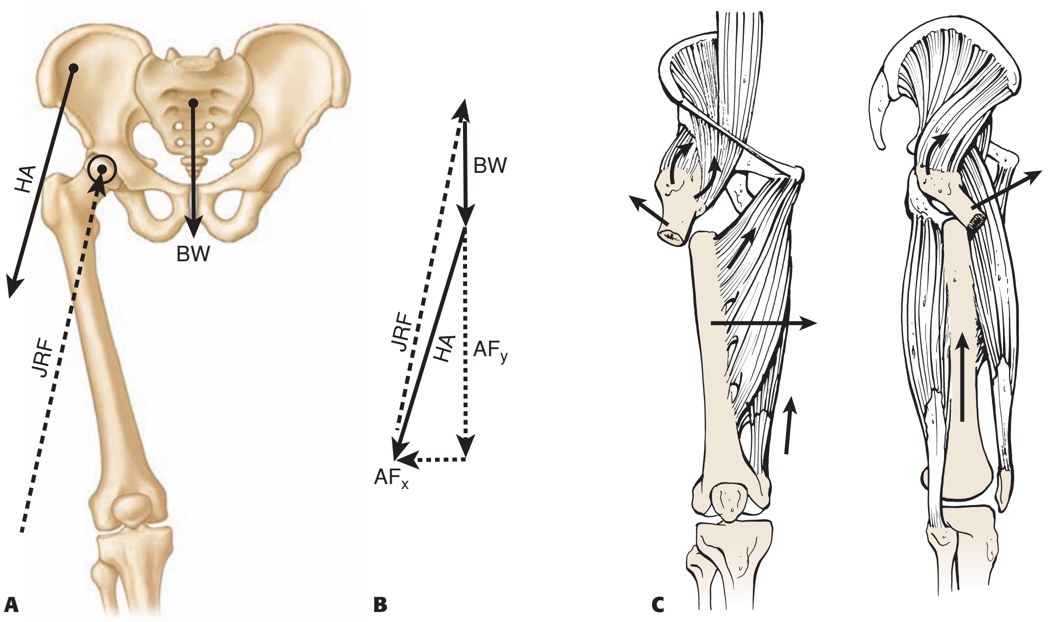

FIGURE 1-1 The force vectors acting on different parts of the body are a culmination of muscle, tendons, ligaments, and external forces. **A:** A simplified example of the force vectors acting on the hip joint. HA, hip abductors; BW, body weight; JRF, joint reactive force. **B:** Using the *x* and *y* vector components of the forces about the hip the joint reactive force (JRF) can be calculated because if the hip is at rest, the sum of all the forces should equal zero. AF_y (vertical component of HA force) AF_x (horizontal component of HA force). **C:** Understanding the forces that are applied about a fracture can help the surgeon understand the deforming forces and assist in reduction and fixation strategies.

force can be calculated using the *x* and *y* components of all the forces. Also, understanding the forces about a fracture help the surgeon to understand the deforming forces, the reduction maneuvers, as well as the proper application of implants to best stabilize the injury. Both the design of the implants as well as the application by the surgeon must be done with these concepts in mind so that they can withstand the mechanical loads applied without failure.

The two major loads acting on a long bone are those that cause it to displace in a linear direction (translation) and those that cause it to rotate around a joint center. Muscles typically cause a bone to rotate (e.g., the biceps causes the forearm to flex and supinate, the anterior tibialis causes the foot to dorsiflex). When a force causes rotation, it is termed a *moment* and has a *moment arm*. The moment arm is the lever arm against which the force acts to cause rotation. It is the perpendicular distance of the muscle force from the center of rotation of the joint. As shown in Figure 1-2, the moment or rotary force is affected not only by the magnitude of the force applied, but also by its distance from the center of rotation. In the example, two moments act on the outstretched arm. The weight carried in the hand as well as the weight of the hand and forearm rotate the arm downward, while the balancing muscle force rotates the forearm upward. Equilibrium is reached by balancing the moments so that the forearm does not rotate and the weight can be carried. Note that to achieve this, the muscle force must be eight times as large as the weight of the object, forearm, and hand because its moment arm or distance from the center of the joint is only one-eighth as long.

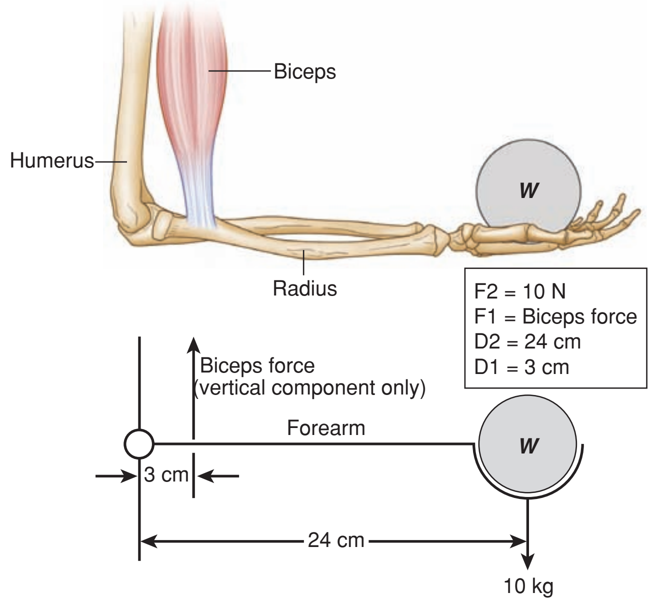

FIGURE 1-2 In this simplified example of a free body diagram, the outstretched arm is a lever and is at rest. The rotational force, or the moment, is centered about the elbow. This moment is defined as the product of the weight (object + forearm + hand) (F_2) and the distance from the elbow (d_2). This moment must be counteracted by a moment in the opposite direction. In this example the vertical component of the biceps force (F_1) is the counteractive force. The lever arm of this force is the distance from the elbow to the insertion of the biceps (d_1). The biceps force is calculated from 10 kg × 24 cm = F_1 × 3 cm. Thus F_1 = 80 N. The biceps force is much greater than the weight of the object, arm, and hand because its lever arm is smaller.

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The basic forces—compression, tension, torsion, and bending—cause the bone to behave in predictable ways. A *compressive force* (Fig. 1-3) results in shortening the length of the bone, whereas *tension* elongates it. *Torsion* causes twisting of a bone about its long axis, whereas *bending* causes it to bow at the center. When these forces are great enough to cause the bone to fracture, it results in characteristic fracture patterns that can be recognized on radiographs. Understanding these forces

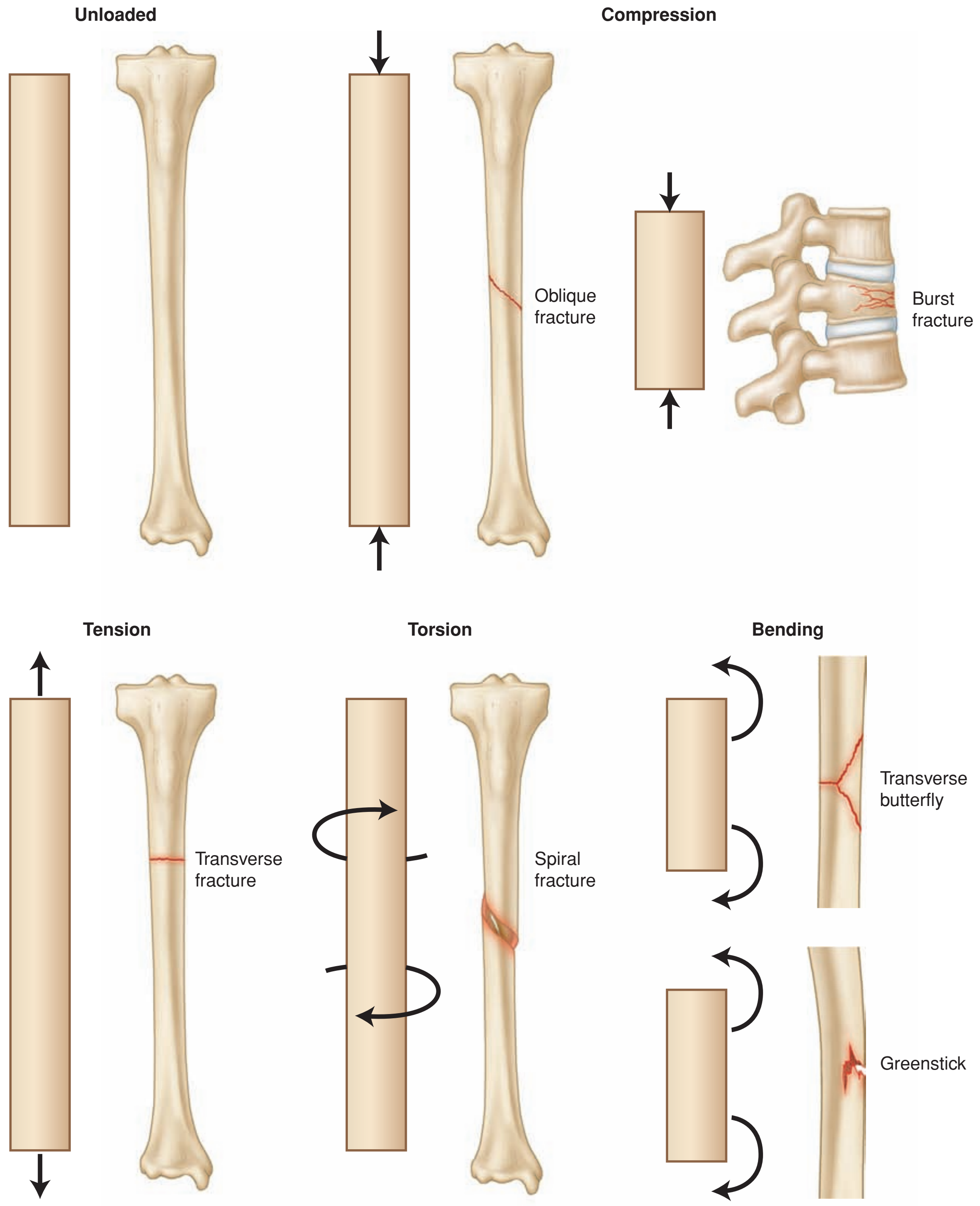

FIGURE 1-3 Basic forces: Unloaded; compression shortens length and can lead to an oblique fracture line or comminution; tension can lead to a transverse fracture. Torsional forces usually cause a spiral pattern. Bending forces cause compressive forces on one side and tensile forces on the other. This can result in a transverse fracture on the tensile side and comminution in a classic butterfly pattern on the compressive side. Bending forces can also result in incomplete or "greenstick" fractures in the pediatric population.

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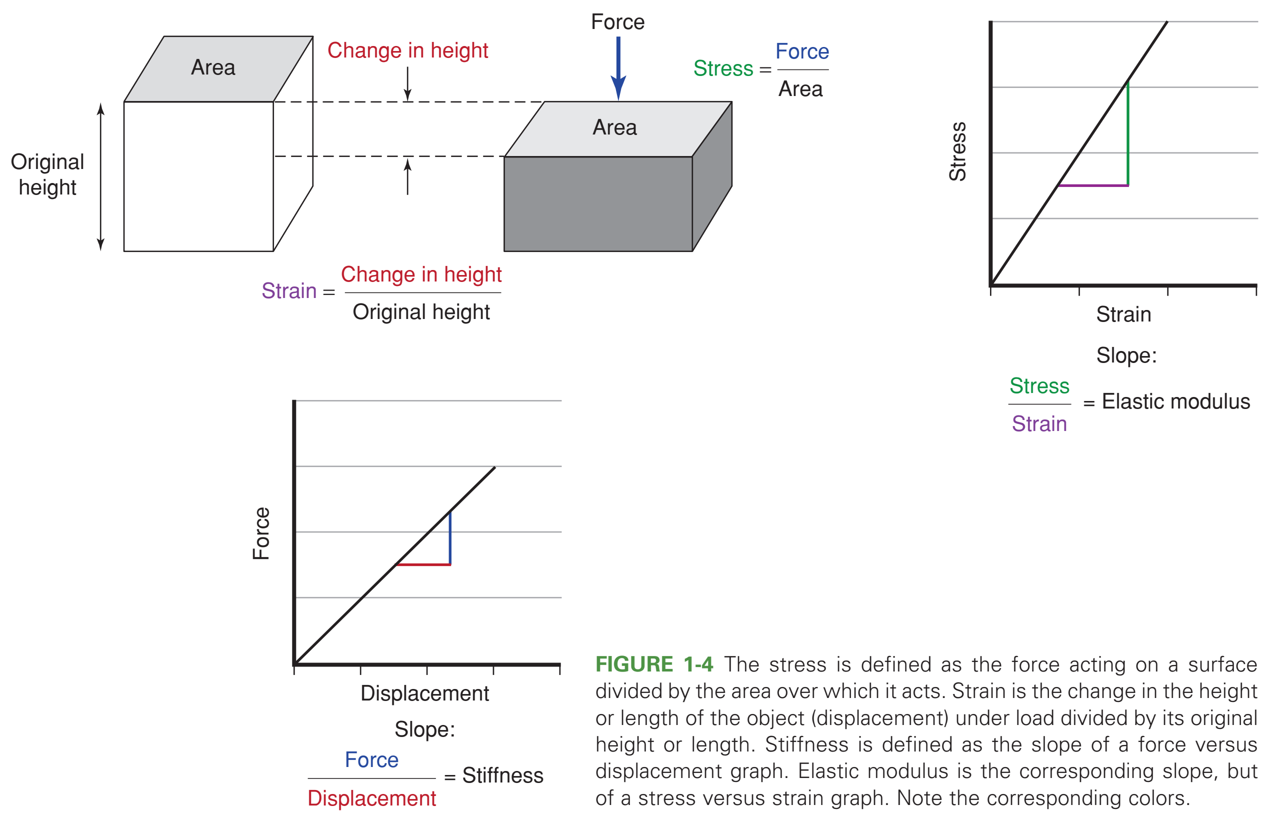

FIGURE 1-4 The stress is defined as the force acting on a surface divided by the area over which it acts. Strain is the change in the height or length of the object (displacement) under load divided by its original height or length. Stiffness is defined as the slope of a force versus displacement graph. Elastic modulus is the corresponding slope, but of a stress versus strain graph. Note the corresponding colors.

can help to understand the circumstances of the forces that occurred at the time of the fracture. Compressive forces can cause oblique fracture lines or can result in comminution and fragmentation of the bone. *Tensile* forces usually cause transverse fracture lines, whereas torsion can cause spiral fractures. Bending forces cause compressive stress on one side and tensile stress on the other side. Bending forces can also cause plastic deformation of immature or flexible bone or result in partial fractures. These partial fractures are also known as "greenstick" fractures and are usually seen in the pediatric population. In a more rigid bone, the tensile forces result in a transverse fracture line and the compressive forces cause comminution, usually in the characteristic butterfly fragment. In many cases an injury is caused by a combination of these forces and the fracture pattern may have a combination of patterns.

Stress, as shown in Figure 1-4, is simply the force divided by the area on an object over which it acts. This is a convenient way to express how the force affects a material locally. For example, when an equal force (hammer blow) is applied to both a sharp and a dull osteotome, the sharp osteotome will concentrate the same force over a smaller surface area than a dull osteotome because of the sharp edge. Therefore, the sharp osteotome will create a greater stress at the osteotome–bone interface, resulting in cutting of the bone. Just as stress is a normalized force (force per unit area), changes in length can also be normalized. *Strain* is simply the change in height or length that a material undergoes during loading, divided by its original height or length. If two plates of different lengths are both subjected to loads that lengthen the plate by 1 cm, the shorter of the two plates will be subjected to more strain as change in length is spread over a shorter distance than it is for the longer plate.

Mechanical testing is used extensively to analyze the properties of different constructs as well as new implant designs.[67] The testing usually consists of a natural or synthetic fractured bone fixed with a certain implant in different configurations. This construct is then loaded into an apparatus that applies a specific load in either a constant or cyclic manner. Sensors can measure the forces applied to the bone as well as any deformity or eventual failure (Fig. 1-5). Depending on the purpose of the experiment the data can be collected measuring the *structural properties* of the bone–fixation construct; that is, the properties of the fixation device and the bone combined. Alternatively, the data can measure the *material properties* which relate to the properties of the substances that make up each component (bone, stainless steel, titanium). In this example, the material properties of the plate are being tested using a fracture model. The corresponding graph represents the data measured in this experiment plotted on a stress–strain graph. The force and displacement are measured and normalized to stress and strain. The initial deformation is termed *elastic* because when the load is removed, the plate will return to its original shape. This is represented by the linear portion of the graph, termed the elastic region. At some load, however, the construct becomes overloaded, entering the *plastic* range. If the load is released after loading in the plastic range but before failure, some *permanent deformation* remains in the construct. The point at which elastic behavior changes to plastic is termed the *yield point.* As previously mentioned, the slope of the stress–strain curve is the elastic (Young's) modulus.

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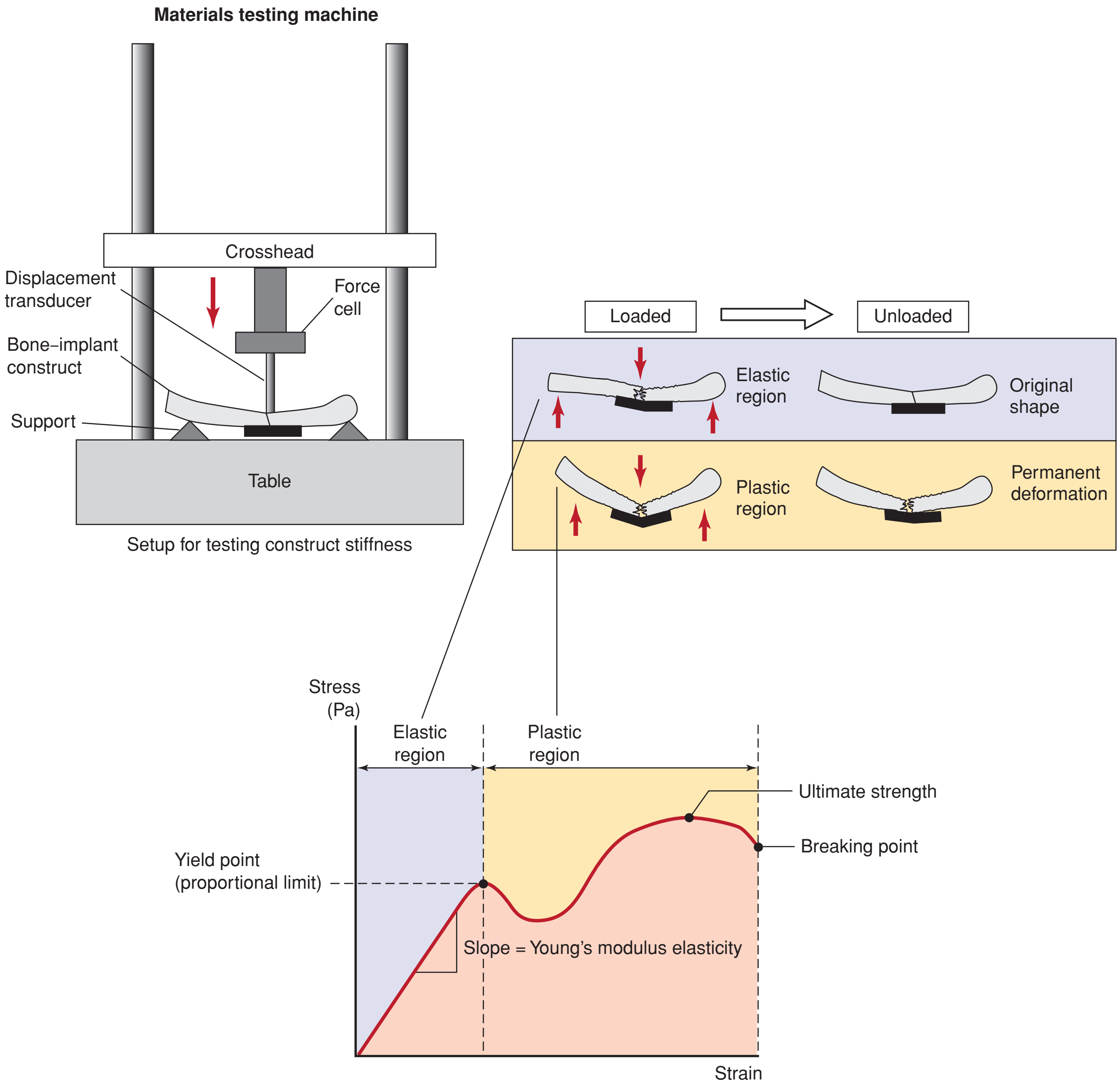

FIGURE 1-5 **Top left:** A fixation construct setup in a mechanical testing machine. In this example, a long bone is fixed with a plate and subjected to bending. **Top right:** The construct during loading in the elastic region and plastic region. **Bottom:** The resulting measurements from the testing machine, which measures stress and strain at the point of the applied load. The graph demonstrates the elastic region, in which the plate acts like a spring, returning to its original shape after the load is released; the plastic region, in which the plate may have permanent deformity; and the failure load, in which the plate fails. The area beneath the curve *(pink area)* is the toughness of the material, or the amount of energy that a material can absorb before failure.

The area under the stress–strain curve is termed the strain energy which is the energy absorbed. *Toughness* is the amount of energy that a material can absorb before failure.

The elastic range represents the working range for the fixation construct. In this region the plate is able to withstand the forces applied to it without losing its shape. The yield point defines the safe maximum functional load before the plate is permanently deformed. A third very important property, fatigue, will be discussed later.

Note that a fixation construct may have different yield points and stiffnesses for loads acting in different directions. An example is a half-pin external fixator construct applied to a tibia, with the pins oriented anteriorly–posteriorly. The stiffness is much greater in anterior–posterior (flexion/extension) bending than medial–lateral (varus/valgus) bending for this construct. Another property to consider is the *work done* in deforming a fixation construct. The product of the force applied and the distance the construct bends is defined as the work done, and is represented by the area under

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TABLE 1-1 Basic Engineering Properties of Common Biologic and Implant Materials

Material	Ultimate Strength Tensile (MPa)	Ultimate Strength Compressive (MPa)	Yield Strength 0.2% Offset (MPa)	Elastic Modulus (MPa)
Muscle	0.2			
Skin	8			50
Cartilage	4	10		20
Fascia	10			
Tendon	70			400
Cortical bone	100	175	80	15,000
Cancellous bone	2	3		1,000
Plaster of Paris	70	75	20	
Polyethylene	40	20	20	1,000
PTFE Teflon	25			500
Acrylic bone cement	40	80		2,000
Titanium (pure, cold worked)	500		400	100,000
Titanium (Al-4V) (alloy F 136)	900		800	100,000
Stainless steel (316 L) (annealed)	>500		>200	200,000
Stainless steel (cold worked)	>850		>700	200,000
Cobalt chrome (cast)	>450		>50	20,000
Cobalt chrome (wrought, annealed)	>300		>300	230,000
Cobalt chrome (wrought, cold work)	1,500		1,000	230,000
Super alloys (CoNiMo)	1,800		1,600	230,000

(Ultimate tensile strength or maximum force in tension, yield strength at 0.2% offset, the strength at which the strain in the material [change in length/original length] is 0.2%, a usual standard for metals, elastic modulus, or stress/strain.)

the force–displacement graph of Figure 1-4. A material may be flexible and tough (e.g., rubber, or a child's bone that deforms but is difficult to break) or stiff but brittle (e.g., glass, elderly bone), if it cannot absorb much deformation without fracturing.

The factors that govern stiffness and yield point are the material from which the fixation device is made and its shape. A construct made of higher elastic modulus materials will be stiffer (e.g., stainless steel is stiffer than titanium) (Table 1-1). The stiffness of a construct is found by dividing the force applied by the deformation that the construct exhibited. The *elastic* (or *Young's*) modulus is determined by dividing the stress applied by the resulting strain (Figs. 1-4 and 1-5). The moduli of some common orthopedic materials are given in Table 1-1. As shown, the elastic modulus of titanium alloy is about one-half that of stainless steel; so, given two plates of the same size and shape, the titanium plate has about one-half the stiffness of the stainless steel plate. This can be important to consider when using new devices made of different materials.

Another concept is how the shape and size of an implant influences the load it can support. As shown in Figure 1-6, a typical plate used in fracture fixation is wider than it is thick. Thus, the plate is actually stiffer when the load is placed against the edge rather than the broad surface of the plate. This is because when the load is applied on the edge of the plate, the material of the plate resisting the load is distributed further away from the center (note that in this example, the material of the plate did not change, just its orientation relative to the load applied). This concept of distribution of material is reflected in the shape property, *moment of inertia*. The moment of inertia provides a measure of how the material is distributed in the cross section of the object relative to the load applied to it. The farther away the material is from the center of the beam, the greater its stiffness. Steel I-beams were developed to take advantage of this concept; that is, gaining greater stiffness for the same amount of material. For solid cylindrical objects like rods, pins, or screws, their stiffness is related to the fourth power of their radius. As shown in Figure 1-6, for rods made of the same materials, a 16-mm diameter intramedullary (IM) rod is 1.7 times as stiff as a 14-mm rod ($[8/7]^4 = 1.7$]).

A third important property of a fracture fixation construct is its ability to resist *fatigue* under cyclic loading. Load can be applied that remains below the yield point of the construct, yet creates a crack that progressively grows. This lowers the yield point of the material and the local stresses will eventually exceed the yield point and the construct will fail (Fig. 1-7). Some materials have an *endurance limit* such that they can support a certain level of load indefinitely without failure. An important aspect of fatigue performance of a fixation construct is the effect of a *stress riser*. In completely uniform materials, the stresses will be almost identical throughout the material. But typical fixation devices have holes, screw threads, and other features in which the shape changes and leads to a change of

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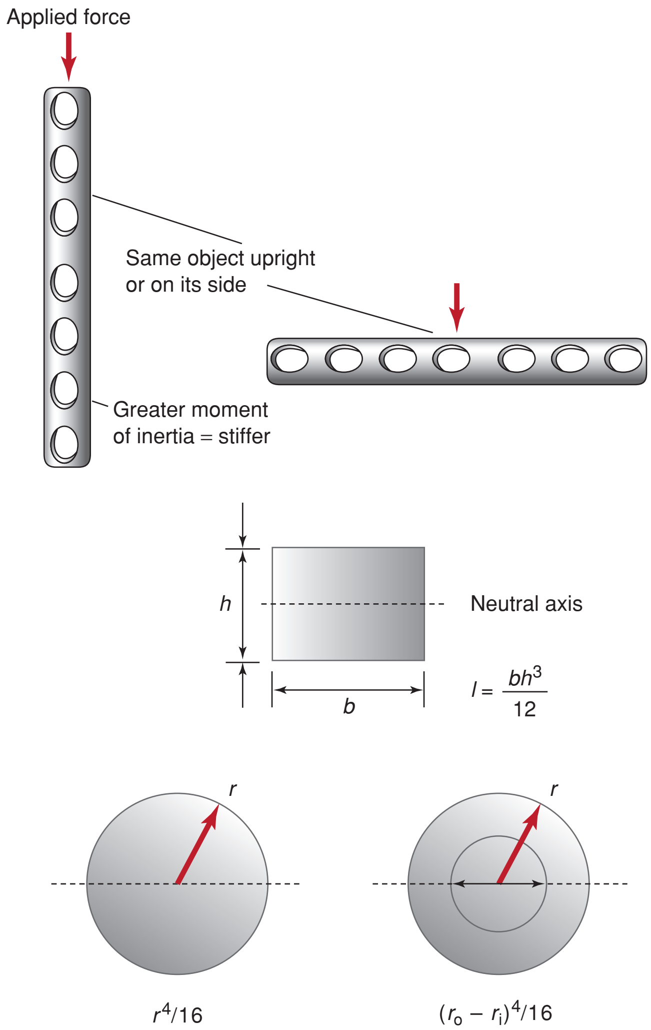

FIGURE 1-6 The concept of moment of inertia or the effect of the geometry of an object on its stiffness. **Top:** Looking at a typical plate used in fracture fixation, when the load is applied on the broader surface the plate is less stiff than when the load is applied to the narrower edge. This is because the distribution of the material is farther from the load applied. **Bottom:** The moment of inertia is a term used to describe how the material is distributed within an object. For a solid rectangular object such as a plate, the moment of inertia (I) and the stiffness increase directly with the width (b) of the plate and the cube of its height (h). For a solid cylinder, such as a pin or a screw, the moment of inertia increases with the fourth power of its radius (r). Therefore a 16-mm diameter IM rod is 1.7 times as stiff as a 14-mm rod, and 2.3 times as stiff as a 13-mm rod, if all the rods are made of the same material. For a hollow cylinder such as an intramedullary nail, the radius of the inner diameter (r_i) is subtracted from the radius of the outer diameter (r_o). The moment of inertia still increases by the fourth power.

the material properties. It is the transition points which create a stress riser. One must also take into account the interface at the end of a fixation construct. The end of the plate or rod creates an abrupt transition between the metal and bone creating a stress riser. Although this cannot always be avoided, placing the end of the implant in a high-stress area such as the subtrochanteric region of the femur can lead to periprosthetic fractures (Fig. 1-8). These fractures can be secondary to another

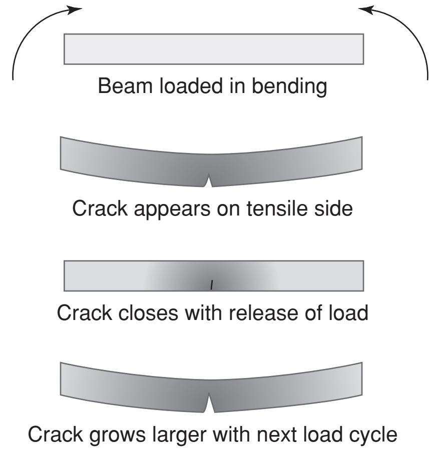

FIGURE 1-7 A stress concentrator is a region of an object in which stresses are higher than in the surrounding material. Taking the example of a fracture plate subjected to bending, the bottom surface elongates under load. In the region of highest tensile forces, a scratch starts to grow into a crack that closes when the load is released, then reopens slightly larger with the next load cycle, eventually growing to a point at which the plate fails. Crack growth is accentuated by stress corrosion, poor bone-to-bone contact at the fracture, and by loads applied by heavier patients.

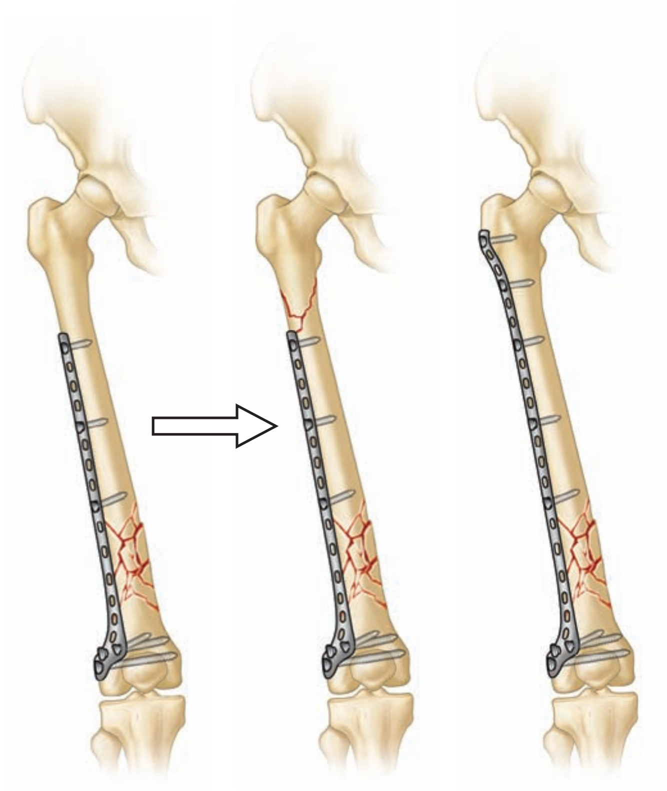

FIGURE 1-8 A stress riser at the end of a fracture construct can cause problems if it is in a region of high stress. In this example, a femoral shaft fracture is fixed using a lateral plate. If the end of the plate is in the high-stress subtrochanteric region, there is a risk that the stress riser can contribute to a periprosthetic fracture. To avoid this, a longer plate can be used to bypass the high-stress area.

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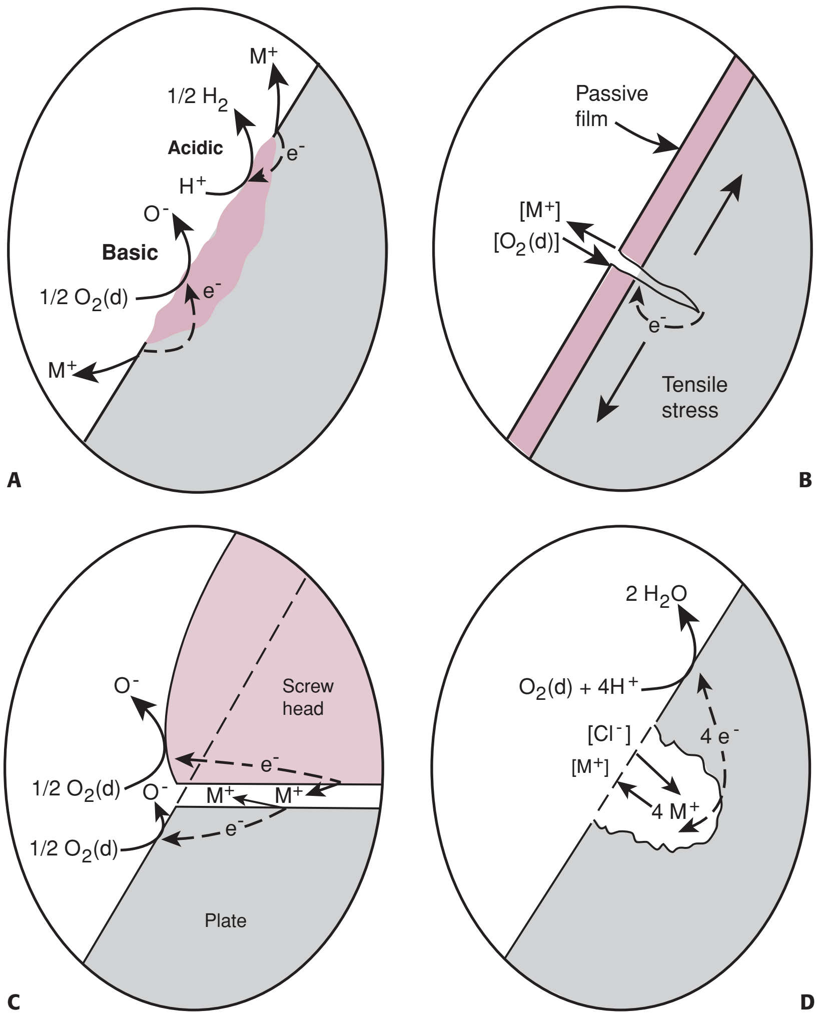

FIGURE 1-9 **A:** Illustration of crevice corrosion, with a local galvanic cell caused by an impurity in the surface of a plate and ions, M^+, being released, resulting in loss of material and formation of a crevice. **B:** Stress corrosion occurs by a local galvanic cell setup between the material at the tip of the crack, which just opened and has not oxidized, and the remaining oxidized surface of the plate. The released ions enhance crack growth occurring from loading. **C:** Fretting corrosion caused by the loss of the oxide layer on the surface of a plate caused by rubbing of the base of the screw against the plate. **D:** Galvanic corrosion around a scratch or pit in the plate.[26]

traumatic event or can be caused by cyclical loading and fatigue failure at the stress riser. Thus, in this situation a longer plate should be used to bypass the high-stress area, particularly in areas of poor bone quality.

A scratch can also cause a local small stress concentrator. When immersed in the saline environment of the body, *stress corrosion* can occur. Stress corrosion combines the effects of the local growth of the crack resulting from cyclic loading with galvanic corrosion. A *galvanic cell* describes a local environment in which electrons flow from the more negative to the more positive material when immersed in a liquid conductor (saline, in this case) (Fig. 1-9). Material is actually removed from the more negative electrode, such as the surface of the plate during galvanic corrosion. In a fixed fracture, the dissimilar materials are the surface of the plate (e.g., stainless steel), which creates an oxide surface coating, and the same material exposed by the fatigue crack that has not yet developed the oxide film. The conductive fluid is the blood and saline found in the surrounding tissues. Galvanic corrosion can accelerate the failure of an implant, even when the implant is loaded well below its yield point, by increasing the rate at which the crack grows. This occurs because in addition to the mechanical propagation at the site of the crack, material at the crack is being removed by the corrosion process. Another mechanism of corrosion, termed *fretting,* results when the surfaces of two implants rub together, such as the head of a screw against the surface of the plate through which it passes. *Crevice corrosion,* which is not common in modern orthopedic materials, results from small galvanic cells formed by impurities in the surface of the implant, causing crevices as the material corrodes.[26]

Another basic property is *viscoelasticity* (Fig. 1-10). Biologic materials do not act as pure springs when load is applied to them. A spring deforms under load and then returns to its original shape when the load is released. For example, if a load is applied to a tendon, and the load is maintained for a period of time, the tissue will continue to deform or *creep.* This is the basic principle behind stretching exercises. Under a constant load, a metal fixation plate will deform and remain at that deformation until the load is removed (elastic behavior). In contrast, the tendon both deforms elastically and creeps, exhibiting both viscous and elastic behavior. This property has important implications for certain types of fixation, especially those that rely on loading of soft tissues, such as in certain types of spinal fixation (to be discussed later).

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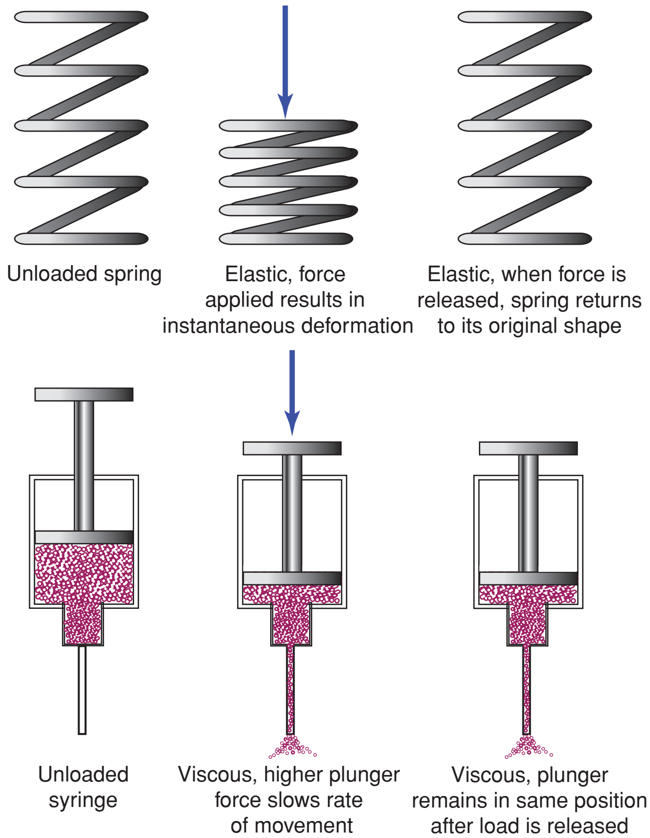

FIGURE 1-10 Viscoelastic response in a biologic tissue can be explained by considering and combining the properties of two devices, a simple spring and a fluid-filled syringe. The elastic or spring component instantly compresses when a load is applied to it. When the load is released, the spring returns to its original shape. When a load is applied to the viscous component, represented by the syringe, fluid is forced out of the needle. If the load is released, the plunger does not return, but remains in its final position, representing the creep property of the tissue. Further, if the force is applied to the plunger more rapidly, there is greater resistance to motion, explaining the increased stiffness of tissue to increased rates of loading. Combinations of these simple components can be used to describe the mechanical properties of biologic tissues.

A second characteristic of viscoelastic behavior is *loading rate dependence.* In simple terms, stretching a soft tissue can be thought of as stretching two components, an elastic one and a viscous one, which make up that tissue. For example, consider a spring connected in series to the handle of a syringe. When a compressive force is applied, the spring instantly compresses, representing the elastic response of the tissue. The syringe plunger starts to displace and continues as it pushes fluid through the orifice. If the force is held constant, the plunger will continue to move, representing the viscous creep of the tissue. If the compressive force is applied slowly, the syringe handle offers little resistance. As the rate of force application increases, the resistance of the syringe to motion increases. This represents the increase in stiffness of the tissue at higher loading rates. Simply put, the stiffness of the tissue depends upon the rate at which the load is applied.

A well-known example of loading rate dependence relates to the failure of ligament and bone. At low loading rates, the ligament is weaker than the bone and the ligament generally fails in the midsubstance. At higher loading rates, the ligament becomes stiffer, and failure may occur by avulsion of the bony attachment of the ligament. *Stress relaxation* occurs if the applied force, instead of increasing, is held constant. As the fluid flows out of the syringe, without further movement of the plunger, the internal force decreases. These three properties—creep, stress relaxation, and load rate dependence—make up the basic tissue viscoelastic properties. It should be appreciated that the model used in this discussion is a simple linear series model, for explanation purposes only. Nevertheless, more complex models using combinations of these basic components have successfully described the observed properties of tissues. Another example of tissue viscoelasticity, besides tendon and other soft tissues, is found in trabecular bone (e.g., as found in vertebrae). In this case, the trabecular structure acts as the spring component, whereas forcing the interstitial fluid through the porous matrix as the trabeculae deform represents the viscous component. Under higher loading rates, there is resistance to flow, increasing the internal pressure and therefore the stiffness of the structure. These effects have been observed at high loading rates, such as during fracture (Fig. 1-11).[34]

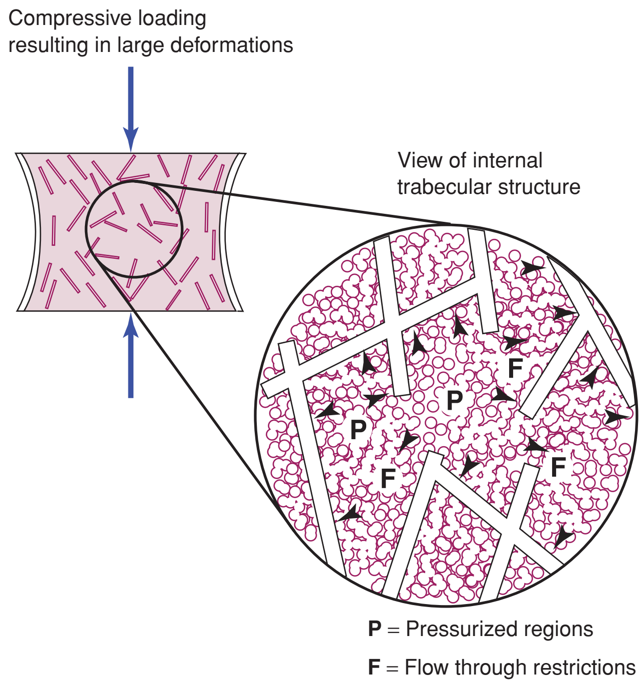

FIGURE 1-11 The trabecular bone possesses some features of the spring and syringe viscoelastic model described in Figure 1-10, although it should be appreciated that this is an idealized model. The trabecular structure acts as the spring element. At higher loading rates, the interstitial fluid resists flowing through the trabecular spaces, causing increased internal pressure and greater bone stiffness. This anatomical feature allows vertebrae and the metaphyseal ends of long bones to resist dynamic loads caused by rapidly applied forces.[34]

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