Essential Biomechanics for Orthopedic Trauma

A Case-Based Guide

Brett D. Crist
Joseph Borrelli Jr.
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Preface

Thank you for your interest in *Essential Biomechanics for Orthopedic Trauma: A Case-Based Guide*. The goal of this book is to make the topic of biomechanics, which is so critical to understand in managing fractures, clinically relevant for learners, including medical students, biomedical engineering students, advanced practice providers, orthopedic residents and fellows, and practicing orthopedic surgeons.

We organized this book to work through the progression of fracture care from a biomechanical standpoint. We discuss the principles of bone healing and the different techniques used in managing fractures. By using actual cases, including examples of successes and failures, we hope to solidify the biomechanical principles that affect our ability to succeed as orthopedic surgeons managing fractures and deformity.

We are very fortunate to have so many thought leaders contribute to this book. The contributors were chosen because they are considered experts in the field of fracture and deformity care, and are great educators.

We would like to thank our families for their support and tolerance/patience with us during the time it has taken to complete the project. We would also like to acknowledge Springer, particularly Kristopher Spring and Katherine Kreilkamp, for bringing this book to reality and being patient with the process. Finally, I would like to acknowledge my coeditors—Ed Harvey and Joe Borrelli. They have been critical to the book's success from developing the book concept, choosing contributors, editing and contributing to the chapters, and going the extra mile in getting it across the finish line.

Enjoy,

Columbia, MO, USA Brett D. Crist, MD

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Part I

Stress, Strain, and Young's Modulus—How They Relate to Fracture Healing

Biomechanical Principles of Fracture Healing

1

Sarah H. McBride-Gagyi and Maureen E. Lynch

Introduction

The bone is remarkably sensitive to its mechanical environment [1–3]. Throughout life, the overall structure is constantly adapting to local mechanical stimuli – adding tissue in areas under increased mechanical loading and removing tissue where it is not utilized. Consequently, the bone is a highly organized tissue with minimal weight and maximal strength to meet the demands of daily physical activity. This concept is known as Wolff's law [3]. Wolff's law is also involved during bone repair, where the type and amount of repair tissue differ significantly based on the repair site's mechanical stability and whole limb loading [4–6]. Thus, understanding basic mechanics is very important for orthopedists aspiring to optimize bone repair in their patients.

The forces a single bone, fracture callus, or implant can withstand are a combination of both its material strength and structural strength (Fig. 1.1) [7–13]. Material strength is an intrinsic property, like density or temperature, which is independent of material quantity, whereas structural strength is a function of not only size but also the distribution of that material. For example, consider a solid cantilevered beam with a rectangular cross section subjected to a bending force at one end. If the beam were made of steel, a much larger force would be required to bend or break it than if the same beam were made of Styrofoam (Dow Chemical, Midland, MI, USA) because of a difference in the *material strength* of steel versus Styrofoam. An inch-thick beam of Styrofoam, however, will require less force to bend or break than a foot-thick beam of Styrofoam. The increased strength in the foot-thick Styrofoam beam is due to greater *structural strength* via larger size. Alternatively, if the material of a very large Styrofoam beam was rearranged and optimized to oppose the loading (e.g., an I-beam cross section), it could be as strong as the steel beam because of the way the material was distributed.

Material Strength

Stress-Strain

When material strength is discussed, the terms stress and strain are used rather than force and displacement (or deformation). Stress and strain are the latter terms normalized by an object's dimensions. Stress, σ[sigma] or τ[tau], is force normalized by the cross-sectional area it is

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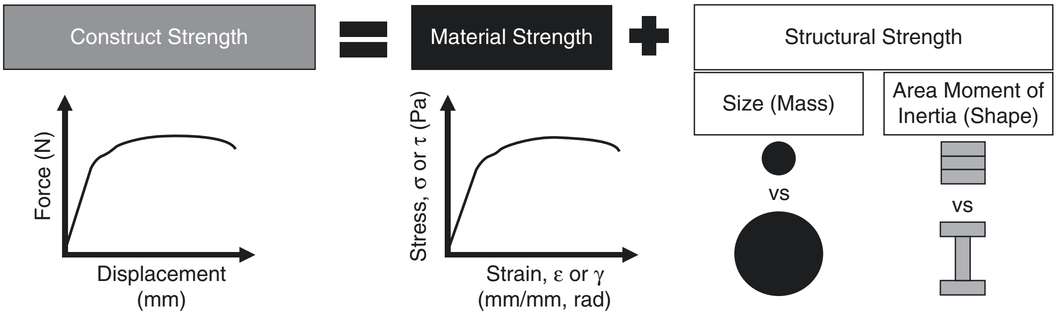
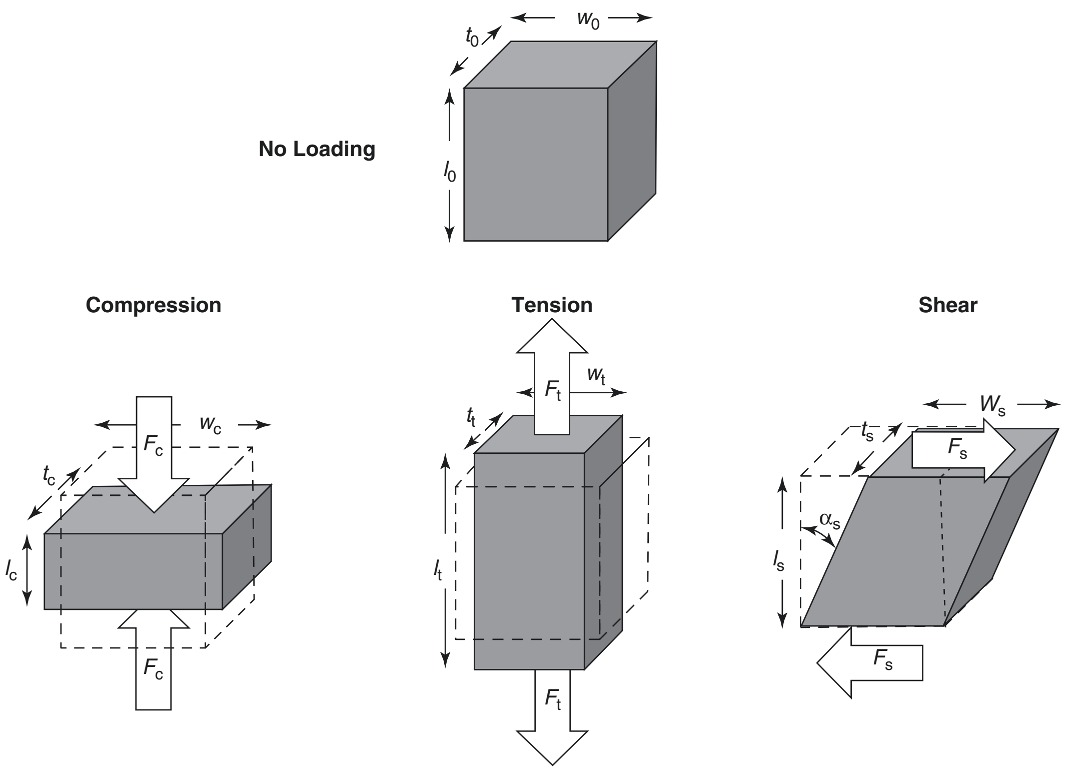


Fig. 1.1 Whole object strength. The forces (and displacements) any object can withstand are a combination of the strength of its constituting material and the structural strength imparted by its size and area moment of inertia (cross-sectional shape)

applied to, while strain, ε[epsilon] or γ[gamma], is the displacement (i.e., change in length) normalized by the original length or the angular displacement, respectively (Fig. 1.2). Stress and strain can be compressive, tensile, or shear depending on whether the object is shortened, lengthened, or angularly deformed, respectively (see Fig. 1.2). Compressive stresses/strains are negative. Tensile stresses/strains are positive. The sign convention for shear stresses/strains depends on the rotational direction (clockwise vs. counterclockwise). Stress uses the unit of Pascals (Pa) which is 1 N/m^2. Biological tissues are usually presented as kPa (10^3), MPa (10^6), or GPa (10^9). Strain is unitless or in radians. Bone strains are usually expressed as microstrain (μ[mu]strain, μ[mu]ε[epsilon] 10^{-6}). Material characteristics, like elasticity, ductility, strength, and toughness, are determined from the stress-strain curve, which is generated from a mechanical test. For practical reasons, most stress-strain curves used for engineering purposes are for tensile or shear forces [13, 14]. Bone, like many other materials, is much stronger in compression than in tensile or shear [7–9, 11]. Therefore, even when subjected to what may seem like strictly compressive forces (e.g., gait loading at the femoral head), objects like bones tend to fail in tension or shear, caused by bending or torsional forces accompanying compression, because those material limits are reached first. For simplification, we will only use tensile or positive shear stresses/strains as examples for the remainder of this section.

There are several important features of the stress-strain curve that are used to determine material properties. The yield point is the stress at which the curve changes from linear to nonlinear, delineating the elastic region (the linear region) from the plastic region (the nonlinear region) (Fig. 1.3). Identifying this transition point can be somewhat subjective. In order to uniformly analyze data, two methods are often employed. In engineering, the intersection of the stress-strain curve and a line parallel to the linear region with a 0.2% offset (x-intercept) is most often used (see Fig. 1.3a) [7, 9, 11, 13]. For biological tissues, the intersection with a line that is 10% less stiff (−10% slope) and that shares the same origin is acceptable as well (see Fig. 1.3b) [12].

When subjected to stresses/strains within the elastic region, the material has not accumulated significant damage and will return to its original shape when the stress is removed. This is referred to as reversible deformation. In theory, the material acts as a perfect spring, releasing all potential energy stored by deformation once the load is removed [13]. The slope of the curve within the elastic region defines the material's stiffness and is formally referred to as the Young's or elastic modulus (see Fig. 1.3c). Compliant or elastic materials will deform a great deal under very little stress; correspondingly, the Young's modulus (the slope) will be low. A stiff material will deform very little under the same stress, thus will have a high Young's modulus (slope). In mineralized tissues such as bone, the Young's modulus is governed predominately by the mineral component



$$\sigma_c = -\frac{F_c}{A_0} = -\frac{F_c}{w_0 * t_0}$$

$$\varepsilon_c = -\frac{l_0 - l_c}{l_0}$$

$$\sigma_t = +\frac{F_t}{A_0} = +\frac{F_t}{w_0 * t_0}$$

$$\varepsilon_c = +\frac{l_0 - l_t}{l_0}$$

$$\tau_S = \pm\frac{F_s}{A_0} = \pm\frac{F_s}{w_0 * t_0}$$

$$\gamma_s \approx \pm\alpha_s$$

Fig. 1.2 Types of stresses and strains. Stresses/strains can be compressive, tensile, or shear, depending on whether they cause shortening, lengthening, or change in shape, respectively. Stress is the applied force divided by the original cross-sectional area to which it is applied (orthogonal for compression/tension and parallel for shear). Compressive/tensile strain is the change in overall length divided by the original length. Shear strain for small deviations is approximately equal to the change in angle

[7–9, 11, 15–17]. Denser bones are generally stiffer, though other components (e.g., noncollagenous proteins, collagen cross-links) have an effect. Furthermore, mineral density is typically spatially inhomogeneous within a single bone; thus the local modulus varies throughout. This is especially true for fracture calluses, where the older woven bone at the far proximal and distal ends of the callus is denser and stiffer than newer woven bone closer to the fracture gap [16, 17].

Stiffness is often confused with strength. Though a material is stiff and has a high Young's modulus, it may not necessarily be strong. Strength refers to the maximal, or "ultimate," stress a material can withstand, which almost universally occurs in the plastic rather than the elastic region (see Fig. 1.3). However, some inferences about strength can be made from the elastic linear region, especially since most materials/structures are meant to operate well below yield for long periods of time. If the elastic region is large and extends to high stresses, then the ultimate stress must be even higher – such a material is considered to be strong. For example,

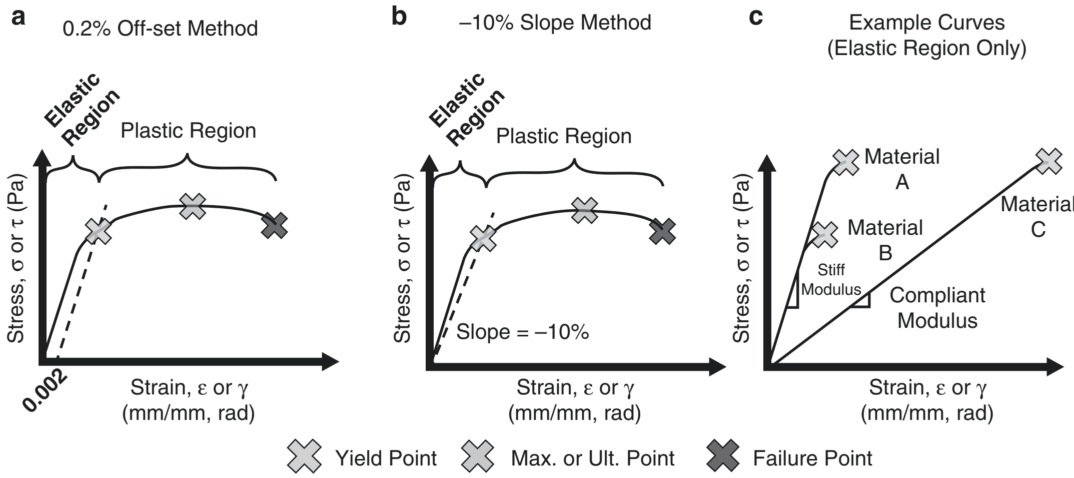


Fig. 1.3 Yield point identification and examples of stiffnesses/strengths. The yield point separates the elastic region from the plastic region. Systematically identifying this point is typically done by (**a**) the 0.2% yield method. (**b**) However, when dealing with biological specimens, the −10% method can be used on either a stress-strain curve or force-displacement curve if the former is not feasible to obtain. (**c**) Materials A and B have the same stiffness (Young's modulus), but Material A is likely stronger than Material B because its yield point is at a much higher stress. Material C is more compliant than the other two materials (lower Young's modulus) but is likely just as strong as Material A because the yield point is at a similar stress

consider three different materials and the linear regions (up to the yield point) of their respective stress-strain curves shown in Fig. 1.3c. Material A and Material B have the same Young's modulus, and both are stiffer than Material C. Material A is probably stronger than Material B since its yield point is much higher. In contrast, because their yield points are similar, Materials A and C are probably closer in strength, though Material C will be more deformed than Material A at any given stress. To truly compare ultimate strengths, however, the entire stress-strain curve, elastic and plastic regions, would need to be assessed.

To the right of the yield point, where the curve is nonlinear, is the plastic region. Once the applied stress/strain crosses into the plastic region, significant damage has occurred, and the material will not return to its original shape when the stress is removed (irreversible deformation) (Fig. 1.4a). In a uniaxial tension test, the damage incurred after the ultimate stress is reached can be visualized by "necking" within the material (Fig. 1.4b) [14]. This region defines a material's ductility. If it can withstand significantly more (plastic) deformation after yield, it is called ductile (Fig. 1.5a, b). If it quickly reaches fracture without much plastic deformation, then it is called brittle (Fig. 1.5c, d). The ultimate stress, which is the maximal stress on the curve, is almost always found in this region. For ductile materials, ultimate stress may be a good deal greater than the yield or failure stress (see Fig. 1.5b); for very brittle materials, it may be the same as or only a little higher than the yield or failure stress (see Fig. 1.5c). In bone and most biological tissues, ductility is governed by the collagen aspect [7, 9, 11, 18, 19]. Changes in collagen with disease and aging tend to make the bone more brittle [18].

The final property that is determined from the stress-strain curve is toughness. Mathematically, toughness is the area under the whole stress-strain curve. It represents the energy per unit volume that the material can withstand before fracturing. Toughness is sometimes divided into pre- and post-yield values since it is usually desirable to operate in the elastic zone where no damage is incurred. Toughness is a function of compliance, strength, and ductility, so it is often a suitable single measure when comparing resistance to fracture across materials or groups.

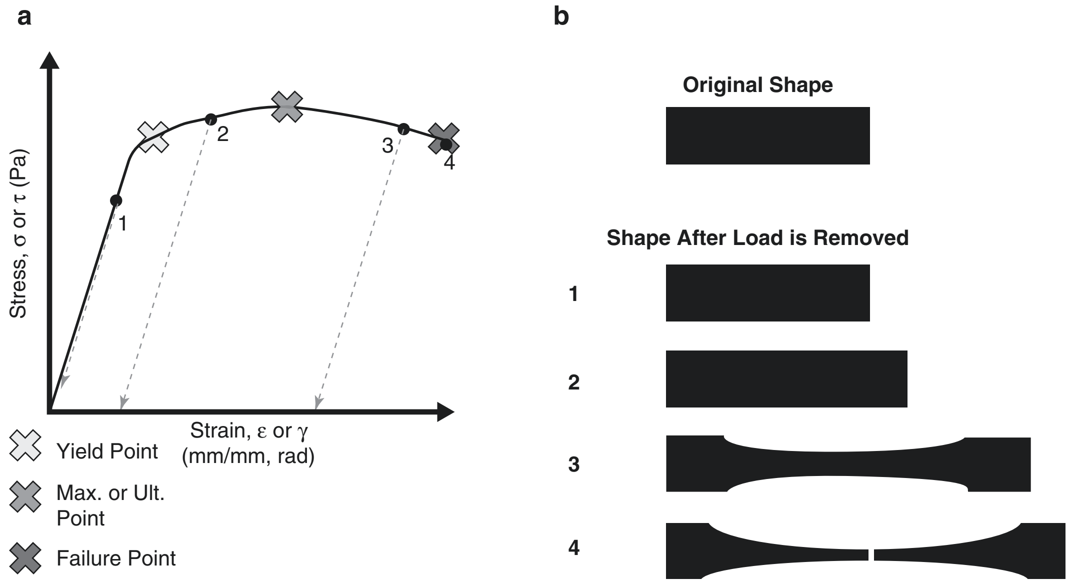


Fig. 1.4 Example of elastic and plastic deformation for a ductile material. (**a**) The stress-strain curve for a material showing four different points it could be loaded to and then the load removed. (**b**) When the object is stretched to point 1, which is within the elastic region, it returns to its original shape. When stretched to point 2, it has passed yield and entered the plastic region. Some permanent damage has been done, so the object is a little longer, but basically the same shape after loading. Point 3 is past the ultimate stress, so not only is it longer when the load is removed, it has begun to "neck." At the final point, point 4, failure has been reached and the object has fractured into two pieces that are longer and have significant "necking"

When discussing material properties, two important concepts should be kept in mind. First, many of the aforementioned terms are relative. Assessment of stiffness, ductility, and strength is highly dependent upon what the normal or control condition is. For example, a "tough" piece of tendon probably absorbs orders of magnitude *less energy* than a "weak" piece of the bone. A "ductile" piece of metal will probably *deform less* past its yield point than a "brittle" piece of plastic. So, it is important to report absolute numbers for each tissue/material when possible for comparison across studies/materials (Table 1.1) [20]. Second, the shape of force-displacement curves is visually very similar to that of stress-strain curves, and each intrinsic material property discussed here has an extrinsic, non-normalized analogue on the force-displacement curve. Many terms are used interchangeably. For example, "stiffness" can refer to the slope of the stress-strain curve which has units of Pascals (Pa/unitless) or the slope of the force-displacement curve which has units N/mm. However, Young's modulus only refers to the slope of the stress-strain curve. For biological specimens, analogues from the force-displacement curve are often reported due to technical limitations prohibiting normalization, such as isolating samples of uniform geometry, accurately characterizing complex morphologies, or meeting testing standards/assumptions [10, 12].

Anisotropy

Isotropy describes material properties and mechanical behavior that are identical in all loading directions (Fig. 1.6a). This is the case for most homogeneous materials, such as the metals or plastics implants are made of. In reality, however, biological tissues are rarely homogeneous at the tissue level. They are typically a composite of very organized collagens and other matrix proteins and, in the case of the bone and later stage fracture cartilage, mineral. Composite materials are typically anisotropic, indicating

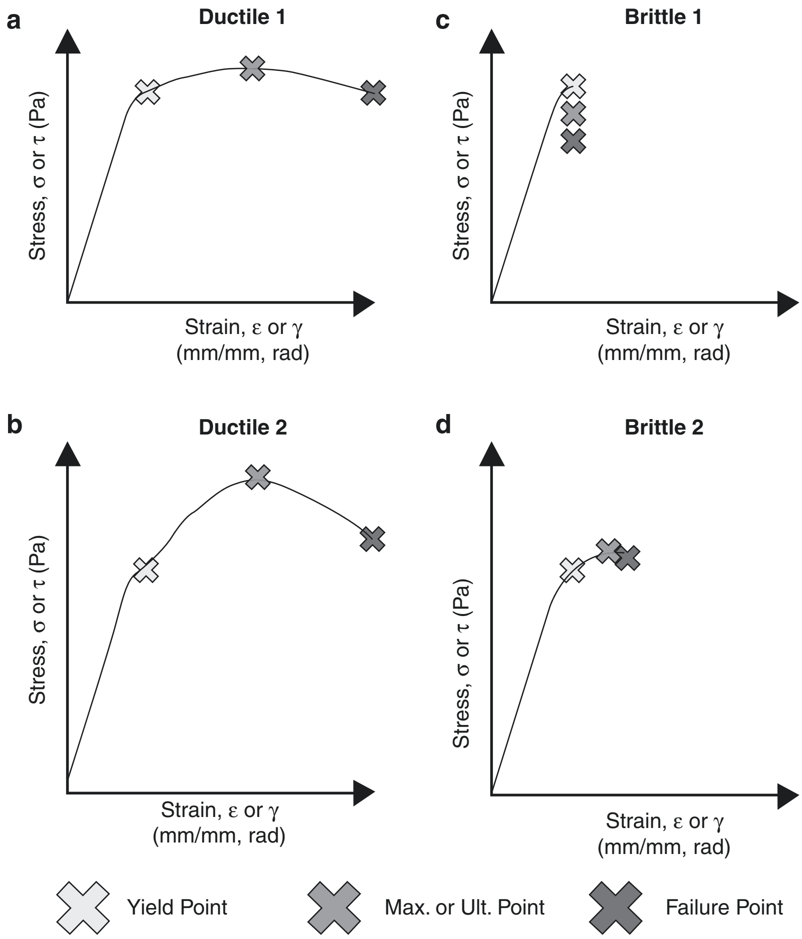


Fig. 1.5 Examples of ductile and brittle stress-strain curves. Here all four materials are of a similar compliance (same Young's modulus) with the same yield point and will behave the same while operating in the elastic region. However, the materials in panels A and B are ductile and those in panels C and D are brittle. Meaning that Materials A and B will deform much further before failing. (**a**) Ductile material 1 will not tolerate a stress much larger than yield. (**b**) Ductile material 2 reaches a much higher ultimate stress. (**c**) Brittle material 1 is extremely brittle. The yield point is also the ultimate stress point and failure point. (**d**) Brittle material 2 has an ultimate stress and failure stress slightly above yield

that their properties and behavior depend on the loading direction (Fig. 1.6b). Mechanical anisotropy is important evolutionarily because it allows optimization of biological tissues to their primary loading direction(s) [21, 22]. Thus, they can withstand their daily loading demands with minimal material. For example, in diaphysial cortical bone, this is achieved via the lamellar and osteonal hierarchy where the collagen fibrils are predominately aligned with the bone's longitudinal axis [18]. Cortical bone is considered orthotropic or transversely isotropic. It is strongest in the axial direction with weaker, but similar to each other, material properties in either transverse direction (anterior-posterior and medial-lateral) (Fig. 1.6c, see Table 1.1) [11]. Cancellous bone achieves its anisotropy via the trabecular orientation. It will be strongest in whatever direction the majority of trabeculae are aligned [7]. For example, in the femoral head, the force of one's body weight causes bending, resulting in tension on the lateral side and compression on the medial side of the femoral shaft, and the trabeculae follow these two primary directional patterns (Fig. 1.6d). Depending on the length scale, this could be considered a structural effect rather than a material property. Some tissues important in bone repair, such as clots, some cartilages, and woven bone, are not anisotropic. In these tissues, the matrix fibers are relatively disorganized since they were created quickly [1, 21–25]. So, although they are composites, they are likely mechanically similar in all directions [26].

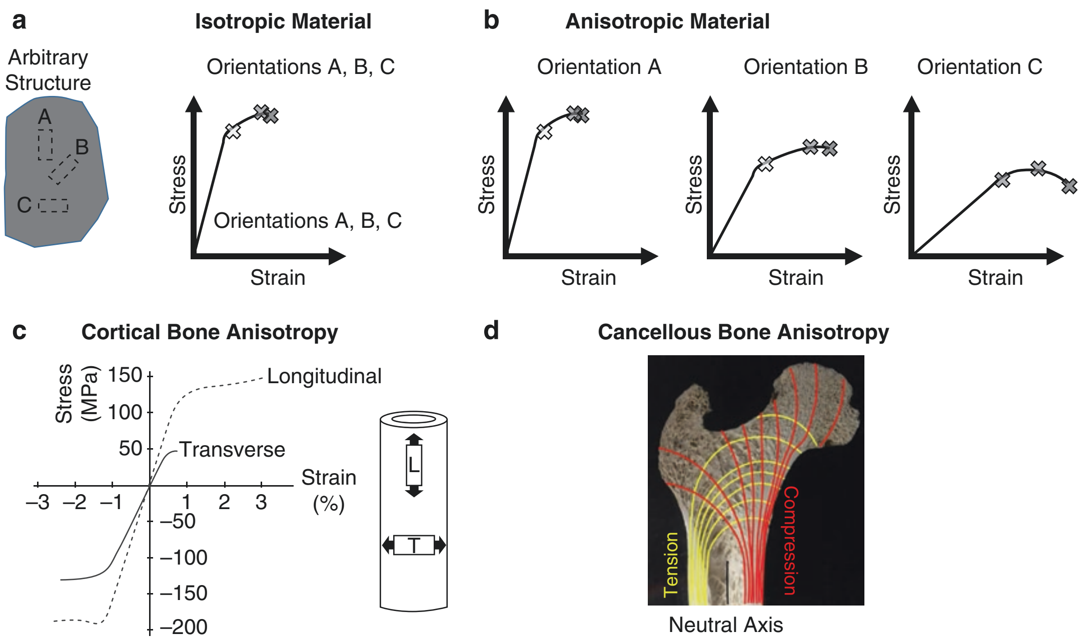


Fig. 1.6 Isotropic vs. anisotropic material properties. Uniform specimens A, B, and C are cut from an arbitrary material at different orientations for uniaxial testing. (**a**) If the material is isotropic, identical stress-strain curves will be obtained for all specimens, and the material will behave identically no matter from which direction stress is applied. (**b**) If the material is anisotropic, the stress-strain curves will differ among specimens A, B, and C, and the material will behave differently depending on the direction of an applied load. (**c**) Cortical bone exhibits anisotropy in both direction of specimen (longitudinal versus transverse) and in loading direction (tension versus compression). (**d**) Cancellous bone exhibits anisotropy based on principal orientation of the trabecular architecture. Shown are the principal trabecular orientations throughout the femoral head, following tensile and compressive loading directions due to applied bending

Table 1.1 Average material properties for bone and common implant materials

Implant materials	Modulus (GPa)	Ultimate stress (MPa)	
		Tension	Compression
Cortical longitudinal	17	133	193
Cortical transverse	11.5	51	133
Cortical shear	3.3	68	
Stainless steel	190–210	586–1351	
Cobalt-chromium alloy	210	770–1500	
Titanium	97–116	240–965	
UHMWPE	1	45	
PMMA	2.3–2.7	35	

Values from Bartel et al. [20]
UHMWPE Ultra-high-molecular-weight polyethylene, *PMMA* Poly(methyl methacrylate), also known as acrylic or acrylic glass

Fatigue

Fatigue loading is repetitive loading below a material's ultimate or fracture point that weakens the material over time [11, 13] The stress-strain curves, and associated material properties, discussed thus far have been for monotonic or single loading bouts. Failure under fatigue loading occurs because the applied stress-strain during an initial cycle incurs some irreversible deformation or damage. When unloaded, the material does not return to its original form, and some strains remain (i.e., the strain values do not return to zero, the origin on the stress-strain curve) (Fig. 1.7a). For purely elastic materials, the unloading curve parallels the elastic region, stopping at a point on the positive x-axis. The next loading cycle begins elastically from the new starting point until reaching the (new) yield point. More damage (plastic deformation) incurs, thus pushing the unloading curve further to the right for each subsequent cycle, and this repeats until the fracture point is reached. Thus, the material fails without ever reaching the monotonic ultimate stress [13].

Most biological materials and implants are meant to be repetitively loaded for decades without breaking. So, they must operate well below the ultimate stress-strain point. To determine how far below they must operate, fatigue graphs are used (Fig. 1.7b). Fatigue graphs, also known as S-N curves (stress-number of cycles), display how many cycles a material can withstand at any given stress value before fracturing [13]. Typically, the graph is on a log scale and the curve has two parts. The curve's first part begins at the ultimate strength and decreases as the number of cycles increases. At some point the curve becomes horizontal. This stress value is known as the endurance limit. This is the maximal stress which, in theory, the material could be loaded to for an infinite number of cycles. However, it should be noted that these curves are developed for perfectly intact specimens with no flaws to cause stress concentrations (see Stress Concentrations below). Additionally, S-N curves are determined from ex vivo specimens, thus neglecting the remodeling and repair capabilities of living bone [27, 28]. Targeted repair of damaged tissue allows living bone to operate at higher stresses for far more cycles than explanted, devitalized bone (provided there is ample time between loading bouts for repair). Metals and plastics obviously have no such self-repair mechanism.

Viscoelasticity

Viscoelastic materials exhibit material properties that behave more like a viscous liquid than a purely elastic material [29]. Specifically, the material will exhibit rate-dependent compliance and stress/strain relaxation. For purely elastic

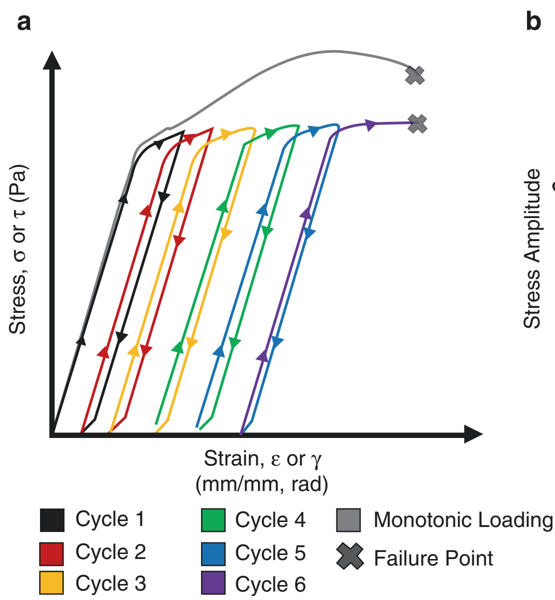


Fig. 1.7 Fatigue failure. (**a**) The material is loaded repeatedly to a stress that is technically less than its ultimate stress. However, because each cycle is causing some new plastic deformations, the damage accumulates until the material fails at the lower stress level. (**b**) S-N curves like the one shown here are used to determine how many cycles a material can withstand when loaded up to any particular stress. Since monotonic loading is a single cycle and the most a material can withstand, the curve always starts at the ultimate load from the stress-strain curve. Many materials have an endurance limit, which is the maximal stress at which a material could, in theory, be cyclically loaded to and never fail

materials like implant metals, the stress-strain curve will be the same and result in the same Young's modulus regardless of how slowly or quickly the load is applied. For viscoelastic materials, the material stiffens when the loads are applied faster, resulting in a higher Young's modulus than if the loads were applied slower (Fig. 1.8a). Additionally, a purely elastic material loaded at a constant stress/strain within the elastic range will hold the corresponding strain/stress indefinitely (Fig. 1.8b, middle panel). In contrast, a viscoelastic material held at a constant stress/strain will slowly relax, causing the dependent parameter to lessen over time (see Fig. 1.8b, bottom panel). Any hydrated material, like the bone or other biological tissues, will have some viscoelastic behaviors. However, viscoelastic effects on bone are typically minimal and are often ignored outside of selecting a physiologically relevant loading rate [11]. On the other hand, more compliant tissues (e.g., tendon, cartilage, and hematoma) or polymer plastics can exhibit large changes in modulus for small changes in loading rate and have significant stress-strain relaxation [23, 24, 29]. Therefore, when dealing with these tissues or materials, viscoelastic effects should be considered.

Structural Strength

Structural strength is an extrinsic property of any object and helps define the absolute force-displacement that object can withstand. Structural strength is the combination of two components, size and material distribution.

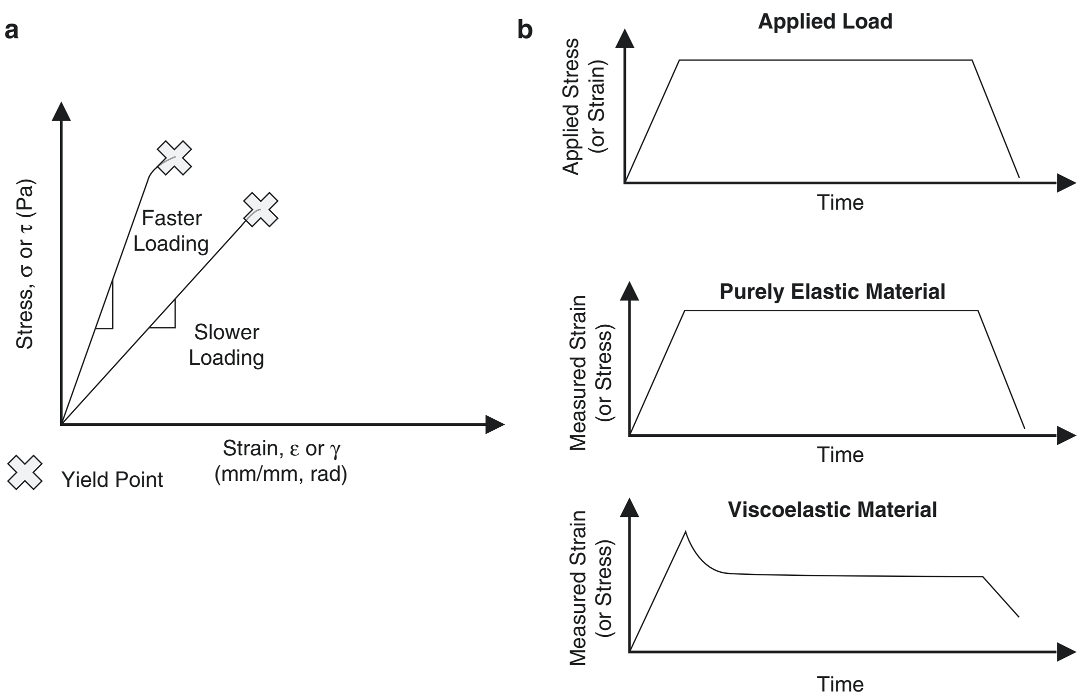


Fig. 1.8 Viscoelastic behaviors. Most biological materials and some engineered materials like polymer plastics do not behave as a pure elastic material. They have some behaviors like a viscous liquid. This usually manifests as rate-dependent compliance and stress/strain relaxation. (**a**) Rate-dependent compliance is when the Young's modulus is higher when the force is applied at a faster rate. (**b**) Stress/strain relaxation is when the material is held at a constant stress (or strain) the corresponding strain (or stress) is not static. It will lessen over time as the material "relaxes." When the load is removed, the material may or may not return to its original state

Size

The effect of size on strength is usually easy to understand. A larger object will take more force to deform and break than a smaller object made of the same material, as was described at the beginning of this chapter using the example of steel and styrofoam beams. Thus, the larger object is "stronger" in an absolute sense [7, 11]. Of course, if each object were normalized to its size (i.e., cross-sectional area), the stress-strain curves would be identical. Thus, if one is trying to construct a stronger and therefore more stable implant, this can be achieved by simply increasing the overall mass. The larger size will increase the forces the object can withstand in all loading modes (pure compression/tension, bending, and torsion).

Material Distribution

The consequences of altered material distribution or cross-sectional shape are more complex than for size. Also, its effects are mostly applicable to bending or torsional loading and not pure compression/tension. This is extremely important for skeletal and fracture callus tissues as well as implants, because bending and torsion are the most predominate loading modes in vivo [7]. When an object is loaded in bending, one side is under compression (negative values) while the opposite side is under tension (positive values) (Fig. 1.9a). The location where the two loading modes meet is under zero loading and is termed the neutral axis. For torsion, the neutral axis is at the axial center of the object with only tension/shear stresses extending radially (Fig. 1.9b). Much like a seesaw or a lever arm, the farther away from the neutral axis (fulcrum) tissue is located, the greater its moment arm and therefore the more resistance to bending or torque it can provide. In bending, this is quantitatively represented by the object's area moment of inertia, I_x or I_y, depending on the loading direction. For torsional loads, the polar moment of inertia is used, J or I_z, which is the sum of I_x and I_y. Thus, it will take much less force to bend or twist a given amount of tissue to failure if it is a solid rod rather than a hollow tube (Table 1.2, columns A and B) [7, 10–13]. The tubular shape allows the mass or area to be distributed further away from the neutral axis, so the area and polar moments of inertia are larger. However, because the cross-sectional area is identical, both objects will behave the same under pure compression/tension. Likewise, weaker tissues or less tissue can provide more structural strength by being distributed further away from the neutral axis (Table 1.2, column C). This is hypothesized as the reason why fracture callus size scales with fixation stability [4, 5]. Lamellar bone is much stronger but more time-consuming to create than woven bone or cartilage. Thus, the body uses the weaker materials of cartilage and woven bone, which can be created rapidly, at greater volume and over a wider area to restore close to pre-injury strength relatively quickly [30, 31]. Also, if there is a primary loading direction, then more tissue/material can be devoted to opposing that load, resulting in a functionally stronger construct for less mass (Table 1.2, column D).

Stress Concentrations

Sometimes objects fail or fracture at forces that should cause stresses lower than the yield or ultimate. While fracture can certainly be due to fatigue failure if the force is causing plastic stresses throughout the object, it can also be a consequence of localized stress concentrations (or the combination of the two). Stress concentrations, sometimes called stress risers, are geometrical features (e.g., holes, cracks, and fillets) or sudden changes in material properties that cause a localized area of increased stress/strain [13]. So, in a way, this is another example of structural strength. These small regions can exhibit stresses severalfold higher than the majority of the object. Thus, localized failure at these regions occurs immediately or over time, which can coalesce to cause macroscopic fracture. A classic example are circular holes. When a tensile force is applied to an object with a hole (such as from screw holes in cortical bone), the stresses in most of the object may be well below yield stress. However, the material on either side of that hole is calculated to undergo 3× higher stress, which could be well

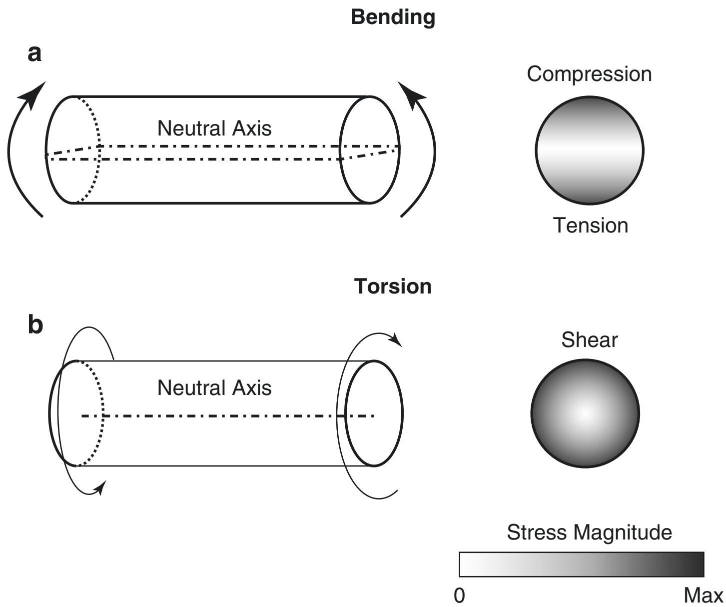
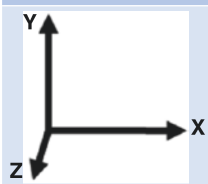
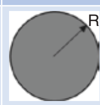
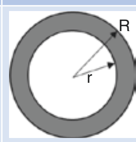
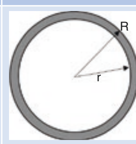
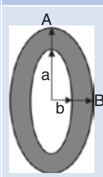


Fig. 1.9 Stress distributions. (**a**) Under bending loads, also called bending moments, the neutral axis is a plane at some point in the middle of the object. Compression is on one side and tension on the other. (**b**) Under torsional loads, the neutral axis is a vector with shear strains increasing radially outward

Table 1.2 Comparisons of structural strength (Equations from Avallone and Baumeister [13])

	A	B	C	D
				
I_x	$\frac{\pi R^4}{4}$	$\frac{\pi\left(R^4-r^4\right)}{4}$	$\frac{\pi\left(R^4-r^4\right)}{4}$	$\frac{\pi\left(A^3B-a^3b\right)}{4}$
I_y	$\frac{\pi R^4}{4}$	$\frac{\pi\left(R^4-r^4\right)}{4}$	$\frac{\pi\left(R^4-r^4\right)}{4}$	$\frac{\pi\left(B^3A-b^3a\right)}{4}$
J	$\frac{\pi R^4}{2}$	$\frac{\pi\left(R^4-r^4\right)}{2}$	$\frac{\pi\left(R^4-r^4\right)}{2}$	$\frac{\pi\left[AB\left(A^2+B^2\right)-ab\left(a^2+b^2\right)\right]}{4}$
Area	100%	100%	85%	100%
Resistance to compression/tension	100%	100%	85%	100%
Resistance to bending about *x*-axis	100%	299%	528%	344%
Resistance to bending about *y*-axis	100%	299%	528%	125%
Resistance to torsion	100%	299%	528%	234%

beyond the yield point and thus cause failure [13]. Furthermore, if multiple holes are within close enough proximity, a single loading event could cause fracture at a much lower force than if the object were solid. For example, perforated paper like stamps or notebook paper is much easier to tear than a solid sheet because the perforations cause stress concentrations that align.

Mature bone, fracture callus, and implants all have stress concentrators on multiple length scales [32]. Mature bone has osteocyte lacunae/canaliculi, Haversian canals, macropores, and muscle insertion sites. The fracture callus has vascular channels and transition areas between tissues of differing compliances. Finally, common orthopedic implants have screw holes and sharp corners. Failure is more likely to occur in any of these regions than elsewhere.

Clinical Implications

Understanding basic mechanical principles is important clinically for several key reasons. First, implants need to be designed and installed so that they do not fail prematurely. Conducting a revision surgery not only increases the risk of morbidity and mortality for the patient, but they are also costly. Second, it is important to create an appropriate mechanical environment that optimizes healing. The bone and fracture callus are sensitive to the loads imposed upon them and will adapt as best they can to those loads. Too much or too little mechanical stimulus can have detrimental effects on ultimate bone repair. As discussed in detail in later chapters, the mechanical environment is greatly impacted by both fixation methods (implant installation) and physical therapy.

References

1. McBride SH, Silva MJ. Adaptive and injury response of bone to mechanical loading. Bonekey Osteovision. 2012;1:pii: 192.
2. Knothe Tate ML, Falls TD, McBride SH, Atit R, Knothe UR. Mechanical modulation of osteochondroprogenitor cell fate. Int J Biochem Cell Biol. 2008;40(12):2720–38.
3. Chen J-H, Liu C, You L, Simmons CA. Boning up on Wolff's Law: mechanical regulation of the cells that make and maintain bone. J Biomech. 2010;43(1):108–18.
4. Glatt V, Evans CH, Tetsworth K. A concert between biology and biomechanics: the influence of the mechanical environment on bone healing. Front Physiol. 2017;7:678. https://doi.org/10.3389/fphys.2016.00678.
5. Morgan EF, Gleason RE, Hayward LN, Leong PL, Palomares KTS. Mechanotransduction and fracture repair. J Bone Joint Surg Am. 2008;90(Suppl 1):25–30. Review.
6. Morgan EF, Lei J. Toward clinical application and molecular understanding of the mechanobiology of bone healing. Clin Rev Bone Miner Metab. 2015;13(4):256–65.
7. Cole JH, van der Meulen MCH. Whole bone mechanics and bone quality. Clin Orthop Relat Res. 2011;469(8):2139–49.
8. Keaveny TM, Morgan EF, Yeh OC. Bone mechanics. In: Kutz M, editor. Standard handbook of biomedical engineering and design. New York: McGraw Hill Professional; 2003. Access Engineering. https://www.accessengineeringlibrary.com/content/book/9780071498388.
9. Karim L, Hussein AI, Morgan EF, Bouxsein ML. The mechanical behavior of bone. In: Marcus R, Feldman D, Dempster DW, Luckey M, Cauley JA, editors. Osteoporosis. 4th ed. San Diego: Academic; 2013. p. 431–52.
10. Sharir A, Barak MM, Shahar R. Whole bone mechanics and mechanical testing. Vet J. 2008;177(1):8–17.
11. Morgan EF, Unnikrisnan GU, Hussein AI. Bone mechanical properties in healthy and diseased states. Annu Rev Biomed Eng. 2018;20(1):119–43.
12. Jepsen KJ, Silva MJ, Vashishth D, Guo XE, van der Meulen MC. Establishing biomechanical mechanisms in mouse models: practical guidelines for systematically evaluating phenotypic changes in the diaphyses of long bones. J Bone Miner Res. 2015;30(6):951–66.
13. Avallone EA, Baumeister T III, editors. Marks' standard handbook for mechanical engineers. 10th ed. New York: McGraw-Hill; 1996.
14. Roylance D. Stress-strain curves [internet]. MIT OpenCourseWare. http://ocw.mit.edu. Cambridge, MA: Massachusetts Institute of Technology; 2001. Available from: http://web.mit.edu/course/3/3.11/www/modules/ss.pdf. Accessed 18 Sept 2018.
15. Manjubala I, Liu Y, Epari DR, Roschger P, Schell H, Fratzl P, et al. Spatial and temporal variations of mechanical properties and mineral content of the external callus during bone healing. Bone. 2009;45(2):185–92.
16. Leong PL, Morgan EF. Measurement of fracture callus material properties via nanoindentation. Acta Biomater. 2008;4(5):1569–75.
17. Leong PL, Morgan EF. Correlations between indentation modulus and mineral density in bone-fracture calluses. Integr Comp Biol. 2009;49(1):59–68.
18. Viguet-Carrin S, Garnero P, Delmas PD. The role of collagen in bone strength. Osteoporos Int. 2006;17(3):319–36.
19. Garnero P. The contribution of collagen crosslinks to bone strength. Bonekey Rep. 2012;1:182. https://doi.org/10.1038/bonekey.2012.182.

20. Bartel DL, Davy DT, Keaveny TM. Orthopaedic mechanics and design in musculoskeletal systems. Pearson/Prentice Hall: Upper Saddle River; 2006.
21. Currey J. The many adaptations of bone. J Biomech. 2003;36(10):1487–95.
22. Thomopoulos S, Marquez JP, Weinberger B, Birman V, Genin GM. Collagen fiber orientation at the tendon to bone insertion and its influence on stress concentrations. J Biomech. 2006;39(10):1842–51.
23. Mfoumou E, Tripette J, Blostein M, Cloutier G. Time-dependent hardening of blood clots quantitatively measured in vivo with shear-wave ultrasound imaging in a rabbit model of venous thrombosis. Thromb Res. 2014;133(2):265–71.
24. Weisel JW. The mechanical properties of fibrin for basic scientists and clinicians. Biophys Chem. 2004;112(2–3):267–76.
25. Silva MJ, Brodt MD, Wopenka B, Thomopoulos S, Williams D, Wassen MHM, et al. Decreased collagen organization and content are associated with reduced strength of demineralized and intact bone in the SAMP6 mouse. J Bone Miner Res. 2006;21(1):78–88.
26. García-Rodríguez J, Martínez-Reina J. Elastic properties of woven bone: effect of mineral content and collagen fibrils orientation. Biomech Model Mechanobiol. 2017;16(1):159–72.
27. Herman BC, Cardoso L, Majeska RJ, Jepsen KJ, Schaffler MB. Activation of bone remodeling after fatigue: differential response to linear microcracks and diffuse damage. Bone. 2010;47(4):766–72.
28. Colopy SA, Benz-Dean J, Barrett JG, Sample SJ, Lu Y, Danova NA, et al. Response of the osteocyte syncytium adjacent to and distant from linear microcracks during adaptation to cyclic fatigue loading. Bone. 2004;35(4):881–91.
29. Özkaya N, Leger D, Goldsheyder D, Nordin M. Mechanical properties of biological tissues. In: Fundamentals of biomechanics: equilibrium, motion, and deformation. 4th ed. Cham: Springer International; 2016. p. 361–83.
30. White AA, Panjabi MM, Southwick WO. The four biomechanical stages of fracture repair. J Bone Joint Surg Am. 1977;59(2):188–92.
31. Silva MJ, Touhey DC. Bone formation after damaging in vivo fatigue loading results in recovery of whole-bone monotonic strength and increased fatigue life. J Orthop Res. 2007;25(2):252–61.
32. Currey JD. Stress concentrations in bone. J Cell Sci. 1962;s3–103(61):111–33.

2 Perren's Strain Theory and Fracture Healing

Sascha Halvachizadeh and Hans-Christoph Pape

Introduction

Fractured bone behaves differently in biological and mechanical environments. This fact influences the choice and method of treatment by the surgeon. Surgical procedures alter the biological environment, fixation of broken bone alters the mechanical environment; both directly contribute to the course of healing and are determined by the surgeon. The basic knowledge of biology and biomechanics of fracture healing are essential for all trauma surgeons since this knowledge defines the first treatment strategies (Table 2.1) [1].

This chapter serves as a summary for active clinicians rather than a pure scientific review. The primary goal of fracture fixation is to achieve prompt and if possible full function of the injured limb. The functional recovery does not only base on fracture healing but also its mechanics, biomechanics, and biology since these factors define a promising outcome for the patients. During fracture fixation, it is not always possible to achieve full mechanical and optimal biological environment. Strength and stiffness need to be sacrificed if the biological environment is aimed to be as optimal as possible. On the other hand, mechanical requirements may impair the biological environment. Osteosynthesis should not aim to permanently replace a broken bone; it should provide a temporary support allowing early functional rehabilitation with healing in a proper anatomical position [1–4]. Further, the type of implant influences the biomechanical environment: Steel with mechanical strength and ductility stands opposed to the electrochemical and biological inertness of titanium. With the treatment decision, the trauma surgeon determines a compromise between mechanical stability and the biological environment as well as the combination of technology and procedure best fit for the patient.

Table 2.1 Relationship of fracture morphology, stabilization, and fracture healing [1]

	Simple	Multifragmentary
	Small (<2 mm)	Large (>2 mm)
Relative stability	Bone resorption, healing delay, or nonunion	Secondary bone healing (callus)
Absolute stability	Primary bone healing, osteonal remodeling	Bone resorption, healing delay, or nonunion

Bone

The bone supports and protects soft structures while enabling locomotion and mechanical functioning of the limbs. The mechanical characteris-

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tics of the bone are a combination of stiffness (little deformation under load) and strength (tolerate high load without failure). The characteristics of the bone are more similar to glass than to rubber [1]: The strong material breaks under very small deformation. The repeated displacement of fractured elements inhibits the bridging of the fracture gap. Relative stabile fixed fracture shifts the balance of mechanical versus biological functions more toward the biological environment. That leads to a sequence of biological events, mainly the formation of soft callus followed by hard callus. That leads to a reduction of strain and the deformation of the less stiff repair tissue. This lower-strain environment promotes the formation of bridging callus which subsequently increases the mechanical stability of the fracture. Full function is restored after the fracture is solidly bridged. Bone remodeling restores the original bone structure [5, 6].

Fracture

A fracture is an unphysiologic discontinuation of the bony integrity. It is the result of repetitive or a single overload. A traumatic fracture is always combined with a soft tissue damage to certain degree. The rapid separation of fracture surfaces creates a void (cavitation) that also contributes to soft tissue damages.

Mechanical and Biomechanical Effects

The loss of continuity results in deformation, pain, and impaired supportive function of the bone. Stabilization may restore function and relieve pain. This may lead to pain-free mobility and reduce complications such as complex regional pain syndromes.

With the disruption of the bone, arteries and veins are damaged. That leads to a spontaneous release of biochemical factors helping to induce bone healing. The fresh fractured bone has enough active and effective factors that support and coordinate fracture healing [7]. Biologically, surgical fracture stabilization should be nothing more than a guidance and a support in this healing process.

Blood Supply The pure mechanical process of a fracture triggers biological reactions (callus formation and bone resorption). Equally to other tissue, the healing and remodeling depends on an intact blood supply. Table 2.2 summarizes factors that influence blood supply at the fracture site [1, 8–11].

Necrotic bone induces internal (Haversian) remodeling, allowing the replacement of dead osteocytes with temporary loss of strength (porosis). Large plate-bone contact areas show this effect just beneath the plates. A higher periosteal blood supply (e.g., with the help of a limited-contact dynamic compression plate [LC-DCP]), may reduce the amount of avascular bone. Injury to the bone reduces cortical circulation by nearly 50% [11] that attributes to a vasoconstriction in the periosteal but also the medullary vessels [12]. The healing process, however, increases blood flow in the adjacent soft tissue that nourishes the callus formation [13, 14]. This supports the statement

Table 2.2 Overview of factors that influence blood supply in a fractured bone [1, 9–11]

Mechanism of injury	Fracture type and its associated soft tissue injury depend on the energy and the direction of the injury
Initial patient management	Motion during rescue leads to additional damage
Patient resuscitation	Fluid resuscitation and oxygen therapy may support healing Late correction of fracture displacements heightens the risk of additional soft tissue damages
Surgical approach	The additional damages through surgical exposure of the fracture should be minimized. Anatomical knowledge is one of the most essential tools of any surgeon [8]
Implant	Contact of the bone and implant may damage bone circulation and depend on the type of implant [9]
Consequence of trauma	Elevated intra-articular pressure and increased hydraulic pressure reduce blood supply and should be addressed early

that the approach to the fracture should damage and strip off as little soft tissue as possible. The essential angiogenesis with subsequent callus perfusion is elementary in the healing process of the fractured bone. This, however, depends on the method of treatment and the induced mechanical conditions (Table 2.3) [15–22].

Biology of Fracture Healing

Fracture healing is divided either into primary (direct) healing by internal remodeling or secondary (indirect) healing by callus formation.

Table 2.3 Factors influencing angiogenesis during treatment and healing of fractures

Flexible fixation leads to greater vascular response (larger volume of callus)
Large strain reduces blood supply (instability in great fracture gap) [15]
Surgical exposure alters fracture hematoma and soft tissue blood supply
Extent of reaming influences cortical perfusion [16–18]
Implant-bone contact reduces bone perfusion
Fragment manipulation, minimal invasive approaches, and external or internal fixators may reduce damages to the blood supply [19–22]

The biological process of osteonal bone remodeling is the basis of primary bone healing and occurs only with absolute stability. Relative stability, however, promotes secondary bone healing through flexible fixation methods. The biology behind secondary bone healing is similar to the process of embryological bone development including both intramembranous and endochondral bone formation [23]. This can be seen during the callus formation in the process of fracture healing of diaphyseal fractures (Table 2.4) [1, 24].

Biomechanics of Bone Healing

Methods of Fracture Stabilization

Surgeons interpret stability as the degree of displacement at the fracture site induced by load [25]. This degree defines the course of fracture healing. The stabilized fracture does not displace under physical load. The aim of fracture stabilization is to maintain the achieved reduction, restore stiffness at the fracture site, and minimize pain related to instability at the fracture site [26–28].

Table 2.4 The four stages of bone healing [1, 24]

Inflammation	Until 1 week after fracture Inflammatory exudation from damaged vessels Soft tissue injury and degranulation of platelets release inflammatory mediators Vasodilatation, hyperemia, PMN, macrophages Granulation tissue replaces gradually fracture hematoma Osteoclasts remove necrotic bone
Soft callus formation	2–3 weeks after fracture in adequately stabilized fracture Growing callus can be observed Progenitor cells are stimulated to become osteoblasts Intramembranous bone growth starts from both fracture sites Mesenchymal progenitor cells proliferate and migrate through the callus At the end: stability is adequate to prevent shortening not however angulation
Hard callus formation	3–4 months Starts as soon as soft callus has bridged the fracture gap Soft tissue within the gap undergoes endochondral ossification Callus is converted into calcified tissue (woven bone) At the point of lowest strain (periphery of fracture site), bone callus growth begins Hard callus formation starts peripherally and progressively moves toward the center of the fracture gap Endochondral ossification replaces soft tissue in the gap by woven bone joining original cortex
Remodeling	Months to years Begins after the fracture has solidly united with woven bone The balance of pressure and tensile forces rebuilt the bone to its original morphology

PMN polymorphonuclear neutrophils

Absolute stability shifts the balance toward neutral mechanical fixation preventing callus formation. Relative stability shifts the balance toward the biological environment that stimulates secondary bone healing. This however can only occur if the motion at the fracture site is not too extensive preventing callus formation and delaying healing [29].

Conservative Fracture Treatment

This type of treatment requires closed reduction with subsequent stabilization to maintain reduction and reduce fragment motion. Indirect bone healing will be stimulated with a nearly uninterrupted biological environment. Traction or external splinting may achieve stabilization in conservative fracture treatment. A curved cast produces a straight bone, and a straight cast produces a bent bone [1, 30]. The pressure of the surrounding tissue reduces movement of the fragments.

Relative Stability

Relative stability allows movement of the bone fragments when physiological load is applied. Rigidity of the fixation decreases displacement. If a fixation method is considered flexible, it allows controlled interfragmentary movement under physiological load [2, 24, 31].

External Fixator Usually, external fixators provide relative stability. The stiffness of fracture stabilization by external fixation depends on the type of implant (e.g., Schanz screws and bars), the geometrical arrangements of these elements related to one another and the bone (e.g., uniplanar, biplanar, circular), and the coupling of the implant to the bone (e.g., tensioned fine wires) [32]. Important factors that increase the stability of fixation include the following:

- Stiffness of the connecting rods.
- Short distance between the rod and the bone axis.
- Number, spacing and diameter of the Schanz screws/wires, and their tension.
- Unilateral external fixator combines axial, bending, and shearing displacements. The external fixator is the only system that allows the surgeon to modify flexibility by adjusting the implant without additional surgery (dynamization). External fixators provide quick and relative safe stability and are used when the biologic environment and the soft tissue situation allow little manipulation [33–35].

Intramedullary Nailing The classic intramedullary nail achieves stability against bending and shear forces perpendicular to its long axis; it however is not immune to torque and is unable to prevent axial shortening. With low torsional stiffness and the loose coupling of intramedullary nail and bone, intramedullary nailing was indicated to simple transverse or short oblique fractures, which cannot shorten and will interdigitate to prevent rotation [36, 37]. The development of locked intramedullary nails and cannulated nails overcame many of these restrictions. Locked nail has the ability to tolerate torsional forces and improved axial loading [38]. The diameter of the nail defines the degree of stability additional to its geometry and the number of interlocking screws as well as their spatial arrangements. The advantages of the locked intramedullary nail come to the price of nonlinear stiffness of the nail-bone construction. To promote insertion of interlocking screws, the locking holes are larger than the diameter of the screws, allowing movement at the coupling, even with little load. Further insertion of interlocking screws may decrease motion as well as the use of angular-stable locking systems (e.g., expert tibial nail) [39, 40].

Internal Fixators and Bridging Plates Multifragmented fractures that are stabilized with a plate in the manner of an external fixator provide elastic splinting. The dimensions of the implant as well as the number and the positions of the screws, the coupling between implants, and the implant and the bone define the stiffness of this fixation method. The indication for plating with the goal of relative stability includes multifragmentary fractures. If it is possible, plating with relative stability should be avoided for simple fractures since it increases the risk for delayed union or even nonunion [41].

Indirect Fracture Healing: Perren's Strain Theory

Acceleration of bone healing is achieved with the stimulation of the formation of callus with interfragmentary movement [42, 43]. The maturing callus stiffens and reduces interfragmentary motion with the possibility of hard bony callus formation. In the early stages of fracture healing, the fracture tolerates greater deformation or higher tissue strain. Later stages when the callus is calcified, the tissue does not tolerate these deformations or strain forces. Figure 2.1 reveals the dynamics of Perren's strain theory. Cell disruption occurs in small gap fractures, whereas in large gap fractures, the strain forces are distributed among cells with each cell experiencing less traction force [2].

Strain is defined as the deformation of material when a given force is applied. Since normal strain is the relative difference in length when a given load is applied, it has no dimension as is expressed by percentage. Before it fractures, intact bone has a normal strain tolerance of 2% before it fractures, compared to the strain tolerance of 100% in granulation tissue [29]. When the movement of fracture ends is too great, the local strain rises over tolerable limit of forming woven bone leading to impaired bridging by hard callus [44]. This leads to an increased volume of the soft callus resulting in a decreased local tissue strain to a level that allows bony bridging. This adaptive mechanism of increasing soft callus volume is impaired in considerable narrow fracture gaps with its movements occurring mostly at the gap leading to high-strain environment. The tolerance decreases after overloading the fracture with too much interfragmentary movement in the healing process [45]. Strain and fluid pressure have an inhomogeneous distribution within the callus. The applied load regulates the callus formation with biophysical stimuli that are sensed by the cells. Different signals are produced with these biophysical stimuli that alter extracellular matrix and tissue properties. After ossification

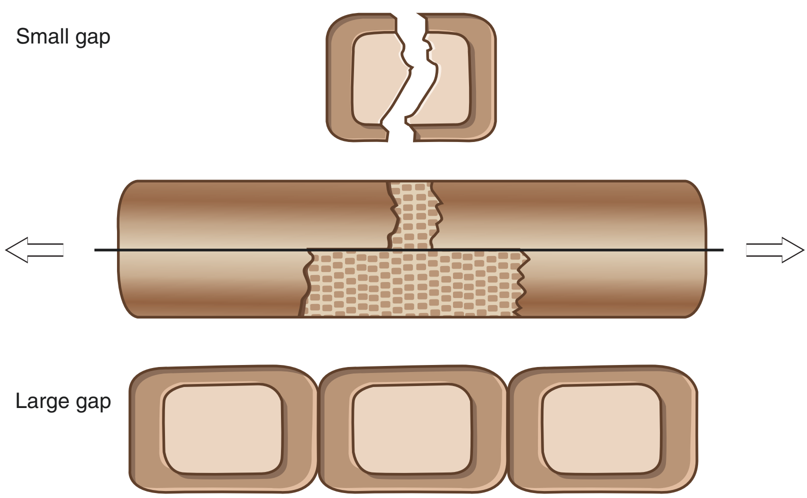

Fig. 2.1 Perren's strain theory. Cell disruption occurs in small gap fractures, whereas in large gap fractures, the strain forces are distributed among cells, resulting in each cell experiencing less traction force. (Adapted from AO principles of fracture management [2], with permission from AO Foundation, Duebendorf, Switzerland)

of the callus, these signals reach a steady state and the original cortex regenerates. Excessive interfragmentary strain as well as too wide fracture gaps impair bridging by hard callus leading to the development of hypertrophic nonunions [46]. On the other hand, some mechanical stimulations are needed to form callus. This is impaired in low-strain environments after either too stiff fixation or too wide fracture gaps [44] resulting in delayed healing or nonunions.

Absolute Stability

The only effective method to abolish fracture movement is interfragmentary compression leading to a no-strain environment. This leads to direct bone healing. The fracture heals without visible callus formation. Osteonal remodeling is the consequence of this fixation method (compressive preload and friction).

Compressive Preload The compressive preload prevents displacement of the fracture fragments leading to absolute stability if the compression is greater than traction produced by function. The static compression does not produce necrosis, neither in lag screws nor in compression plates, even in overloaded bone [47].

Friction Friction counteracts shear forces that act tangentially and thus avoids sliding displacements. The amount of resistance to shear displacements depends on the compression-induced friction and the geometry of the surfaces in contact. Normal smooth bone surfaces produce less than 40% friction [47]. Rough surfaces allow firm fixation with additional counteracting displacements due to shear forces.

Lag Screws and Plates The lag screw stabilizes fractures by compression after the approximation between the thread and the head of the screw. These forces exceed the time direct bone healing requires. Lag screws, however, provide small lever arm to resist functional loading. When viewed from the center of the screw, the area of compression is too small to withstand bending and shearing. This can be overcome with protection plates (neutralization plates) that protect the lag screws from these forces. Thus the plate protects, has ability to compress, and can be used as tension band, bridging of buttress. Simple transverse or short oblique fractures can be treated by a plate that is applied to one side of a fracture and then tensioned (excentric placement of screws). However, this method on a straight plate on a straight bone produces compression underneath the plate with slight tension on the opposite cortex leading to an instable situation. This problem can be overcome by overbending the plate to produce a small gap between the plate and the bone at the level of the fracture. If the plate is placed at the tension side of the bone, it acts as tension band and converts tension into compression at the far cortex producing absolute stability. The buttress is a construct that resists axial load by applying force at 90° to the axis of potential deformity. It is often combined with lag screws.

Direct Fracture Healing: Biomechanics

In the diaphysis, absolute stability is achieved by interfragmentary compression. Early functional treatment is possible within a few days of surgery. Radiological, only minor changes can be observed with minimal or no callus formation [48]. The gradual disappearance of the fracture sign shows progredient fracture healing, whereas a widening of this line may indicate insufficient stability. In the first days after surgery, minimal activity can be observed near the fracture site. The hematoma is resorbed/transformed into repair tissue. After few weeks Haversian systems remodel the bone [49] with simultaneous lamellar filling of fragments. In the following weeks, the osteons reach the fracture and cross it as soon as there is contact [50]. The newly formed osteons crossing the gap provide a micro-bridging or interdigitation.

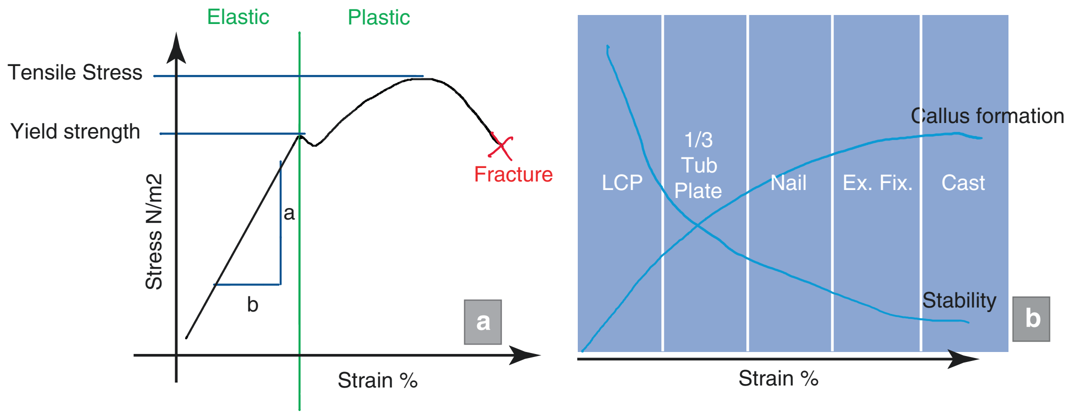

Fig. 2.2 (**a**) Stress-strain diagram with different stages of deformation. Young's modulus is the ability of material to withstand deformation (a/b). Tensile stress is the maximal tension material can withstand. At the *yield* strength, elastic (reversible) deformation ends and plastic (irreversible) deformation starts. Here first fibers start to beak. (**b**) Different therapeutic interventions influence callus formation as well as stability as the function of strain. The most stable (rigid) fixations lead to minimal callus formation. The less flexible the fixation, the more strain is observed that leads to more callus formation. (N = Newton, m = meter, LCP = lateral compression plate, Tub plate = tubular plate, Ex. Fix. = External fixator)

Summary

Each tissue has elastic and rigid properties. The biomechanical function of tissues depends on the proportion of these properties. Within a tissue, elastic and rigid properties define the amount of stress this tissue can withstand. The bone has the ability to withstand a certain amount of stress as well as strain with reversible deformation (elastic deformation). However, when a certain point is reached (yield strength), single cells or cell compounds start to break and the deformation is irreversible (plastic deformation). If the tissue suffers more strain, after the maximum of stress it can withstand (tensile stress), the tissue will fracture.

Depending on the treatment strategy, as well as the fracture pattern and properties, the degree of rigidity and elasticity within the fracture gap can be defined by the fixation method. For example, the lateral compression plate minimizes due to absolute stability within a fracture gap strain leading to minimal callus formation. Contrary, casting allows more strain leading to more callus formation due to less stable fracture fixation (Fig. 2.2). The treatment strategy depends on the fracture properties as well as the patients' concomitant injuries. Each treatment should be evaluated and assessed individually.

References

1. Ito K, Perren SM. Biology and biomechanics in fracture management. In: Rüedi TP, Buckley RE, Moran CG, editors. Arbeitsgemeinschaft für Osteosynthesefragen. AO principles of fracture management. 2nd ed. Stuttgart/New York: Thieme; 2007. p. 9–31.
2. Stephan Perren. AO principles of fracture management. Stuttgart: Thieme; 2000.
3. Giannoudis P, Tzioupis C, Almalki T, Buckley R. Fracture healing in osteoporotic fractures: is it really different? A basic science perspective. Injury. 2007;38(Suppl 1):S90–9.
4. Jagodzinski M, Krettek C. Effect of mechanical stability on fracture healing–an update. Injury. 2007;38(Suppl 1):S3–10.
5. Hadjidakis DJ, Androulakis II. Bone remodeling. Ann N Y Acad Sci. 2006;1092:385–96.
6. Wang W, Yeung KW. Bone grafts and biomaterials substitutes for bone defect repair: a review. Bioact Mater. 2017;2(4):224–47.
7. Dimitriou R, Tsiridis E, Giannoudis PV. Current concepts of molecular aspects of bone healing. Injury. 2005;36(12):1392–404.
8. Hoppenfeld S, DeBoer P, Buckley R. Surgical exposures in orthopaedics: the anatomic approach. 4th ed. Philadelphia: Wolters Kluwer Health; 2012.

9. Gautier E, Cordey J, Mathys R, Rahn BA, Perren SM. Porosity and remodelling of plated bone after internal fixation: result of stress shielding or vascular damage? In: Ducheyne P, van der Perre G, Aubert AE, editors. Biomaterials and biomechanics 1983:proceedings of the fourth European conference on biomaterials. Amsterdam: Elsevier; 1984. p. 195–200.
10. Gautier E, Perren S, Cordey J. Effect of plate position relative to bending direction on the rigidity of a plate osteosynthesis. A theoretical analysis. Injury. 2000;31(Suppl 3):C14–20.
11. Grundnes O, Reikerås O. Blood flow and mechanical properties of healing bone: femoral osteotomies studied in rats. Acta Orthop Scand. 1992;63(5):487–91.
12. Kelly PJ, Montgomery RJ, Bronk JT. Reaction of the circulatory system to injury and regeneration. Clin Orthop Relat Res. 1990;254:275–88.
13. Brookes M, Revell WJ. Blood supply of bone: scientific aspects. London: Springer; 1998.
14. Rhinelander FW. Tibial blood supply in relation to fracture healing. Clin Orthop Relat Res. 1974;105:34-81.
15. Eckert-Hübner K, Claes LJ. Callus tissue differentiation and vascularization under different conditions. (Abstract from Sixth Meeting of the International Society of Fracture Repair; 23–26 Sep 1998, Strasbourg, France). J Orthop Trauma. 1999;13(4):282–3.
16. Kessler S, Hallfeldt K, Perren S, Schweiberer L. The effects of reaming and intramedullary nailing on fracture healing. Clin Orthop Relat Res. 1986;212:18–25.
17. Pfeifer R, Sellei R, Pape HC. The biology of intramedullary reaming. Injury. 2010;41(Suppl 2):S4–8.
18. Klein M, Rahn B, Frigg R, Kessler S, Perren SJ. Reaming versus non-reaming in medullary nailing: interference with cortical circulation of the canine tibia. Arch Orthop Trauma Surg. 1990;109(6):314–6.
19. Claes L, Heitemeyer U, Krischak G, Braun H, Hierholzer GJ. Fixation technique influences osteogenesis of comminuted fractures. Clin Orthop Relat Res. 1999;365:221–9.
20. Perren SM, Buchanan JS. Basic concepts relevant to the design and development of the Point Contact Fixator (PC-Fix). Injury. 1995;26(Suppl 2):1–4.
21. Tepic S, Perren SJ. The biomechanics of the PC-Fix internal fixator. Injury. 1995;26(Suppl 2):5–10.
22. Farouk O, Krettek C, Miclau T, Schandelmaier P, Guy P, Tscherne HJ. Minimally invasive plate osteosynthesis: does percutaneous plating disrupt femoral blood supply less than the traditional technique? J Orthop Trauma. 1999;13(6):401–6.
23. McKibbin B. The biology of fracture healing in long bones. J Bone Joint Surg Br. 1978;60-B(2):150–62.
24. Allgöwer M, Perren SM, Rüedi T. Biophysikalische Aspekte der normalen und der heilenden Knochencorticalis. Langenbecks Arch für Chirurgie. 1970;328(1–2):109–14.
25. Perren S. Physical and biological aspects of fracture healing with special reference to internal fixation. Clin Orthop Relat Res. 1979;138:175–96.
26. Perren SM, Huggler A, Russenberger M, Straumann F, Müller ME, Allgöwer M. A method of measuring the change in compression applied to living cortical bone. Acta Orthop Scand Suppl. 1969;125:7–16.
27. Perren S, Russenberger M, Steinemann S, Müller M, Allgöwer M. A dynamic compression plate. Acta Orthop Scand Suppl. 1969;125:31–41.
28. Perren SM. Optimizing the degree of fixation stability based on the strain theory. Orthopade. 2010;39(2):132–8. [Article in French].
29. Perren SM. Evolution of the internal fixation of long bone fractures: the scientific basis of biological internal fixation: choosing a new balance between stability and biology. Bone Joint J. 2002;84(8):1093–110.
30. Browner BD, Jupiter JB, Krettek C, Anderson PA. Skeletal trauma e-book: basic science management and reconstructions. 5th ed. Philadelphia: Elsevier Saunders; 2015.
31. Rüedi TP, Buckley RE, Moran CG; Arbeitsgemeinschaft für Osteosynthesefragen. AO principles of fracture management, 2nd ed. Vol 2 Specific fractures. Stuttgart/New York: Thieme; 2007.
32. Fleming B, Paley D, Kristiansen T, Pope MJ. A biomechanical analysis of the Ilizarov external fixator. Clin Orthop Relat Res. 1989;241:95–105.
33. Hildebrand F, Giannoudis P, Kretteck C, Pape HC. Damage control: extremities. Injury. 2004;35 (7):678–89.
34. Pape H-C, Giannoudis P, Krettek C. The timing of fracture treatment in polytrauma patients: relevance of damage control orthopedic surgery. Am J Surg. 2002;183(6):622–9.
35. Pape H-C, Tornetta P, Tarkin I, Tzioupis C, Sabeson V, Olson SA. Timing of fracture fixation in multitrauma patients: the role of early total care and damage control surgery. J Am Acad Orthop Surg. 2009;17(9): 541–9.
36. Hansen ST, Winquist RA. Closed intramedullary nailing of the femur. Küntscher technique with reaming. Clin Orthop Relat Res. 1979;138:56–61.
37. Bick EM. The intramedullary nailing of fractures by G. Küntscher. Translation of article in Archiv für Klinische Chirurgie, 200:443, 1940. Clin Orthop Relat Res. 1968;60:5–12.
38. Schandelmaier P, Krettek C, Tscherne HJ. Biomechanical study of nine different tibia locking nails. J Orthop Trauma. 1996;10(1):37–44.
39. Johnson KD, Tencer AF, Blumenthal S, August A, Johnston DJ. Biomechanical performance of locked intramedullary nail systems in comminuted femoral shaft fractures. Clin Orthop Relat Res. 1986;206:151–61.
40. Bong MR, Kummer FJ, Koval KJ, Egol KA. Intramedullary nailing of the lower extremity: biomechanics and biology. J Am Acad Orthop Res. 2007;15(2):97–106.
41. Piątkowski K, Piekarczyk P, Kwiatkowski K, Przybycień M, Chwedczuk BJ. Comparison of different locking plate fixation methods in distal tibia fractures. Int Orthop. 2015;39(11):2245–51.

42. Claes L, Augat P, Suger G, Wilke HJ. Influence of size and stability of the osteotomy gap on the success of fracture healing. J Orthop Res. 1997;15(4):577–84.
43. Claes LJ. Dynamisierung der Osteosynthese: Zietpunkt und Methoden. Der Unfallchirurg. Ausgabe 1/2018. Dynamisierung der Ostosynthese. 2018;121(1):3–9.
44. Perren SM, Cordey J. The concept of interfragmentary strain. In: Uhthoff HK, Stahl E, editors. Current concepts of internal fixation of fractures. Berlin: Springer; 1980. p. 63–77.
45. Goodship A, Kenwright J. The influence of induced micromovement upon the healing of experimental tibial fractures. Bone Joint J. 1985;67(4):650–5.
46. Schenk R, Müller J, Willenegger HJ. Experimental histological contribution to the development and treatment of pseudarthrosis. Hefte Unfallheilkd. 1968;94:15–24. [Article in German].
47. Perren SM, Huggler A, Russenberger M, Allgöwer M, Mathys R, Schenk R, et al. The reaction of cortical bone to compression. Acta Orthop Scand Suppl. 1969;125:19–29.
48. van Frank Haasnoot E, Münch TW, Matter P, Perren SM. Radiological sequences of healing in internal plates and splints of different contact surface to bone. (DCP, LC-DCP and PC-Fix). Injury. 1995;26(Suppl 2):28–36.
49. Schenk R, Willenegger HJ. Zum histologischen Bild der sogenannten Primärheilung der Knochenkompakta nach experimentellen Osteotomien am Hund. Experientia. 1963;19(11):593–5.
50. Rahn BA, Gallinaro P, Baltensperger A, Perren SM. Primary bone healing: an experimental study in the rabbit. J Bone Joint Surg Am. 1971;53(4):783–6.

3 Case Studies in Fracture Healing and Nonunions

Joseph Borrelli Jr. and Brent L. Norris

Introduction

The process of fracture healing in general consists of three interdependent phases: inflammation, repair, and remodeling. Normal fracture healing is initiated by the local hematoma that forms when a fracture occurs followed by an inflammatory response that occurs in response to the fracture and associated soft tissue injury. These initial events, hematoma formation and inflammation, have been shown to direct the downstream processes of the fracture repair and remodeling phases. The signaling cascades initiated during this initial inflammatory phase play a critical role in triggering bone regeneration and ultimately fracture healing. This local inflammation is influenced by both the acute systemic inflammatory response to injury and any chronic inflammatory states, commonly seen in certain acute and chronic conditions (i.e., polytrauma, sepsis, diabetes, rheumatoid arthritis, obesity, etc.). The inflammatory phase has been recognized as a prerequisite for successful bone healing [1, 2]. Factors that affect local inflammation include the surrounding soft tissue injury, the fracture hematoma, and the biomechanical stability of the fracture following initial treatment. Additionally, medical interventions, including the use of anti-inflammatory drugs, the administration of corticosteroids, smoking, and the use of alcohol by the patient, have also been shown to affect the local and systemic inflammation and ultimately fracture healing. The inflammatory stage serves primarily to prepare the site for the upcoming bone healing process by attracting large number of cells to the area. This enhanced chemotaxis is accomplished by the liberation of numerous inflammatory mediators. Polymorphonuclear leukocytes (PMNs), lymphocytes, blood monocytes, and macrophages are readily attracted to the site and attract additional inflammatory cells as well as mesenchymal cells, which ultimately leads to enhanced angiogenesis and the production of extracellular matrix [3].

Historically, fracture healing has been described as occurring either via so-called primary or direct fracture healing and/or by *secondary or indirect fracture healing*. Primary bone healing involves a direct attempt by the components of cortical bone to unite directly with the opposing cortical bone to reestablish the mechanical integrity of the bone. This process is thought to occur *only* when absolute stability of the reduced fracture fragments has been established by rigid internal fixation. Primary bone healing is allowed to proceed as the accurate reduction and

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stable fixation results in a substantial decrease in the strain at the fracture site. In general, absolute stability between the fracture fragments is obtained either with the placement of inter-fragmentary lag screws (with a neutralization plate) or with the use of dynamic compression plates used to create inter-fragmentary compression directly. Bone production by osteoblasts fills the microscopic gaps between the fracture fragments in the same manner that the Howship lacunae are filled after the action of "cutting cones." This type of bone healing occurs less frequently than secondary or indirect bone healing [3].

Secondary or indirect fracture healing involves an indirect method of fracture healing that employs the surrounding external soft tissues as well as the local periosteum to unite opposing fracture fragments. This type of fracture healing occurs in the absence of absolute stability between the major fracture fragments. Secondary bone healing includes the development of a fracture callus (an intermediate step), which is primarily made up of cartilage. During the healing process, this cartilage is replaced by woven and then lamellar bone. This intermediate step, which involves the formation of callus, is absent, for the most part, in primary fracture healing. Secondary bone healing generally proceeds in four stages: the hematoma phase, which serves to activate the coagulation cascade, change the local environment, and attract inflammatory cells; the granulation stage where the healing fracture is supported by active osteoprogenitor cell proliferation, angiogenesis, and abundant extracellular matrix production; the stage of callus formation, soft and hard, which contains—depending upon the mechanical environment—different types of differentiating mesenchymal stem cells (MSCs); and the remodeling phase (a rather long process that may take years to complete) where there is resorption of the remaining callus and restoration of the normal internal boney architecture without scar formation. During secondary bone healing the most important response to the fracture takes place within the periosteum. Here, both committed osteoprogenitor cells and uncommitted undifferentiated mesenchymal cells contribute to fracture healing by recapitulation of embryonic intramembranous ossification and endochondral bone formation. The response from the periosteum is a fundamental reaction to a fracture; this response is enhanced by micromotion between the fracture fragments and is inhibited by rigid fixation. Secondary bone healing is considered to be rapid and capable of bridging gaps between the fracture fragments as large as half the diameter of the local bone. In general, relative fracture stability occurs when fractures are treated nonoperatively with splints, casts, or fracture braces or when treated operatively with indirect reduction techniques utilizing bridge plate constructs, intramedullary nails, and external fixators.

Primary or Direct Fracture Healing

Case 1 (Fig. 3.1)

Primary or direct fracture healing does not occur commonly in nature. In fact, primary or direct bone healing was originally identified over a century ago with the introduction of rigid internal fixation of fractures [4]. Primary bone healing requires a near anatomical reduction of the fracture fragments that is without any significant gap between the fragments and stable or rigid fixation. This type of fracture healing is the goal of treatment when open reduction and internal fixation (ORIF) of intra-articular, peri-articular, and some diaphyseal fractures are treated with plates and screws. When near anatomical reduction and stable, if not rigid, fixation is achieved, direct bone healing can occur by direct remodeling of the lamellar bone, the Haversian canals, and osseous blood vessels. This type of healing generally takes a few months to a few years before complete healing (including the slow remodeling process) is achieved. Therefore, this healing process occurs considerably more slowly than secondary bone healing.

Both contact healing, where the fracture fragments are brought into direct contact with each

Fig. 3.1 Attempted posteroanterior (PA) (**a**) and lateral (**b**) forearm radiographs demonstrating displaced, comminuted fractures of the ulna and radius diaphysis. Intraoperative radiograph (**c**) of both bone forearm fractures after open reduction and internal fixation of the ulna has been performed. PA (**d**) and lateral (**e**) forearm radiographs, 14 months following ORIF with anatomic reduction, inter-fragmentary compression with the use of lag screws and dynamic compression plates. Each fracture has healed via primary bone healing without evidence of callus formation or implant failure

other, and gap healing, where the fracture fragments are brought into very close proximity to each other, occur during primary fracture healing. In both cases the healing process involves an attempt to directly reestablish a biomechanically competent lamellar bone structure across the fracture site. Cortical bone on one side of the fracture must unite with cortical bone on the other side of the fracture to reestablish mechanical continuity. If the gap between bone ends is less than 0.01 mm and inter-fragmentary strain is less than 2%, the fracture generally unites by so-called contact healing [5]. Under these conditions, cutting cones are formed at the ends of the osteons closest to the fracture site. The tips of the cutting cones consist of osteoclasts, which cross the fracture line, generating longitudinal cavities at a rate of 50–100 μm/day [6, 7]. These cavities are later filled by the bone produced by osteoblasts residing behind the osteoclasts and lining the sides of the cutting cone (Fig. 3.2a) [8]. This results in the simultaneous generation of a bony union and the restoration of Haversian systems formed in an axial direction. The reestablished Haversian systems allow for penetration of blood vessels carrying additional osteoblastic precursors. The bridging osteons later mature by direct remodeling into lamellar bone, resulting in fracture healing without the formation of periosteal callus or scar.

Gap healing, often a component of primary bone healing, differs from contact healing in that bony union and Haversian system remodeling occurs simultaneously. Gap healing occurs in the setting of stable, if not rigid, conditions with a near anatomical reduction of the fracture fragments. The gaps, however, must be less than 800 μm to 1 mm for gap healing to occur [5]. In gap healing, gaps between the fracture fragments are first filled with lamellar bone oriented perpendicular to the long axis of the bone and subsequently require secondary osteonal reconstruction, unlike during the process of contact healing (Fig. 3.2b) [8]. This preliminary bone

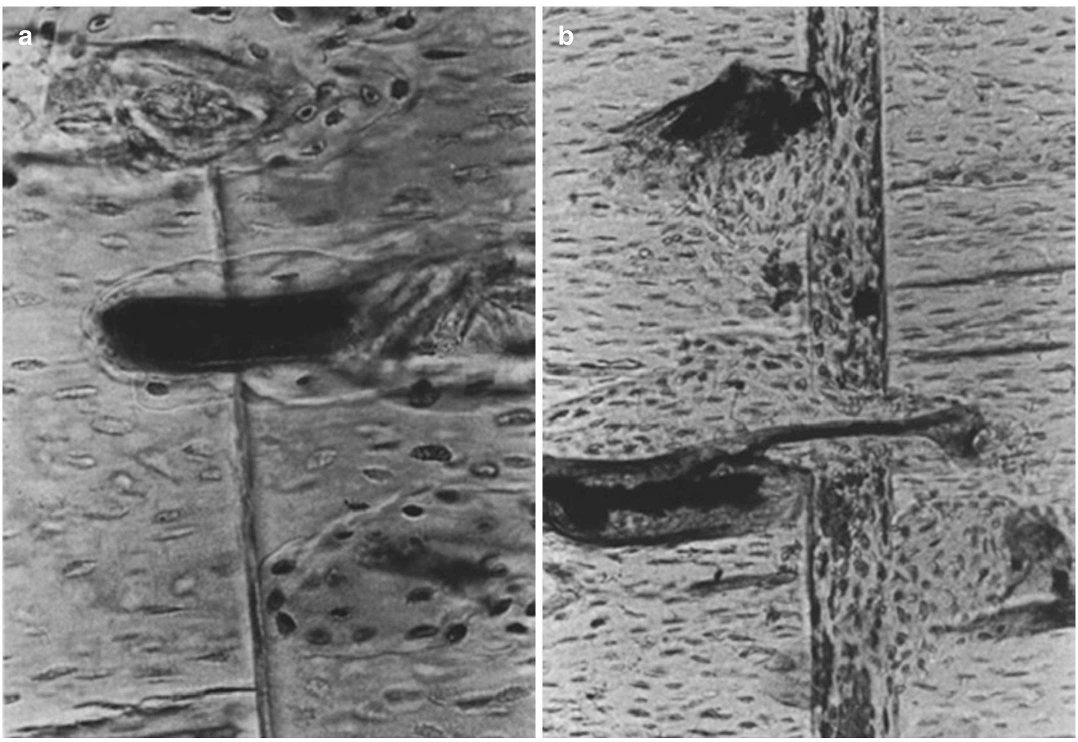

Fig. 3.2 Historic photomicrographs of contact healing and gap healing from a classic experimental study in rabbits. (**a**) Photomicrograph of contact healing demonstrates cutting cones lead by osteoclasts and trailed by osteoblasts proceeding across bone fragments brought directly into contact with each other following an osteotomy. (**b**) Photomicrograph of gap healing demonstrating the woven bone initially developed between the bone fragments and then subsequently remodeled into lamellar bone as healing progresses. (From Rahn et al. [8], with permission)

structure is then gradually replaced by longitudinal revascularized osteons carrying osteoprogenitor cells, which differentiate into osteoblasts and produce lamellar bone. The lamellar bone produced within these tiny gaps is mechanically weak. This initial process of filling in the gaps takes approximately 3–8 weeks, after which secondary remodeling, resembling contact healing, takes place [6]. This phase is necessary in order to fully restore the anatomical and biomechanical properties of the bone.

In both bone forearm fracture case presented above (section "Case 1"), each of the fracture fragments was anatomically reduced, and absolute stability was achieved with inter-fragmentary lag screws and dynamic compression plates. Fracture healing, which started shortly after the fracture occurred, proceeded via primary fracture healing after ORIF. Although not visible on plain radiographs, this primary bone healing proceeded with both contact and gap healing. To reiterate, contact healing requires that the strain between the fracture fragments not exceed 2% and gap healing will only proceed if the fracture "gap" is <1 mm. Achieving an anatomical reduction of each fracture and obtaining fixation with appropriate-sized lag screws and a limited contact-dynamic compression plate (LC-DCP) met the biomechanical conditions necessary for primary bone healing of this complex fracture.

Failure of Fracture Healing

Case 2 (Fig. 3.3)

Fracture healing is a complicated sequence of events involving many factors and therefore is influenced by both patient and fracture environmental factors. Patient factors include the age of the patient, presence of comorbidities, certain medications/drugs, smoking and alcohol, and, of course, the patient's genetics. Experimental animal studies have shown that bone healing potential declines with age, and this has been confirmed by several clinical studies that have shown that age is a negative predictor for fracture healing in certain fractures [9–11]. Comorbidities including malnutrition and metabolic deficiencies have also been identified as major risk factors for fracture nonunion. Deficiencies in calcium, phosphorus, vitamins C and D, albumin, and protein have all been found to negatively affect bone healing [12, 13].

Certain medications have been shown to have a direct negative effect on fracture healing. These necessary medications include antineoplastic drugs as well as widely used corticosteroids, which are known to encourage osteoblast apoptosis and osteocyte apoptosis and inhibit osteoblast genesis [14]. Additionally, bisphosphonates, which are widely used for the treatment of osteoporosis, most commonly in the older population, have also been shown to alter fracture healing. These drugs generally work by inhibiting bone resorption by mitigating the effects of osteoclasts. Some investigators have suggested that bisphosphonates might be candidates to actually upmodulate bone healing [15, 16]. Still other investigators as well as many clinicians have raised concerns regarding the effects bisphosphonates on the role of osteoclasts in the process of bone homeostasis and bone remodeling. Atypical femur fractures have been associated with the prolonged use of bisphosphonates. Their recognition and treatment have been well outlined since they were first recognized approximately 10–15 years ago, not long after the use of bisphosphonates was introduced to the population as a means to treat osteoporosis [16–20]. Atypical femur fractures have been found to have a rather consistent radiographic and fracture pattern and have been associated with an elevated risk of delayed union and nonunions. The causes of these complications are felt to be multifactorial and include alteration in bone healing as either a direct result of the bisphosphonates on the healing process, the presence of osteoporosis in a generally older population, and the difficulty of obtaining and maintaining a stable anatomic reduction of the fracture during ORIF and during the healing process.

The patient presented in section "Case 2" developed an impending atypical femur fracture likely as a result of prolonged use of bisphosphonates. Apparently, this impending fracture went unrecognized until the displaced fracture occurred. Unfortunately, the intramedullary nail was performed with the proximal fragment in residual varus and with displacement at the fracture site. The varus alignment of the proximal fracture fragment and perhaps the residual effects of the bisphosphonates contributed to the development of this atrophic nonunion. The second operative procedure included removal of the intramedullary nail, anatomical reduction of the fracture and proximal femur, and plating with inter-fragmentary compression. Her fracture then went on to heal uneventfully, presumably by primary fracture healing means, and she returned to her usual activities of daily living.

There are also fracture-dependent factors that influence fracture healing. These factors include fracture personality, location, surrounding soft tissue damage, and of course the biomechanical features of the fixation methods, techniques, and final construct.

To successfully stimulate fracture healing following the development of a nonunion often depends upon the type of nonunion and the rea-

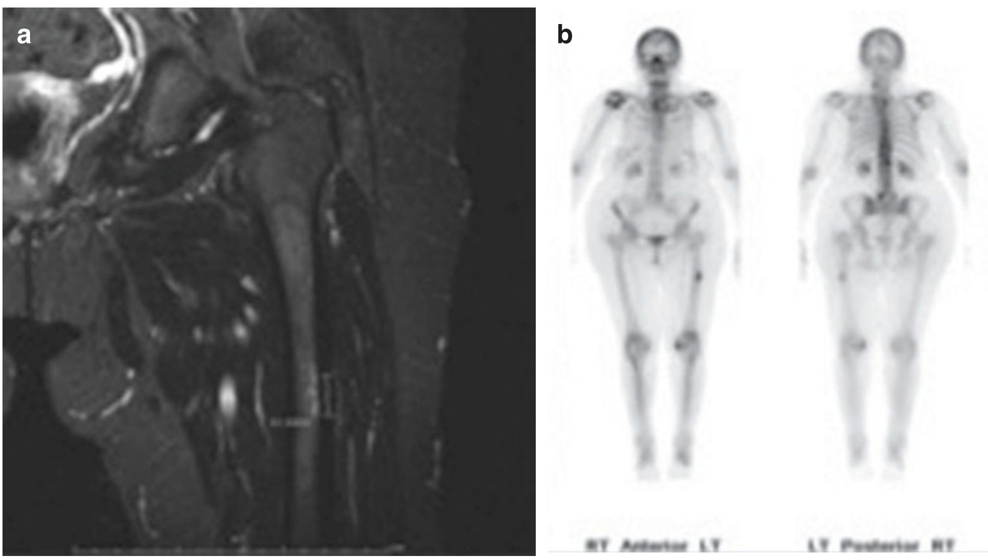

Fig. 3.3 (**a**) T2-weighted magnetic resonance image (MRI) of the left femur in a woman complaining of left thigh pain and who has been on anti-resorptive therapy for her osteoporosis for several years. The MRI indicates the presence of intramedullary edema in the subtrochanteric area. (**b**) Whole-body technetium-99 (99Tc) scan demonstrating marked increased uptake of the 99Tc tracer along the lateral cortex of the subtrochanteric area of the left femur. Anteroposterior (AP) (**c**) and lateral (**d**) radiographs of an atypical femur fracture that occurred in the area of increased 99Tc radioisotope on the bone scan and edema on the MRI. The fracture is transverse laterally and proceeds obliquely as it extends proximally through the medial cortex. Thickening of the lateral cortex can also be seen, consistent with an atypical femur fracture associated with long term anti-resorptive therapy. AP (**e**) and lateral (**f**) radiographs after intramedullary nailing of the fracture seen in (**c**) and (**d**). The fracture is poorly reduced in both the coronal and sagittal planes, and the proximal fragment remains in varus resulting in increased strain at the fracture site. AP (**g**) and lateral (**h**) radiographs after removal of the intramedullary nail and open reduction and internal fixation with interfragmentary compression plating. The fracture has healed by primary bone healing, with ongoing remodeling

Fig. 3.3 (continued)

son why the fracture failed to heal initially. In general, there are several potential applications that can be used to improve fracture and nonunion healing. In addition to improving the biomechanical environment of the nonunion site, additional treatments include the application of osteogenic materials to the fracture/nonunion site, such as autologous bone, bone morphogenetic proteins (BMPs), allograft bone, fibroblast growth factors (FGF), vascular endothelial growth factor (VEGF), platelet-derived growth factors (PDGF), and others. Systemic enhancement of the host has also been shown to improve fracture/nonunion healing in certain circumstances. These enhancements can include the use of parathyroid hormone, bisphosphonates, anti-sclerostin antibodies, anti-Dickkopf-related protein 1 (DKK1) antibodies, as well as others still under investigation. Biophysical stimulation has also been tried for many years to stimulate nonunion healing and to speed routine fracture healing, with mixed results [19, 20]. These modalities include electromagnetic field stimulation, low-intensity pulsed ultrasound stimulation, and extracorporeal stock wave therapy [21].

Secondary Fracture Healing

Case 3 (Fig. 3.4)

An 18-year-old male unrestrained back seat passenger in a motor vehicle collision (MVC). In this MVC the patient sustained an isolated, closed, spiral fracture of his left humerus. This fracture occurred at the junction of the middle and distal third of the humeral shaft (Fig. 3.4a, b). Following this MVC the patient was seen in the emergency department where he underwent closed reduction of the fracture and application of a coaptation splint of his left arm and humerus.

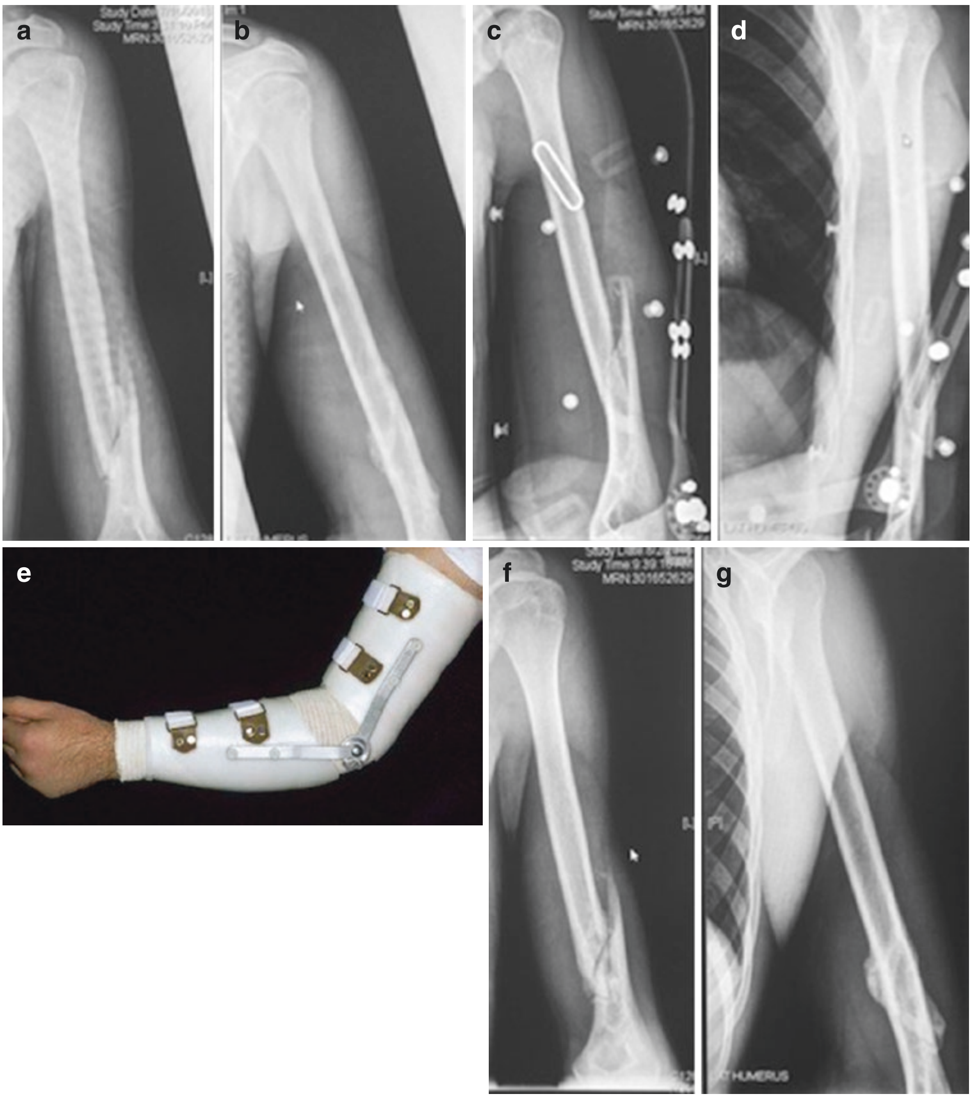

Fig. 3.4 Lateral (**a**) and oblique (**b**) radiographs of a closed, displaced spiral fracture of the left humerus in a young healthy male involved in a motor vehicle accident. Oblique (**c**) and lateral (**d**) radiographs of the spiral humerus fracture depicted in **a** and **b**, 10 days post-injury, the patient has now been placed in a prefabricated fracture brace, with an elbow hinge (**e**). Anteroposterior (AP) (**f**) and lateral (**g**) radiographs, 2 months post-injury. The fracture is beginning to heal via secondary bone healing with callus formation. AP (**h**) and lateral (**i**) radiographs, 3 months post-injury. The fracture is beginning to heal via secondary bone healing with callus formation. AP (**j**) and lateral (**k**) radiographs, 7 months post-injury. The fracture has healed and the patient has returned to his usual activities of daily living without restrictions as a college freshman

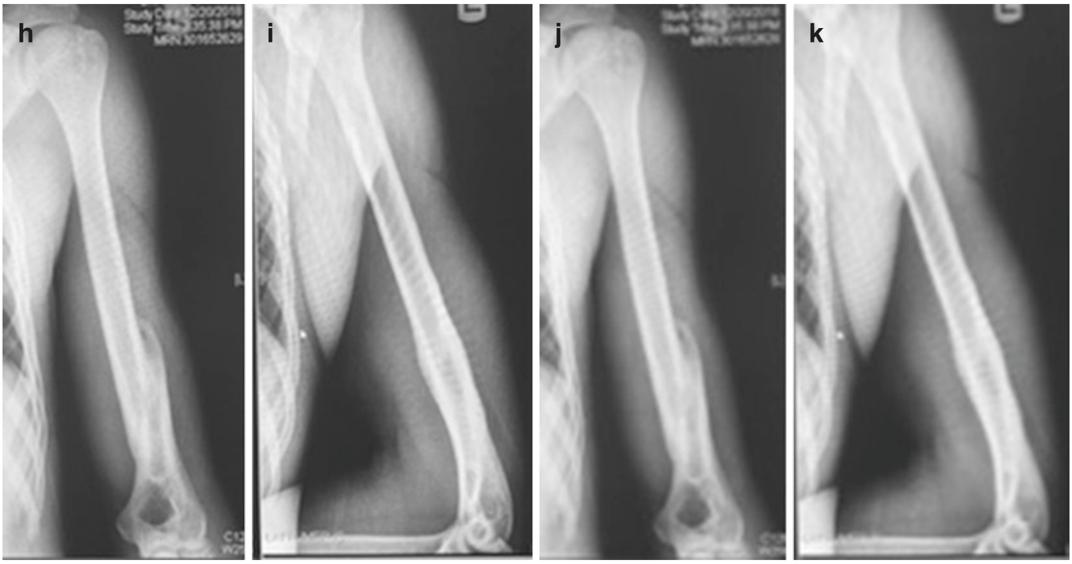

Fig. 3.4 (continued)

Several days after his injury, the patient came, along with his mother, for an office visit at which time his treatment options were discussed. The patient and his mother elected to try and treat this fracture nonoperatively. Approximately, 10 days after his injury, the patient was placed in a long arm Sarmiento-type fracture brace with a hinge at the level of his left elbow. At that time he was started on early active range motion of his left shoulder, elbow, wrist, and hand.

Secondary Fracture Healing

Case 4 (Fig. 3.5)

Case 5 (Fig. 3.6)

Indirect (secondary) fracture healing is the most common form of fracture healing and proceeds with both endochondral and intramembranous bone healing [22]. It does not require anatomical reduction of the fracture fragments nor rigid fixation and stabilization. Instead, indirect fracture healing is actually enhanced by micromotion at the fracture site. Of course, too much motion and/or load is known to result in delayed healing or even nonunion [23]. Indirect bone healing typically occurs in nonoperative fracture treatment, which generally requires the use of casts and braces, as well as certain fixation constructs that permit some motion at the fracture site. In most cases secondary fracture healing follows the use of intramedullary nails, external fixators, or bridge plating, each of which provides relative stability at the fracture site [24, 25].

As with all fractures, immediately following the fracture, a hematoma forms that consists of cells from both the peripheral and intramedullary blood, as well as from the liberated bone marrow. The resultant hematoma clots between and around the fracture ends, and within the medullary canal, ultimately forming a template for callus formation [26]. Pro-inflammatory molecules flood into fracture site and surrounding damaged soft tissues and are important for subsequent tissue regeneration and fracture healing. This acute inflammatory response peaks within the first 24 hours and lasts for approximately 7 days [27]. This initial proinflammatory response helps recruit inflammatory cells and promotes angiogenesis. Tumor necrosis factor alpha (TNF-α) and several interleukins, such as IL-1 and IL-6, are believed to be important in fracture healing by promoting the production of the primary cartilaginous callus and angiogenesis [28–31].

Fig. 3.5 Radiographs of a 23-year-old male involved in a motor vehicle crash in which he sustained a closed, comminuted left femoral shaft fracture (**a**, **b**). Anteroposterior (AP) (**c**, **d**) and lateral (**e**) radiographs, 6 weeks after closed intramedullary nailing of this closed, left femoral shaft fracture. AP (**f**) and lateral (**g**, **h**) radiographs 6 months after closed intramedullary nailing of this closed left femoral shaft fracture. The comminuted fracture is healing via secondary bone healing. AP (**i**) and lateral (**j**, **k**) radiographs 9 months after closed intramedullary nailing of this closed left femoral shaft fracture. Fracture has healed completely and the patient has returned to his usual activities of daily living

Fig. 3.6 Anteroposterior (AP) (**a**, **b**) radiographs of a closed, displaced comminuted fracture of the left femoral diaphysis sustained in an motor vehicle crash by a young, healthy female. AP (**c**, **d**) and lateral (**e**, **f**) intraoperative fluoroscopic images of a bridge plate construct used to treat this comminuted left femoral shaft fracture. AP (**g**, **h**) and lateral (**i**, **j**) radiographs of this comminuted left femur fracture, 3 months postoperatively with evidence of early healing of the major fracture fragments by secondary fracture healing. AP (**k**, **l**) and lateral (**m**, **n**) radiographs of this comminuted left femur fracture, 6 months postoperatively after a medial plate has been added to further stabilize the fracture and support further healing. (AP) (**o**, **p**) and lateral (**q**, **r**) radiographs of this comminuted femur fracture, 2 years post-operatively, demostrate complete fracture healing and restoration of length, alignment and rotation

Fig. 3.6 (continued)

In order for fractures to heal, specific MSCs have to be recruited, proliferate, and differentiate into osteogenic cells. Exactly where these cells come from is not fully understood, although most data now indicate that these MSCs are derived from surrounding soft tissues and bone marrow, as well as the systemic circulation from which they are likely recruited to the fracture site by BMPs [32]. In order for fracture healing to proceed, these MSCs must differentiate into chondrocytes, osteoblasts, or osteoclasts.

Although indirect fracture healing consists of both intramembranous and endochondral ossification, the formation of a cartilaginous callus that later undergoes mineralization, resorption, and then replacement with bone is the key feature of this process. Following the formation of the primary hematoma, a fibrin-rich granulation tissue forms. Within this tissue, endochondral formation occurs in between the fracture ends and external to periosteal sites. Although, initially the fracture is mechanically unstable, this cartilaginous tissue that forms the soft callus improves fracture stability, allowing additional healing to proceed [23].

At the same time, an intramembranous ossification response occurs subperiosteally at each end of the fracture, ultimately generating a hard callus. It is the final bridging of this central hard callus that ultimately provides the fracture with a semirigid structure [22].

In order for bone regeneration to progress, the primary soft cartilaginous callus needs to be resorbed and replaced by a hard bony callus. This step of fracture healing, to some extent, recapitulates embryological bone development with a combination of cellular proliferation and differ-

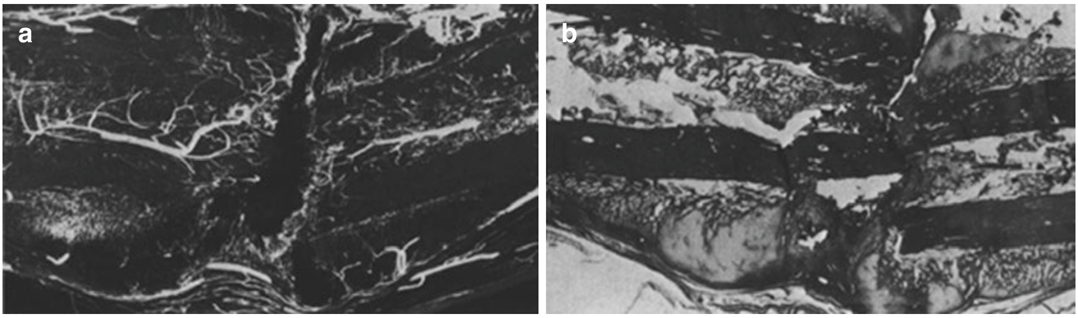

Fig. 3.7 Historic photomicrographs of secondary bone healing from classic texts**.** *(Left)* Photomicrograph of secondary bone healing taking place with a microangiogram demonstrating the abundant vascular supply needed for fracture healing. *(Right)* Photomicrograph of an osteotomy healing by secondary healing with abundant early callus formation on either side of the osteotomy as healing progresses. (From Rhinelander [37, 38] with permission)

entiation, increasing cellular volume and matrix deposition [33].

Although the hard callus is a rigid structure providing biomechanical stability, it does not fully restore the biomechanical properties of normal bone. In order to achieve this, the fracture-healing cascade initiates a second resorptive phase, this time to remodel the hard callus into a lamellar bone structure with a central medullary cavity [26]. A balance of hard callus resorption carries out the remodeling process by osteoclasts and lamellar bone deposition by osteoblasts. This remodeling may take years to be completed to achieve a fully regenerated bone structure [34]. For bone remodeling to be successful, an adequate blood supply and a gradual increase in mechanical stability is crucial [35]. This is clearly demonstrated in cases where neither is achieved, resulting in the development of an atrophic fibrous nonunion. However, in cases in which there is good vascularity but unstable fixation, the healing process progresses to form a large cartilaginous callus and results in the development of a hypertrophic nonunion or a pseudoarthrosis (Fig. 3.7) [36–38].

Hypertrophic Nonunion

Case 6: Failed Bone Healing (Fig. 3.8)

Hypertrophic nonunions are thought to develop due to insufficient (relatively unstable) mechanical environment. This relative instability prevents MSCs from differentiating into osteoblasts and generally leads to the formation of considerable soft callus in and around the fracture. Fracture healing has been recognized as a complex physiological process, and the successful treatment of many nonunion is just as complex. Recent advances have been made in the understanding of the molecular biology and genetics that directly influence fracture healing, including an improved understanding of the spatial and temporal actions of several of the key cell types, proteins, and the hundreds of gene expressions. Standardized treatment approaches to provide solutions for impaired fracture healing in the past included the utilization of growth factors, scaffolds, and MSCs. This approach was commonly referred to as the triangular concept. More recently, Giannoudis et al. have added an additional facet to this approach to fracture healing, emphasizing the importance of the mechanical environment [39]. This modified "triangular concept" is now referred to as the "diamond concept" and recognizes the importance of osteogenic cells, scaffolds, and growth factors for successful fracture healing *and* the mechanical environment of the fracture or nonunion.

Although initially the diamond concept was proposed for the treatment of acute fractures, surgeons have now extended its use to the treatment of fracture nonunions. According to the diamond concept, optimizing mechanical stability is particularly important in the treatment of

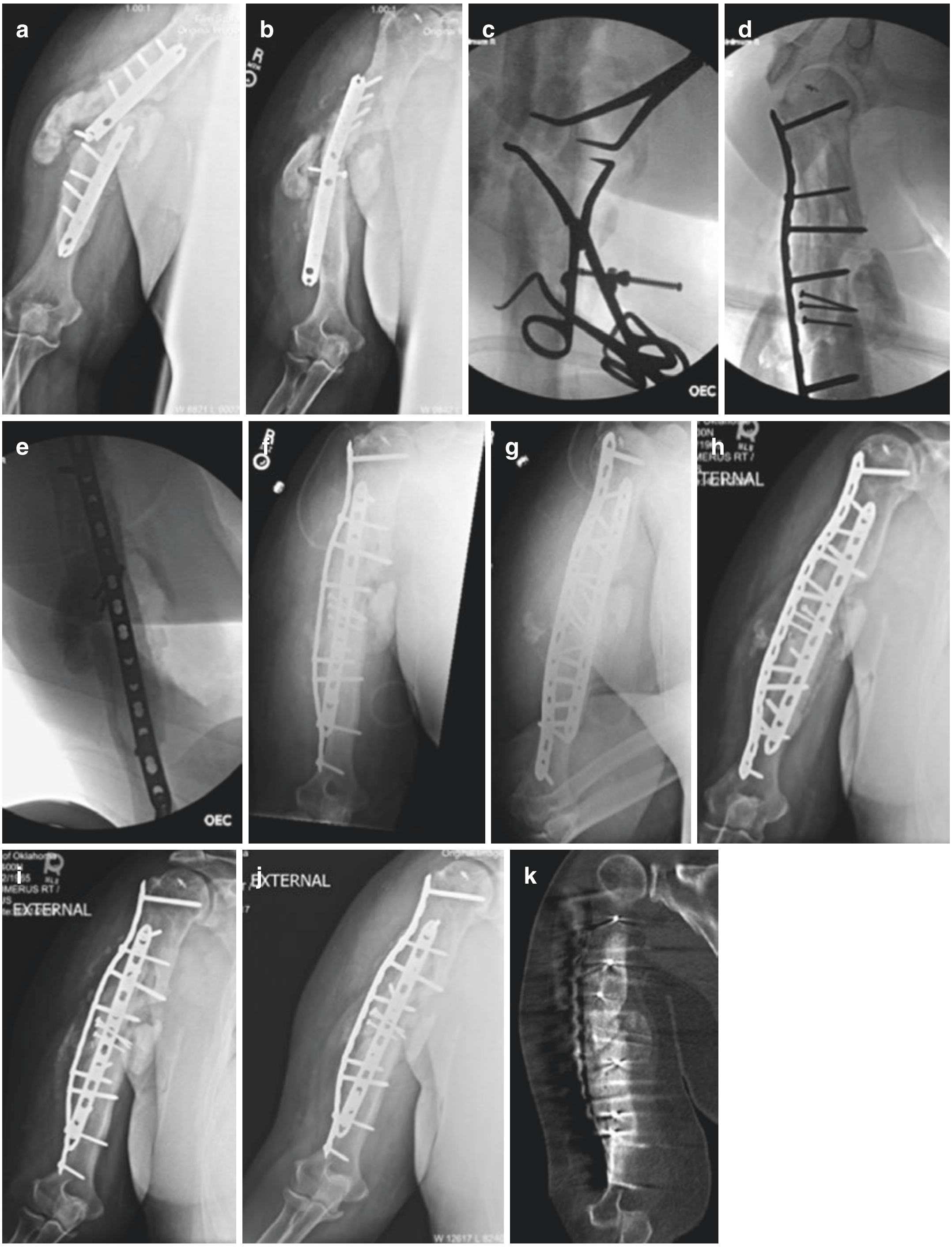

Fig. 3.8 Anteroposterior (AP) (**a**) and oblique (**b**) radiographs of a hypertrophic humeral shaft nonunion with failed plate. (**c–e**) Intraoperative fluoroscopic images during the repair of this nonunion. Inter-fragmentary compression between the fracture fragments is obtained with pointed reduction forceps and then lag screws. Large fragment neutralization plates are applied to provide additional stability and to protect the lag screws. Immediate postoperative AP (**f**) and lateral (**g**) radiographs of the humeral nonunion. AP, internal, and external rotation (**h–j**) radiographs taken 3 months postoperatively demonstrating primary bone healing progressing across the previous nonunion site. (**k**) CT scan at 8 months postoperatively confirming healing of the nonunion site. At this time the patient is pain free and has returned to his usual activities of daily living

hypertrophic nonunions. In these cases with an unstable mechanical environment following failed osteosynthesis, repeat osteosynthesis to improve the mechanical environment of the fracture/nonunion site is indicated. In “simple” hypertrophic nonunion cases, dynamization of the intramedullary nail with full weight-bearing is generally successful if performed within a reasonable time frame [40]. The dynamization process generally involves removing a locking bolt or two to allow the major fragments to come into contact with each other, which restores some of the stability and supports the healing process. In more complex hypertrophic nonunion cases (including those with failed osteosynthesis following ORIF with a plate and screws), repeat osteosynthesis is necessary, in addition to the opening and reaming of the medullary canals of each major fragment and placement of a larger intramedullary nail or replating to obtain absolute stability across the nonunion site(s). In the hypertrophic case presented above, the nonunion was “taken down,” the medullary canals opened up, and the nonunion reduced; rigid internal fixation was applied according to the diamond concept. This systematic approach has been recently shown to be very successful in a large series of humeral nonunions [41].

Conclusion

It is estimated that 7.9 million fractures occur each year in the United States. Impaired healing of these fractures is thought to occur in approximately 10% and is often felt to be the result of unfavorable healing environments—local, systemic, and biomechanical factors at the fracture site. Because fracture healing and bone regeneration is a complex process that involves multiple interacting biologic and biomechanical factors, it is critical that we continue to seek a better understanding of how each factor and factors that are still unknown affect fracture healing so that we may better treat our patient in an effort to speed fracture healing and restore limb function. Having a better understanding, the variety of different mechanisms by which fractures can be treated, combined with a better understanding of how different biomechanical environments affect fracture healing, will go a long way in decreasing these nonunion rates and improving fracture patient care.

References

1. Mountziaris PM, Mikos AG. Modulation of the inflammatory response for enhanced bone tissue regeneration. Tissue Eng Part B Rev. 2008;14(2):179–86.
2. Xing Z, Ku C, Hu D, Miclau T 3rd, Marcucio RS. Rejuvenation of the inflammatory system stimulates fracture repair in aged mice. J Orthop Res. 2010;28(8):1000–6.
3. Pountos I, Giannoudis PV. Fracture healing: back to basics and latest advances. In: Giannoudis PV, editor. Fracture reduction and fixation techniques. Cham: Springer International; 2018.
4. Lane WAL. The operative treatment of fracture. 2nd ed. London: The Medical Publishing Co., Ltd; 1914.
5. Shapiro F. Cortical bone repair. The relationship of the lacunar-canalicular system and intercellular gap junctions to the repair process. J Bone Joint Surg Am. 1988;70(7):1067–81.
6. Kaderly RE. Primary bone healing. Semin Vet Med Surg (Small Anim). 1991;6(1):21–5.
7. Kitaori T, Ito H, Schwarz EM, Tsutsumi R, Yoshitomi H, Oishi S, et al. Stromal cell-derived factor 1/CXCR4 signaling is critical for the recruitment of mesenchymal stem cells to the fracture site during skeletal repair in a mouse model. Arthritis Rheum. 2009;60(3):813–23.
8. Rahn BA, Gallinaro P, Baltensperger A, Perren SM. Primary bone healing. An experimental study in the rabbit. J Bone Joint Surg Am. 1971;53(4):783–6.
9. Aho AJ. Electron microscopic and histologic studies on fracture repair in old and young rats. Acta Chir Scand Suppl. 1966;357:162–5.
10. Parker MJ. Prediction of fracture union after internal fixation of intracapsular femoral neck fractures. Injury. 1994;25(Suppl 2):B3–6.
11. Robinson CM, Court-Brown CM, McQueen MM, Wakefield AE. Estimating the risk of nonunion following nonoperative treatment of a clavicular fracture. J Bone Joint Surg Am. 2004;86(7):1359–65.
12. Einhorn TA, Gerstenfeld LC. Fracture healing: mechanisms and intervention. Nat Rev Rheumatol. 2015;11(1):45–54.
13. Einhorn TA, Bonnarens F, Burstein AH. The contributions of dietary protein and mineral to the healing of experimental fractures. A biomechanical study. J Bone Joint Surg Am. 1986;68(9):1389–95.
14. Pountos I, Georgouli T, Blokhuis TJ, Pape HC, Giannoudis PV. Pharmacological agents and impairment of fracture healing: what is the evidence? Injury. 2008;39(4):384–94.
15. Burke D, Dishowitz M, Sweetwyne M, Miedel E, Handkenson KD, Kelly DJ. The role of oxygen

as a regulator of stem cell fate during fracture repair in TSP2-null mice. J Orthop Res. 2013;31(10):1585–96.
16. Jeffcoach DR, Sams VG, Lawson CM, Enderson BL, Smith ST, Kline H, et al. Nonsteroidal anti-inflammatory drugs impact on nonunion and infection rates in long bone fractures. J Trauma Acute Care Surg. 2014;76(3):779–83.
17. Goh SK, Yang KY, Koh JS, et al. Subtrochanteric insufficiency fractures in patients on alendronate therapy: a caution. J Bone Joint Surg Br. 2007;89(3):349–53.
18. Schilcher J, Michaelsson K, Aspenberg P. Bisphosphonate use and atypical fractures of the femoral shaft. N Engl J Med. 2011;364(18):1728–37.
19. Tyler W, Bukata S, O'Keefe R. Atypical femur fractures. Clin Geriatr Med. 2014;30(2):349–59.
20. Einhorn TA, O'Keefe RJ, Buchwalter JA, American Academy of Orthopaedic Surgeons. Orthopaedic basic science: foundations of clinical practice. 3rd ed. Rosemont: AAOS; 2007. p. 331–46.
21. Kwong FN, Harris MB. Recent developments in the biology of fracture repair. J Am Acad Orthop Surg. 2008;16(11):619–25.
22. Gerstenfeld LC, Alkhiary YM, Krall EA, Nicholls FH, Stapleton SN, Fitch JL, et al. Three-dimensional reconstruction of fracture callus morphogenesis. J Histochem Cytochem. 2006;54(11):1215–28.
23. Green E, Lubahn JD, Evans J. Risk factors, treatment, and outcomes associated with nonunion of the midshaft humerus fracture. J Surg Orthop Adv. 2005;14(2):64–72.
24. Pape HC, Giannoudis PV, Grimme K, van Griensven M, Krettek C. Effects of intramedullary femoral fracture fixation: what is the impact of experimental studies in regards to the clinical knowledge? Shock. 2002;18(4):291–300.
25. Perren SM. Evolution of the internal fixation of long bone fractures. The scientific basis of biological internal fixation: choosing a new balance between stability and biology. J Bone Joint Surg Br. 2002;84(8):1093–110.
26. Gerstenfeld LC, Cullinane DM, Barnes GL, Graves DT, Einhorn TA. Fracture healing as a post-natal developmental process: molecular, spatial, and temporal aspects of its regulation. J Cell Biochem. 2003;88(5):873–84.
27. Cho TJ, Gerstenfeld LC, Einhorn TA. Differential temporal expression of members of the transforming growth factor beta superfamily during murine fracture healing. J Bone Miner Res. 2002;17(3):513–20.
28. Sfei C, Ho L, Doll BA, Azari K, Hollinger JO. Fracture repair. In: Lieberman JR, Friedlaender GE, editors. Bone regeneration and repair. Totowa: Humana Press; 2005. p. 21–44.
29. Kon T, Cho TJ, Aizawa T, Yamazaki M, Nooh N, Graves D, et al. Expression of osteoprotegerin, receptor activator of NF-kappaB ligand (osteoprotegerin ligand) and related proinflammatory cytokines during fracture healing. J Bone Miner Res. 2001;16(6):1004–14.
30. Lee SK, Lorenzo J. Cytokines regulating osteoclast formation and function. Curr Opin Rheumatol. 2006;18(4):411–8.
31. Yang X, Ricciardi BF, Hernandez-Soria A, Shi Y, Pleshko CN, Bostrom MP. Callus mineralization and maturation are delayed during fracture healing in interleukin-6 knockout mice. Bone. 2007;41(6):928–36.
32. Bais MV, Wigner N, Young M, Toholka R, Graves DT, Morgan EF, et al. BMP2 is essential for post natal osteogenesis but not for recruitment of osteogenic stem cells. Bone. 2009;45(2):254–66.
33. Dimitriou R, Tsiridis E, Giannoudis PV. Current concepts of molecular aspects of bone healing. Injury. 2005;36(12):1392–404.
34. Breur GJ, VanEnkevort BA, Farnum CE, Wilsman NJ. Linear relationship between the volume of hypertrophic chondrocytes and the rate of longitudinal bone growth in growth plates. J Orthop Res. 1991;9(3):348–59.
35. Wendeberg B. Mineral metabolism of fractures of the tibia in man studied with external counting of Sr85. Acta Orthop Scand Suppl. 1961;52:1–79.
36. Carano RA, Filvaroff EH. Angiogenesis and bone repair. Drug Discov Today. 2003;8(21):980–9.
37. Rhinelander FW. Circulation in bone. In: Bourne GH, editor. The biochemistry and physiology of bone, vol. 2. New York, London: Academic Press; 1976.
38. Rhinelander FW. Tibial blood supply in relation to fracture healing. Clin Orthop Relat Res. 1974;105:34-81.
39. Giannoudis PV, Einhorn TA, Marsh D. Fracture healing: the diamond concept. Injury. 2007;38(Suppl 4):S3–6.
40. Andrzejowski P, Giannoudis PV. The 'diamond concept' for long bone non-union management. J Orthop Traumatol. 2019;20(1):21.
41. Miska M, Findeisen S, Tanner M, Biglari B, Studier-Fischer S, Grutzner PA, et al. Treatment of nonunions in fractures of the humeral shaft according to the Diamond Concept. Bone Joint J. 2016;98-B(1):81–7.

Part II

External Fixation Principles with Case Examples

4 Biomechanics of External Fixators for Fracture Fixation: Uniplanar, Multiplanar, and Circular Frames

Predrag Grubor and Joseph Borrelli Jr.

Introduction

An external fixator is a device used in bone and joint surgery and it serves to stabilize bone fragments using pins that pass through the parts of skeleton. Externally, it is attached to the construction of the external fixator [1–3]. This therapeutic method with the external fixator is referred to as an external fixation. It stabilizes and maintains broken bone fragments in the desired position. With the use of the fixator, the following can be achieved in bone fragments: neutralization, compression, dynamization, distraction, angulation, rotation, osteotaxis, ligamentotaxis, elastic fixation, and biocompression [1–3]. The notion of neutralization refers to the maintenance of the length of the limbs in order to avoid shortening. It is performed in comminuted, small fragment fractures that are too small to be stabilized directly in the external fixator frame and in the stabilization of fractures with bone defects. The external fixator over the fragment keeps a direct contact with the broken bone without moving. The force of pressure over the frame can accelerate the bone healing process [4]. Dynamization of the external fixator enables the axial forces to transfer from the apparatus to the bone, allowing micromotion at the fracture site [2]. Dynamization is achieved differently with each external fixator, depending on the frame construct. A fixator that uses thin wires generally does not need to be dynamized, in order to permit micromotion at the fracture site [4].

Distraction of bone fragments is used on the fracture site or after osteotomy, with the appearance of a "sticky" callus, in order to achieve the length of the injured bone, or the replacement of the bone fragment defect. It can be used in metaphyseal and intra-articular fractures in bone trauma [3]. The frame of the external fixator in its construction has the possibility to correct the angulation of the extremities. A universal joint can be used in the frame construction, so that, apart from the angulation, there is a possibility of correction of rotation as well [5].

As a term, osteotaxis was introduced by Hoffmann, and this term refers to the closed reposition of the fracture without opening the fracture site. Ligamentotaxis is a closed method where the ligaments and capsules of the joint are used for the reposition of fracture fragments, and the percutaneous application of the external fixator maintains the reposition until healing [6]. With the application of elastic fixation, in 1979

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Burny et al. showed that for the optimum healing, of bone fractures, a permanent presence of micromovement at the fracture site up to 1 mm is necessary [7]. It is achieved by the alternating stiffness and elasticity of the external fixator, by its mounting [7]. Biocompression means allowing the transmission of muscle force and body weight directly through the bone and not through the frame of the apparatus. In this, movements of broken bones are prevented from bending, rotating, dislocating, elongating, etc. [8].

Indications for External Fixation

Indications for external fixation are specific. Each problem must be individualized and familiar enough with other conventional, long-tested methods to solve the problem in the best possible way. Most authors consider that external fixation is the method of choice for open fractures (type II and III according to Gustilo), soft tissue compromise in primary surgery, and fractures that are associated with burns or skin loss that require reconstructive skin surgery [4]. External fixation gives good clinical results in the treatment of infected fractures, chronic bone infections, pseudoarthrosis, deformity correction, treatment of the discrepancies, and bone defects [3–5].

Biomechanics of External Fixator

The term biomechanics in orthopedics and traumatology of the locomotor system implies the study of physical properties of bones, muscles, cartilage, fasciitis, tendons, and joints in physiological and pathological state [1]. The most commonly used term in biomechanical testing is the rigidity and stiffness with regard to the apparatus for fixed bone fragments. The rigidity and stiffness of an external fixator construct are commonly considered when external fixators are used to treat an acute fracture and post-traumatic reconstruction [2–4].

The term *biomechanical rigidity* usually implies the externally stabilized fracture's resistance to the effect of three different forces:

- Axial; compression and distraction
- Bending; anteroposterior and lateromedial
- Torsion

Rigidity depends on the type of fracture and degree of reposition of fragments [8]. When working with the external fixators, biomechanical rigidity is dependent upon how the frame was constructed and which and how many components were utilized. Rigidity is higher in triangular, semicircular, and circular mounting of the frame of external fixators [8, 9].

The Importance of Pins and Wires in Biomechanics of External Fixators

The rigidity of external fixation depends not only on the type of mounting but also on the following:

- Type of fracture (the rigidity is higher in transverse, well-reduced fractures than in comminuted fractures that are not reduced)
- Configuration, number, and thickness of the pins
- Contact between the bone and the pins
- Distance between the bone and the frame
- Materials from which the pins are made
- Method of grouping the pins and the placement [7–10]

Thin wires (pins) and half-pins have a very important role in biomechanics of external fixators. In general the thin wires and half-pins and the pin to bar connections as well as the comminuted bone fragments are the biomechanical weak points of the external fixators [1]. Half-pins are most frequently fixed to the frame over the clamps from the outside, while there are different ways to be fixed to the bone. The construction of clamps ensures that their joint with the pins is firm and reliable. Pins without any threads, which were previously used, provided a highly unreliable bond. After 10–15 days, when certain bone resorption occurs, spontaneous slipping and loss of fixation occur. In order to minimize this slipping, pins are placed at an angle at the same level [10]. Prestressing of pins, especially in the case of bilateral mounting (most often in terms of compression at the fracture site), provides greater stability that is sufficient for the use of the Charnley fixator in arthrodesis, but prestressing is insufficient in unstable, comminuted, or fractures with bone defect [11]. Better stability between the half-pin and the bone is provided with half-pins with a thread at one end and centrally threaded pins with a thread in the middle [11].

In order to achieve the firmer connection between the bone and the pin, three types of threads are most commonly produced on the pins of the external fixators:

- A large pitch thread
- Conical thread
- A pin with a short thread at the top so that it is tightened only in the second cortex

A large pitch thread on the pin of the external fixator provides good conditions for the preservation of vitality of the bone tissue after placement of the half-pin. With a larger pitch, the volume of bone tissue within the thread is larger, the conditions of vascularization are better, and the risk of disruption due to osteolysis is less [11].

In order to make the bone thread as rigid as possible, a conical thread is used for the Orthofix fixator (Orthofix Medical, Lewisville, Texas, USA). This thread construction facilitates the application of the pins and provides good contact within the bone tissue. The removal of these wedges is easier [12].

A short-threaded pin passes through the near cortex and engages the far cortex. This pin embeds its threads only in the far cortex of the bone. In vitro studies have shown that this arrangement provides better fixation conditions and that the bending of the pin in the area of the surface of the bone is smaller, thus limiting conditions for loosening the half-pins [13].

As the distance between of the bars or rings increases the stability of the construct of the fixator to the bone through which the pin passes is greater, is decreased. The excessive micromotion allowed may encourage the development of a fracture nonunion. Therefore during application of the external fixator, the frame should be as close as possible to the skin (i.e., at the minimum possible distance). This is difficult to achieve in obese people so if there is an indication for an external fixation, choose the place where the bone is most superficial, and further increase the stability of the construct by using multiple pins and by grouping them together [13, 14].

It is clear that a greater number of pins influence the rigidity of the external fixator. The use of thin wires is often the reason for insufficient rigidity of many constructs. Mechanical examinations have shown that the rigidity of the external fixator increases if the thickness of the pin increases from 3 to 6 mm. In the bone model fixed with bilateral mounting with two pins above and below the fracture, the rigidity is four times greater if the fixation is accomplished with pins of thickness of 6 mm, than if it is accomplished with pins of thickness of 4 mm. Further increase of thickness over 6 mm does not seem to be practical, as further rigidity depends on the pin-bone compound [13].

The stability of the external fixators in which the pins are parallel (unilateral and bilateral) shows reversed biomechanical characteristics in relation to natural requirements. Unilateral mounting is the simplest application. In such mounting of the fixator, the stability in the frontal plane (in the pin level) in relation to their stability in the sagittal level (anteroposterior plane) is 12:1. Under the pin configuration, the angle of convergence between the pins is considered along the longitudinal axis. Previous examinations have showed that the most stable stability of the apparatus at the frontal and sagittal level is obtained when the pin convergence is 90° [1, 4].

Proper clustering of pins during the application of the external fixation device is very important for the stability of the fixation. The distance between the pins should not be larger than 4 cm. Pins made of stainless steel are preferred over pins made of titanium, ceramics, etc. What is unique about all external fixators with pins is the following: pins should be placed in intraseptal space between muscle groups and ideally through undamaged skin and outside the fracture hematoma if possible [15].

Semicircular and circular external fixators use stainless steel thin wires. These wires generally have diameters of 1.5 and 1.8 mm and lengths of 150, 170, 250, and 370 mm. These wires can have a spinous or pergamine tips, resulting in a lower heat than a Kirschner wire with a triangular tip while being inserted into the bone. Increasing the diameter to 1.8 mm increases the degree of tension by 5–6% (compression up to 30 kg) or up to 13% (compression greater than 30 kg). These are standard wires, and they come with stoppers, olive, in the form of a lapel or bayonet are used as well. The olive needles have a diameter of 1.5 mm and 1.8 mm and a length of 250 and 400 mm [15–18]. For the greater apparatus-bone stability, olive needles, the so-called console needles, are used. It is recommended to shrink the olive wire to 1.5–2 cm and shape the tip of the needle with beveling. The bactericidal effect of these wires can be achieved with a thin coating of silver, gold, and platinum [19].

Based upon the desired biomechanics, technical possibilities, geometric configuration, and the components of the external fixator constructed, external fixators may be placed in a variety of ways, including:

- Unilateral
- Bilateral
- V-frame
- Quadrilateral
- Triangular
- Semicircular
- Circular
- Unilateral with convergent pins

Placing one frame of the external fixator in one plane is called unilateral placing and two frames in one plane bilateral placing [2] (Fig. 4.1).

The unilateral external fixator (see Fig. 4.1a) has parallel pins in one plane. The first constructed external fixators were unilateral. The first reported users of unilateral external fixators include Malgaigne 1840, Langebeck 1851, Lambotte 1910, and Chalier 1917. Unilateral mounting of the external fixator creates a biomechanical imbalance between anteroposterior and laterolateral stability [11]. Recognition of this imbalance led to a transformation in the frame construction of the external fixator. To address this imbalance, Hoffmann in 1938, Charnley in 1948, and ASIF (Association for the Study of Internal Fixation) in 1952 showed that in addition to unilateral frames, these components could also be configured into bilateral, triangular, delta, and quadrilateral frames with much improved biomechanical stability and flexibility.

The stability of all unilateral external fixators has been improved by using larger diameter pins and greater number of pins and placing them in a convergent fashion and by placing at least one connecting bar close to the bone/skin. Therefore unilateral frames can provide stability of the fracture, if the external fixator is constructed following these concepts. In terms of axial stability, less stability has been noted with unilateral external fixators. Most unilateral external fixators now use lightweight bars made of carbon fibers that have increased their stability. Additionally, rods and multidirectional pins to bar clamps now available allow bilateral and triangular frames to be constructed. This helps to achieve the adequate stability regardless of the type of fracture or bone defect. By the anterior positioning of the frame, neutralization in anteroposterior level is best achieved [9], and the anterior application of the external fixator on the tibia provides the simplest clinical control [11]. Anterior mounting of the frame in one plane provides greater bending rigidity in that plane [9]. When these types of fixators are exposed to varus-valgus stress or torsional forces, they offer less stability and motion at the fracture site. If a unilateral frame needs to be mounted, in cases of tibia fractures, it should be placed in the midsagittal plane with half-pins placed a half-finger breadth medial to the tibial crest [9]. Frames placed in this manner provide better stability when subjected to axial forces at the fracture site. Asymmetric loads at the fracture site with a unilateral fixator in place do not meet the basic biomechanical standard of fracture repair [12].

Modern unilateral external fixators now use light rods such as “mono-tube,” ASIF tubular, Aesculap (Aesculap Inc. USA, Center Valley, Pennsylvania, USA), Stuhler-Heise fixator,

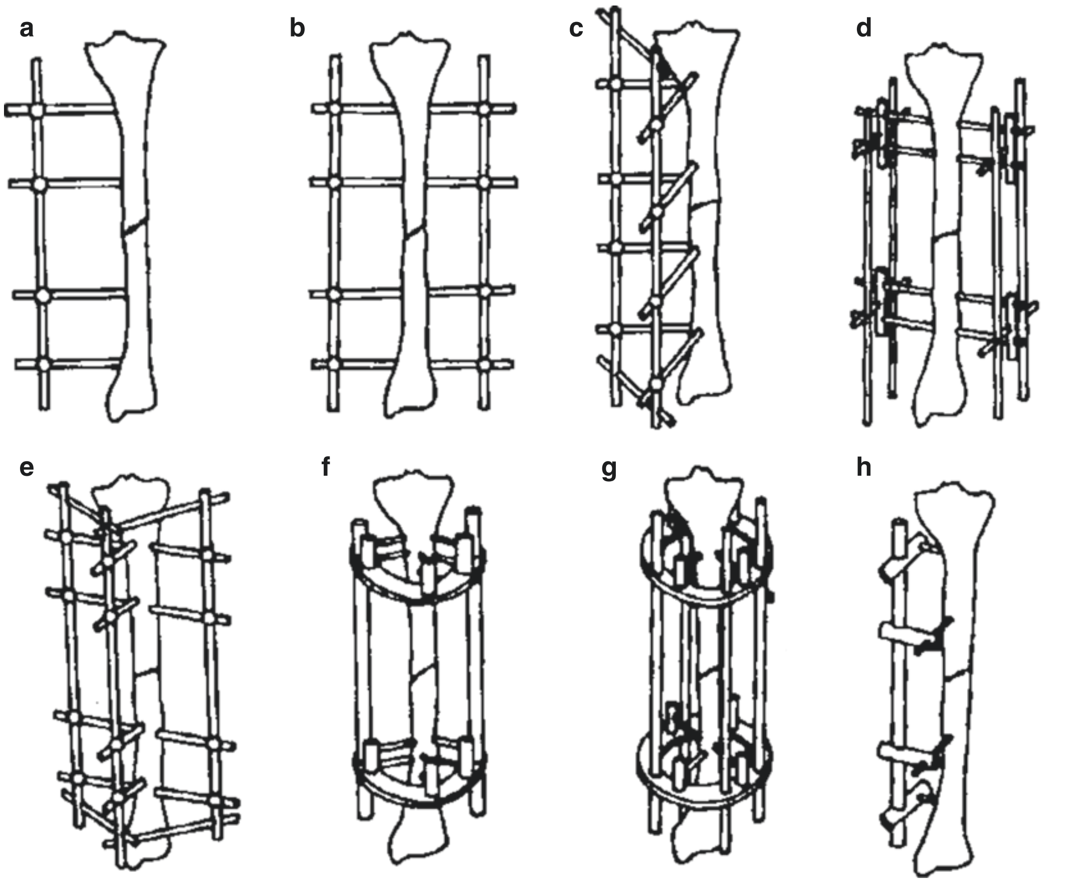

Fig. 4.1 Technical, geometric configurations of the frame of the external fixator

Orthofix, etc. A typical example of a unilateral external fixator is the "French external fixator" (Fig. 4.2), whereby pins are fixed through fixed holes that do not allow variations in pin positions when used.

The unilateral external fixator Orthofix, created by De Bastiani, represents a modern technological solution. It consists of two clamps that are connected to the central telescopic frame through the spherical joints. In each clamp, four pins can be placed in one plane and at the same distance [11]. The mobility of the clamp is provided by spherical joints, movable in all three planes, which allows adjustment of the anatomical segment, with a secure clamp block in the required position (Fig. 4.3). Generally, the employed half-pins are strong and conically cut. These flutes allow easier application, greater stability in the pin-bone contact, easier removal, and smaller infection around the pin. These types of pins also minimize the thermonecrosis of the bone, and generally there is little pain after the first twist of this conical threaded pin during removal. The central part of the fixator is a telescopic frame, which allows for compression, distraction, and biocompression among the fracture fragments. Use of this fixator in comminuted, multiple fragment fractures provides excellent stability of bone fragments and neutralization of weight-bearing forces. The thickness and length of the pins make it attractive only in the treatment of long bone fractures. Stability of the unilateral Orthofix frame is achieved by a distraction-compressive mechanism within the telescopic frame. These frames can be used in bone repair as well as in the treatment of leg length discrepancies, bone defects, and post-traumatic shortening [11, 12].

Fig. 4.2 Unilateral "French external fixator"

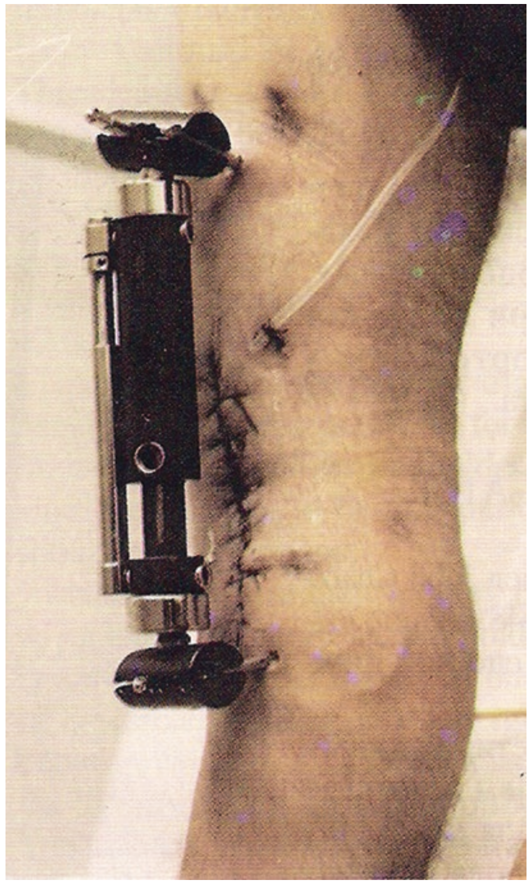

Fig. 4.3 Unilateral external fixator—Orthofix

A bilateral external fixator (see Fig. 4.1b) allows instability in the sagittal but good stability coronal plane. Sagittal plane instability is in general the most pronounced instability and is present only in uni- and bilateral mountings with parallel pins [1, 2]. The simple unilateral Charnley fixator can be configured as a bilateral frame (Fig. 4.4) with parallel pins (bilateral) and provides good, coronal plane stability but reduced sagittal plane stability.

Unilateral frame fixators are generally acceptable for the stabilization of long bones of the hands, humerus, radius, and ulna [20]. For the stabilization of femur structures, delta, V-frame, or quadrilateral frame is more acceptable. Threaded bars and nuts allow compression and distraction between bone fragments. Compression of up to 500N between fragments can be achieved [21]. In cases of bilateral frames, it is mandatory to use centrally threaded transfixation pins. This type of external fixator provides satisfactory stability, neutralization, compression, and distraction [20, 21].

Triangular (see Fig. 4.1e) external fixator frame construction and semicircular frames (Sl.1.F) provide great stability and rigidity in both the sagittal and coronal planes. Torsional stability is also much improved when compared to unilateral and bilateral frames and is sufficient to permit fracture healing [1]. In vivo torsion forces are far less as compared to the bending forces in the sagittal and coronal planes, as well as compression forces. The compression of

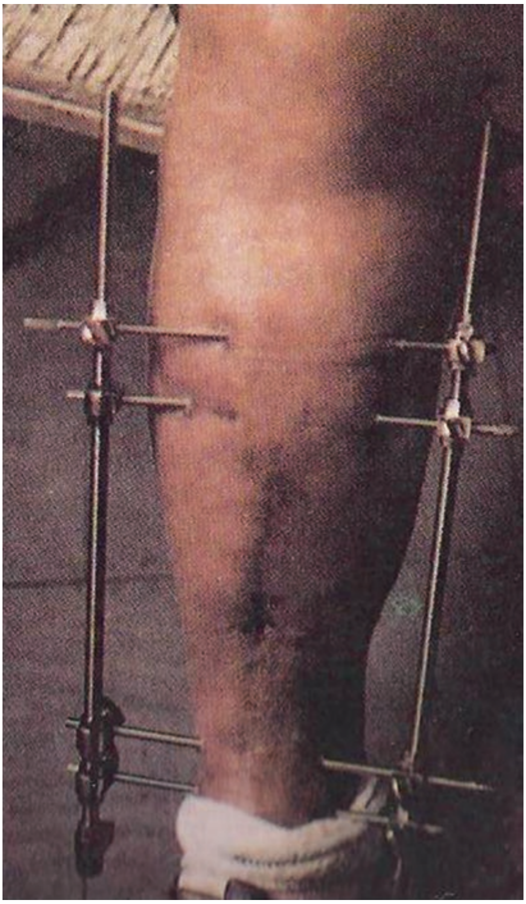

Fig. 4.4 Bilateral mounting of the frame to the Charnley fixator

Fig. 4.5 AO external fixator with triangular (V-frame) mounting of the frame

20–30 kg allows axial cyclic movements of 0.5–1 mm in these types of frames, which has been found to be the most optimal biomechanical stimulating effect in the fracture callus formation [6].

AO external fixation (***A****rbeitsgemeinschaft für* ***O****steosynthesefragen*; Association for the Study of Internal Fixation)—Instead of using solid external connecting bars, hollow tubes are generally employed, and the use of these tubes reduces the weight of the apparatus and allows varus, valgus, and rotation correction with multidirectional pin to bar clamps (Fig. 4.5).

The basic components of the AO external fixator are bars (tubes) of different lengths, four different types of wedge clamps, as well as pin to bar connectors and bar to bar connectors. These bars are attached to the main bone fragments with either half-pins (Schanz screws) or transfixation pins (Steinmann) and can be constructed in three basic configurations: unilateral, bilateral, and triangular [21, 22].

The type of frame configuration is determined by the personality of the fracture and the severity and injury pattern of the surrounding soft tissues. Unilateral mounting provides better stabilization in upper extremity injuries than when used to treat lower extremity fractures. For lower extremity fractures, in order to provide sufficient stability to support fracture healing, it is necessary to mount a bilateral and triangular frame. These types of frames can provide neutralization of weight-bearing forces and allow compression, distraction, and correction of angulation [23].

Unilateral external fixator with convergent pins (see Fig. 4.1h) provides uniform stability in the plane of the pins at the site of the fracture or corticotomy.

A unilateral external fixator frame can be constructed with parallel or convergent pins. Biomechanical tests have shown that mounting these frames with right angle convergence provides the same stability in the sagittal and coronal, as well as the axial plane [1, 7]. Mounting with a smaller pin convergence (approaching parallel) gradually decreases biomechanical stability across the fracture site. The superiority of mounting the pins with right angle convergence has been shown to provide optimum biomechanical conditions for osteogenesis in clinical practice [2, 3, 5]. Clinical results have shown that in the setting of the external fixator with the convergent pins, fractures heal faster with a more abundant, periosteal callus in relation to the external fixator with parallel pins [5]. Unilateral external fixator with converged pins is an external fixator according to Mitković M20 (Fig. 4.6).

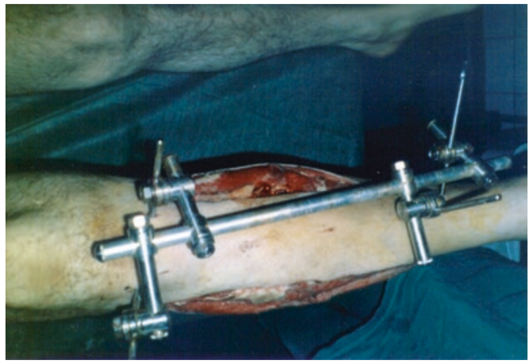

Fig. 4.6 External fixator according to Mitković M20

This fixator has a frame length of 320 mm, four movable clamps of 60 mm, four pins of 150 mm, and a weight of 650 g. This fixator achieves physiological biomechanical stability in all three planes, if the frame, movable clamp, pin, and the bone are placed at a convergent angle of 90°. The spacing between the movable clamps should not be greater than 40 mm and the distance from the skin to the pin holder up to 2 cm. This way, the application of the external fixator provides a uniform stability in the sagittal and coronal planes, creating ideal biological and biomechanical conditions for bone healing. The pins placed in this manner contribute to the osteogenic periosteal reaction at the site distant from the fracture, and since the periosteal reaction is generalized, the osteogenic processes are more extensive than with parallel placed pins (Fig. 4.7).

Threads should be placed at the distance of 40 mm between each other. With this fixator, apart from unilateral frame, bilateral and triangular constructions are possible as well. By using the articulated clamp and M20 in the construction, semicircular construct of the fixator can be performed. Doing so allows greater therapeutic possibility including the use of curved wires and olive wires.

Unilateral external fixator by Shearer, which utilizes convergent pins, is a disposable external fixator (Fig. 4.8). The components are sterilized with gamma rays of 25kGray. It is packaged in two different sizes: a larger set for the lower extremity and a smaller set for the upper extremity.

One set consists of a unilateral frame that is in the form of two tubes that are connected in the middle with a special articulated clamp. The clamp allows the tube of the unilateral frame to be placed at the ideal position and, if needed, allows compression and distraction at the fracture site. Articulated part enables the frame to follow the anatomical shaft of the bone. Six highly movable clamps allow the setting of convergent wedges only at an angle of 60° to the bone axis. The position of the wedge on the mobile clamp is fixed, so the accuracy in the placement of the wedge is required. It is necessary to set the most proximal and distal wedge, fix the frame to the installed wedges, and alternately set the other four proximal and distal wedges. When the reposition is done, the compression device can achieve the required compression between the fragments.

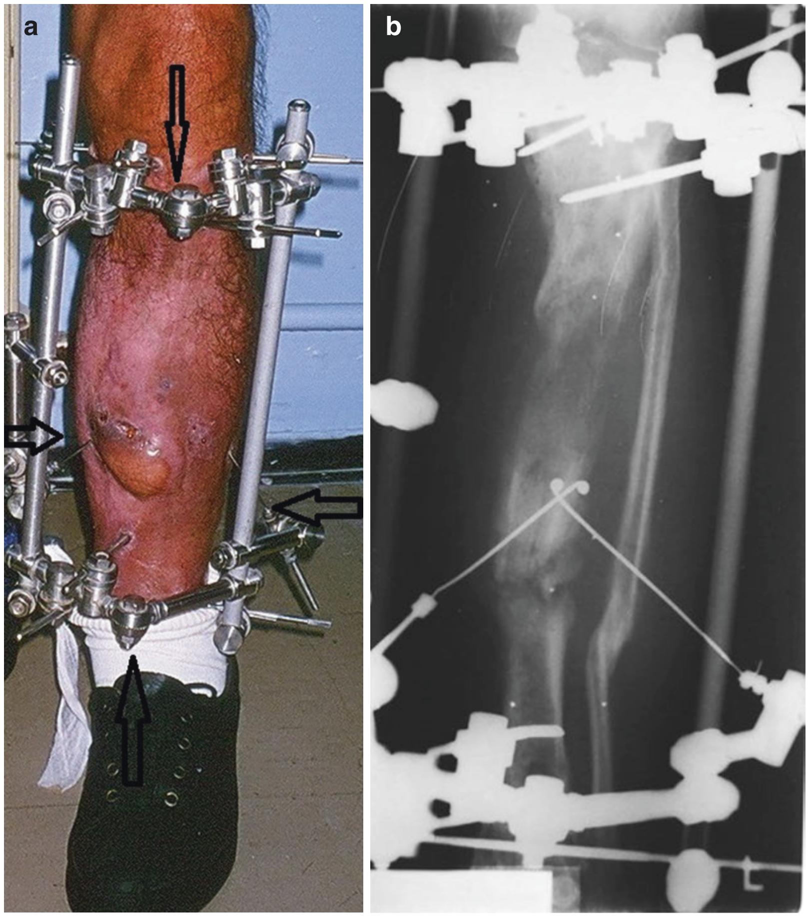

Fig. 4.7 (**a**) Mounting the polycircular frame M20 for distraction osteogenesis, after the reconstruction of skin defect with myocutaneous flap, (**b**) radiography of bone defect repair

If it is necessary to choose between the unilateral fixator with convergent wedges (Shearer and Mitković M20) in the management of high-energy fractures, the authors would prefer the Mitković M20-type fixator as it is simpler and faster to apply and has unlimited number of mounting variations and for its relatively low cost compared to the other types of frames available as well as the simplicity of production when needed in large numbers.

Quadrilateral frame consists of two double bilateral frames (see Fig. 4.1d). This kind of mounting of the external fixator achieves good stabilization of bone fragments in all planes [24]. The most recognizable quadrilateral frame construction is the one with the Hoffmann external fixator. In 1938, Raoul Hoffmann, a Swiss surgeon from Geneva, described an external fixator (Fig. 4.9) that now bears his name. Hoffmann was not only a surgeon but also a doctor of theology and a carpenter. This type of frame is typically applied in the care of patients who have suffered high-energy fracture of the lower extremities such as occur in war and high-speed motor vehicle crashes and when pedestrians are struck by automobiles. The original version of the Hoffmann device has transformed, but the basic construction and principles have remained the same. The transformation was described by Ray, Vidal, and Adrey, who, by working with this fixator, improved the static and dynamic properties as well as the mode of application [23–25]. It is available in three sizes and therefore suitable for application on all segments of the locomotor system, as well as on children. During construction of this frame, it is

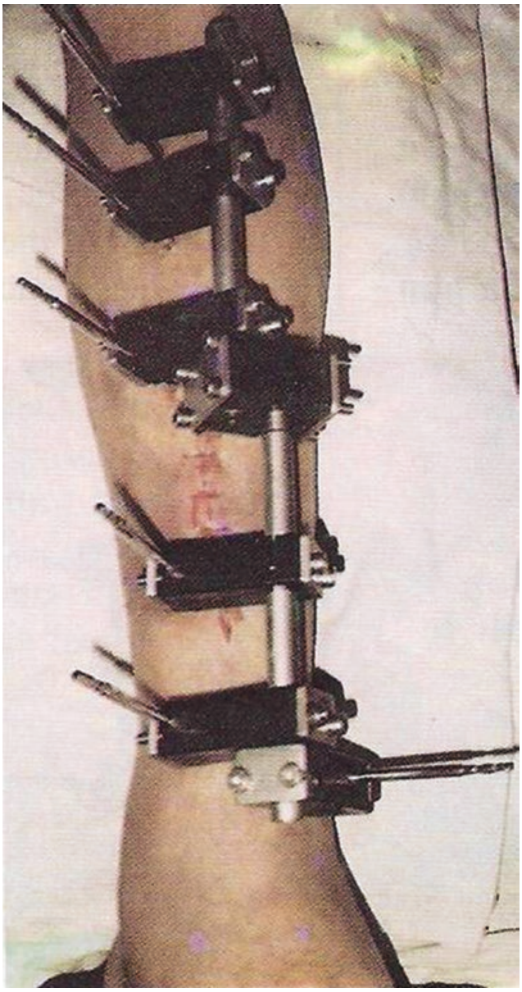

Fig. 4.8 The J. R. Shearer external fixator

very important to place the parallel pins at the predetermined distance [25]. The pin is applied manually with a self-threading wedge. Four pins are grouped in one frame and fixed with one clamp. The distal and proximal clamps are connected with the use of a sliding rod, which allows the neutralization, compression or distraction of bone fragments. The fixator has the capability of being of constructed as a unilateral, bilateral, delta, triangular, or quadrilateral frame. The experience and knowledge of the surgeon influences the setting of a rational number of pins, clamps, and bars to construct a frame that provides the best biomechanical conditions to support bone healing [24, 25]. In general, at least two points of fixation are necessary for the proximal and distal bone fragments. Additional fixation is desirable into each fragment when additional pins or wires can be accommodated without compromising the joint capsule, the fracture hematoma, or the path of future surgical incisions. If a large intercalary fragment of the diaphysis is present and the above conditions can be met, it is sometimes desirable to stabilize this fragment as well. Doing so will improve the stability of the construct and aid in the restoration of the length, alignment, and rotation of the limb.

For fractures of the humerus, radius, and ulna, a unilateral frame construction of the Hoffmann fixator will provide adequate stability to support fracture healing. For fractures of the tibia or femoral diaphysis, a delta, triangular, or, if deemed necessary, a quadrilateral frame can be constructed. Generally, the type of frame constructed will depend upon the personality of the fracture, size, and demands of the patient and the purpose of the frame (definitive vs. temporary). In the case of proximal humerus injuries, *spina scapulae* frames can be used to set proximal clamps. Historically, pelvis fractures were almost exclusively stabilized with the Slätis construction of the Hoffmann fixator, that is, the use of a trapezoidal frame [24–26].

Closed Frame Can Be Circular or Semicircular The circular and semicircular fixator (Fig. 4.10) with a wire has good anteroposterior and laterolateral stability and the possibility of using dynamization. Permanent elastic compression causes the physiological compression between bone fragments, supported by the contraction of muscles of the extremities [27].

Circular external fixator with ***Kirschner wires*** includes percutaneous placement of needles with or without an olive, their adequate intersection at an angle of 90°, which is ideal, but not less than 60°, to ensure maximum stability and tension of the ring. The thin wires are positioned in different planes as close as possible to the ring while respecting the path and position the neurovascular structures and soft tissues [27–29].

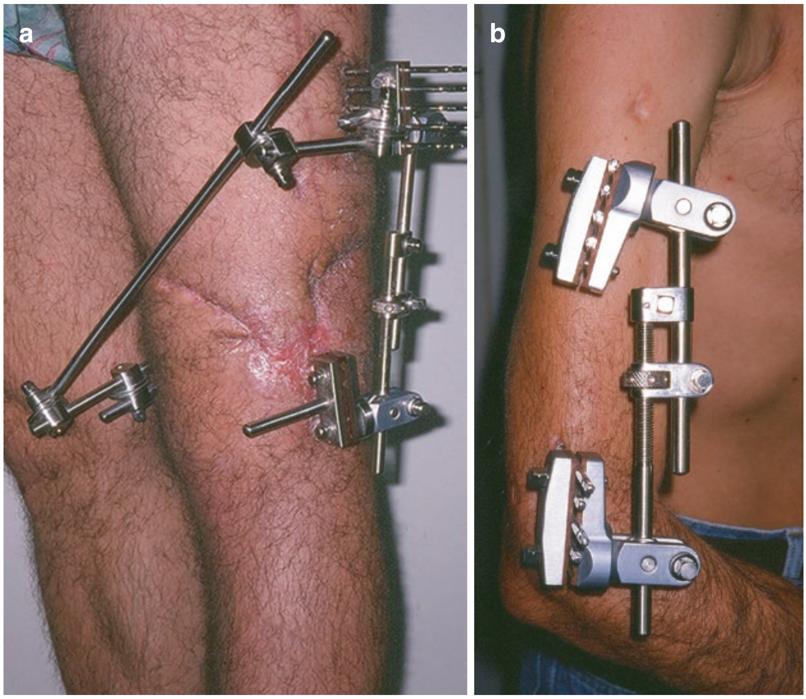

Fig. 4.9 (**a**) Delta and (**b**) unilateral mounting of the Hoffmann fixator

Aseptic techniques must be followed during placement of these wires in an effort to avoid secondary complications (primarily infections). Each wire should be placed, particularly when they are located near the joint, with extreme care to assure that the wire does not penetrate the joint capsule, pierce nearby tendons, or restrict joint movement or cause soft tissue impingement [28]. In children, thin wires used for the construction of circular frames should be positioned to assure that they are not inserted into the growth plates. Additionally, to minimize the risk of soft tissue injury, the lowest possible energy as possible should be used to place these wires. Many surgeons choose to carefully drill the wires through the bone and once the wire has penetrated the far cortex to tap them the rest of the way with a light mallet.

Proper tensioning of the thin wires (approximately 90–130 kg) is essential for maximizing the stability of the construct [29, 30]. The thickness of the wires chosen is influenced by the application site and the patient's age and size. That being said in general 1.8 mm wires are the most commonly used in conjunction with the so-called Rule of Two—when applying the external fixator of Ilizarov:

- 2 cm distance between the frames of the fixator and the skin
- Two rings per anatomical segment
- Two wires per ring of the fixator
- Two to four distancers per apparatus

In adults, wires of 2, 4, or 6 mm diameters are more commonly used most, especially in Western Europe and the USA. These wires are suitable for the stabilization of most fractures in the area of the diaphysis [30]. Thin wires and wedges impregnated with hydroxyapatite are increasingly being used in an effort to increase the rate of osteogenesis [31]. Another convenience of the circular fixator by Ilizarov is the possibility of angular corrections, as well as being able to perform axial compression and distraction [26–30]. Also, these frames can be applied to small bones including metacarpals and metatarsals, as well as the clavicle [26].

Technological advancements have continually improved the various components of external

Fig. 4.10 Circular external fixator by Ilizarov

fixators available today. The most commonly used external fixators used today include:

- Circular external fixator by Ilizarov, Taylor Spatial Frame® (Smith & Nephew, Andover, Massachusetts, USA)
- Rigid monolateral (unilateral) external fixators: Wagner, Orthofix, mono-tube
- Monolateral external fixators with angulation: Heidelberg
- Intramedullary fixators: Intramedullary skeletal kinetic distractor, Albizzia, Fitbone® (Wittenstein, Igersheim, Germany)

By the use of olive wires, twisted wires, and an adequate number of rings and the use of conical pins, fracture fragments are prevented from translating along the wire. Additionally, the spacing of the telescopic frames on long bones must be the same in order to achieve even load and stability [27, 28]. The best stability is achieved with the use of the four-ring device applied with these principles in mind:

- A basic ring is placed as far proximally and distally from the fracture as possible while avoiding placement of the thin wires within the capsule and growth plates as mentioned previously.
- At least two additional rings are positioned as close to the fracture as possible while trying to avoid the local fracture hematoma and injured soft tissues.
- A minimum of the wires or pins, with or without an olives, are placed through each major bone fragment at various levels and in different planes [30].

The mechanical characteristics of this fixator allow the positive effects of axial microcircuits without the harmful effects of torsion and translational bending [27]. The advantages of an axial micromovement for enhanced improvement and bone remodeling, as well as bone regeneration of the bone, are provided with the use of the technique of dynamization by gradually dismantling the fixator. One way to decrease the stability of the fixation and allow controlled stressing at the fracture site is to remove one wire or half-pin from the ring periodically during fracture healing while maintaining the tension of the remaining wires [28]. When applying these types of frames, it is important to keep in mind that the ring diameter affects the stability of circular fixators. Frames with a smaller ring diameter are more stable than frames with a larger diameter of the same thickness. By reducing the ring diameter by 2 cm, the rigidity of the frame increases by 70% [29]. Therefore, the smallest ring diameter in which the extremity fits should be used during frame construction. Of course postoperative edema of the extremity should be taken into account when choosing the size of the ring to use [30]. In most cases, it is necessary to provide at least 2 cm of space between the skin and the rim of the ring. When this rule of thumb is not followed, the ring can cause compression of the limb leading to additional edema as well as ulceration of the skin. Different ring diameters can be mounted to the same extremity in order to give comfort and optimize the stability of bone fragments. In each case the bone segment should ideally be positioned within the center of the ring. In segmental fractures, it is recommended to position two rings on each segment of the bone for the adequate stability [26–30].

One of the Most Commonly Used Semicircular External Fixators Is the Volkov-Oganesyan Fixator This type of frame construct can be used to manage stiff nonunions, fibrosed joints, and neglected joint dislocations, as well as open and closed fractures, and in the treatment of corrective osteotomies for the treatment of congenital limb abnormalities. This type of fixator is composed of two or three connections between the different structures (Fig. 4.11). These segments are joined with three semirings through which Kirschner wires are placed. These wires reach the bone fragments at an angle of 90°. Two axial connections provide great mobility in the sagittal and coronal planes, and they also act as a hinged joint that can be used for correction of flexion and extension deformity in the sagittal plane. If indicated, compression and distraction can be performed with this type of external fixator. This type of external fixator has also shown good clinical results in the treatment of contracted or ankylosed joints. When using this type of fixator for the treatment of an ankylosed joint, it is necessary to initially distract the joint 1–2 mm. In joints where distraction is not possible, *arthroscopic arthrolysis* will need to be performed. Postoperatively, with this fixator in place, attempts are made to slowly eliminate contractures and achieve full range of motion of the joint. The contractures are addressed with the use of a flexible/extension clamps, 1 m per day, which corresponds to approximately 3° less flexion. Every day, with the movement of 1 mm of middle clamp of the frame, an increase in the function of the joint by 3° is expected and continued until full flexion and extension is achieved. In the second round, obtaining a full function of the joint can be increased daily on the third frame by 3–4 mm, that is, 9°–12°. The treatment lasts 7–8 weeks, depending on the contracture, age of the patient, and stiffness of the joint. The fixator is worn until complete flexion and extension is recovered. While treatment with the fixator takes a considerable amount of time, during which the patient should be participating in supervised physical therapy. When full joint function has been achieved within the frame, active exercises should be continued for an additional 8–10 days. At this time generally the frame can be removed while active physical therapy is continued.

Fig. 4.11 Volkov-Oganesyan semicircular external fixator

Osteo Mechanic, Kotajev external fixator uses the thin wires and wedges to stabilize bone fragments. It is made of perforated two-thirds rings; these semicircular rings allow placement of an adequate number of thin wires to be placed while avoiding joint impingement, and if necessary, additional rings can be stacked to further increase strength. The semicircular rings are interconnected with telescopic rods, which can produce interfragmentary compression or distraction. To further improve the adaptability of this type of frame, perforated plates can be added to the rings, to which the wedges and thin wires are mounted (Fig. 4.12). In primary fixation, this external fixator uses thin wires and pins to stabilize the fracture or osteotomy fragments. During treatment, this otherwise rigid fixator can be transformed

Fig. 4.12 Igor Kotajev external fixator uses needles and wedges

into a more elastic one, with the use of telescopic holders between the semicircular rings.

Summary

Over the years a wide variety of different types of external fixators have been developed and used to address a variety of skeletal problems. Although these frames vary considerably in appearance and in their application, all of these frames share similar components. In general, each frame uses some sort of thin wire or half-pin inserted into the many fracture fragments or bone segments. These wires or pins are then attached either by "pin to bar clamps" or ring to wire clamps to attach them to either rings, partial rings (semicircular), or bars. In those frames that use bars, these bars are further attached to other pin to bar clamps, and thus other half-pins along the length of the bone being stabilized. In the frames that use rings, the series of rings are connected either with fixed threaded rods that can be manipulated to create compression or distraction across the fracture or osteotomy site or with hinged rods that allow for correction of angulation.

Because of the great flexibility that these frames allow, they can be used for a wide variety of different conditions. However, with the ever-increasing technological advancements of plates and intramedullary nails, and operative techniques such as percutaneous plating, as well as the advancements in the treatment of open fractures, external fixators are being used less commonly as a definitive means of treatment. This being said, external fixators are still widely used in the setting of acute fractures with significant associated soft tissue injury and for open fractures with soft tissue defects or those open injuries where there is an unavoidable delay in the initiation of care. Additionally, external fixators are still widely used in damage control orthopedics (DOC) in polytraumatized patient and for high-energy displaced intra-articular fractures, with axial stability, that are best treated during the light of day and when the soft tissues, and the patient, have had a chance to recover. In these last two scenarios, the external fixators are used to temporarily stabilize and reapproximate the fracture fragments until safe definitive management can be undertaken.

In summary, this chapter provides a review of the main different types of external fixators that have been developed over the years. Although it is generally not necessary for orthopedic surgeons to have mastered the use of each of these types of frames, it is imperative that orthopedic surgeons who treat acutely injured patients, or patients with post-traumatic osseous complications, to have in their armamentarium a working knowledge of the most widely used and readily available external fixators to assure appropriate treatment of each of their patients.

References

1. Mitković MI. Spoljnja fiksacija u traumatologiji: razvoj i primena aparata autora. Niš: Prosveta; 1992.
2. Grubor PR, Grubor MI, Asotic MI. Comparison of stability of different types of external fixation. Med Arh. 2011;65(3):157–9.
3. Perren SM, Boitzy AL. La différenciation cellulaire et la biomécanique de l'os au cours de la consolidation d'une fracture. Anat Clin. 1978;1(1):13–28.
4. Grubor PR. Treatment of war wounds: the extremities. New York: Nova Science Publishers; 2014.
5. Grubor PR. Spoljnja fiksacija u koštano-zglobnoj hirurgiji. Banja Luka: Medicinski fakultet; 2013.
6. Hoffmann R. Rotule à os pour la direction dirigée non sanglante des fractures (ostéotaxis). Helv Med Acta. 1938;5:844–50.
7. Burny F, El Banna S, Evrard H, Vander Chinst M, Degeeter L, Peeters M, et al. Elastic external fixation of tibial fractures. Study of 1421 cases. In: Brooker AF, Edwards C, editors. External fixation: the current state of the art. Baltimore: Williams & Wilkins Co.; 1979. p. 55–73.
8. Mitković M. Fracture treatment with M.9 external fixation system. 90° convergent orientation of the pins. 18th World Congress (SICOT '90), Proceedings. Montreal: Société Internationale de Chirurgie Orthopédique et de Traumatologie; 1990. p. 479–80.
9. Dell'Oca F, Alberto A. Modular external fixation in emergency using the AO tubular system. Montevideo: Mar Adentro; 1989.
10. Behrens FF, Searls K. External fixation of the tibia. Basic concepts and prospective evaluation. J Bone Joint Surg Br. 1986;68(2):246–54.
11. De Bastiani G, Aldegberi R, Renzi Brivio L. Dynamic axial fixation. Clin Orthop. 1986;10(2):95–9.
12. De Bastiani G, Aldegheri R, Renzi Brivio L. The treatment of fractures with a dynamic axial fixator. J Bone Joint Surg Br. 1984;66(4):538–45.
13. Green SA. Complications of external skeletal fixation. Clin Orthop Relat Res. 1983;180:109–16.
14. Aro HT, Hein TL, Chao EY. Mechanical performance of pin in external fixators. Clin Orthop Relat Res. 1989;248:246–53.
15. Carpenter JE, Hipp JA, Gerhart TN, Rudman CG, Hayes WC, Trippel SB. Failure of growth hormone to alter the biomechanics of fracture-healing in a rabbit model. J Bone Joint Surg Am. 1992;74(3):359–67.
16. Gasser B, Boman B, Wyder D, Schneider E. Stiffness characteristics of the circular Ilizarov device as opposed to conventional external fixators. J Biomech Eng. 1990;112(1):15–21.
17. Duda GN, Kassi JP, Hoffmannn JE, Riedt R, Khodadadyan C, Raschke M. [Mechanical behavior of Ilizarov ring fixators. Effect of frame parameters on stiffness and consequences for clinical use.] Unfallchirurg. 2013(10):839–45. *[Article in German]*.
18. Ilizarov GA. Transosseus osteosynthesis. Theoretical and clinical aspects of the regeneration and growth of tissue. Berlin: Springer; 1992. p. 63–136.
19. Saragaglia D, Pernoud A, Tourné Y, Leroy JM, al Zahab MA. [Tibiotarsal arthrodesis: value of external fixator associated with in situ cancellous bone graft. Initial results apropos of 18 cases]. Rev Chir Orthop Reparatrice Appar Mot. 1995;80(1):51–7. *[Article in French]*.
20. Fabrin JE, Larsen KI, Holstein PE. Arthrodesis with external fixation in the unstable or misaligned Charcot ankle in patients with diabetes mellitus. Int J Low Extrem Wounds. 2007;6(2):102–7.
21. Perren SM. Biological internal fixation: its background, methods, requirements, potential and limits. Acta Chir Orthop Traumatol Cech. 2000;67(1):6–12.
22. Jakob RP, Fernandez DL. The treatment of wrist fractures with the small AO external fixation device. In: Uhthoff HK, Em S, editors. Current concepts of external fixation of fractures. Berlin/Heidelberg: Springer; 1982.
23. Benum P, Svenningsen S. Tibial fractures treated with Hoffmann's external fixation: a comparative analysis of Hoffmann bilateral frames and the Vidal-Adrey double frame modification. Acta Orthop Scand. 1982;53(3):471–6.
24. Vidal J, Rabishong P, Bonnel F, Adrey J. Étude biomécanique du fixateur externe d'Hoffmann dans les fractures de jambe. Montpellier Chir. 1970;16:43–52.
25. Finlay JB, Moroz TK, Rorabeck CH, Davey JR, Bourne RB. Stability of ten configurations of the Hoffmann external-fixation frame. J Bone Joint Surg Am. 1987;69(5):734–44.
26. Fleming B, Paley D, Kristiansen T, Pope M. A biomechanical analysis of the Ilizarov external fixator. Clin Orthop Relat Res. 1989;(241):95–105.
27. Podolsky A, Chao EY. Mechanical performance of Ilizarov circular external fixators in comparison with other external fixators. Clin Orthop Relat Res. 1993;(293):61–70.
28. Ilizarov GA, Emilyanova HS, Lebedev BE. Some experimental studies. Mechanical characteristics of Kirschner wires. In: Perosseus compression and distraction osteosynthesis in traumatology and orthopedics. Kurgan Publishers, 1972. p. 14–25.
29. Calhoun JH, Li F, Bauford WL, Lehman T, Ledbetter BR, Lowery R. Rigidity of half pins for the Ilizarov external fixator. Bull Hosp Jt Dis. 1992;52(1):21–6.
30. Tomic SL, Bumbasirevic MA, Lesic AL, Bumbasirevic VE. Modification of the Ilizarov external fixator for aseptic hypertrophic nonunion of the clavicle: an option for treatment. J Orthop Trauma. 2006;20(2):122–8.
31. Popkov AV, Gorbach EN, Kononovich NA, Popkov DA, Tverdokhlebov SI, Shesterikov EV. Bioactivity and osteointegration of hydroxyapatite-coated stainless steel and titanium wires used for intramedullary osteosynthesis. Strategies Trauma Limb Reconstr. 2017;12(2):107–13.

5 Diaphyseal Fractures

Dolfi Herscovici Jr. and Julia M. Scaduto

Introduction

External fixation is synonymous with the management of trauma and has been since Hippocrates described the use of an external device to manage a tibia fracture approximately 400 years BC. More than 2000 years later, formal external fixation was first recorded in 1853 when French surgeon and medical historian Joseph-François Malgaigne used it to manage a fractured patella [1]. Over the next 100 years, multiple European and American surgeons added to the knowledge of the use of external fixation. Although these changes attracted the attention of some surgeons, a 1950 survey, of practicing surgeons commissioned by the American Academy of Orthopedic Surgeons (AAOS) as well as the American Association for the Surgery of Trauma and the Iowa Medical Association (a representative state), found that only 27% believed that external fixation had a place in the management of fractures. The remaining majority felt that frequent pin-site infections, difficulties in application, the struggle to obtain and maintain a reduction, and an inadequate rigidity of fixation systems made this type of treatment inadvisable [2]. It is important to note that during the time the survey was performed, the standard of fracture care consisted of either prolonged traction, a variety of casting techniques, the use of pins, or combinations of pins and plaster in special circumstances. At that time surgery was reserved only for fractures involving the hip and the femoral shaft. Using these techniques, the primary goal was not to obtain length and alignment but to simply get the fracture(s) to heal.

In the 1970s and early 1980s, surgeons in the United States renewed their interest in external fixation and began publishing reports documenting its usefulness for a variety of fractures and dislocations [3, 4]. This led to a recognition that the available literature was unstructured regarding the indications for use, the pitfalls, and the successes associated with external fixation. In the 1980s, Fred Behrens became one of the leading advocates of external fixation, offering general theories, basic concepts, and principles for its use, while describing the potential that external fixation had to offer for the care of trauma patients [5–8]. Along with others, the phrases *half-pins, bars and rods, simple and clamp fixators, unilateral/bilateral* or *multiplanar frames, corridors for pin insertion, ligamentotaxis, axial alignment, indirect reductions, transarticular fixation, spanning frames,* and *soft tissue management* became part of the lexicon of trauma care. It should also be understood that these devices became more popular as a better understanding of biomechanics and fixator principles was developed and as implant companies began to develop

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devices that were more user-friendly and reproducible and the indications were better defined and expanded for the management of trauma patients.

Today external fixation has become such an essential tool in trauma that it would be difficult to conceive of caring for severely injured patients without the use of these devices. Currently, the use of external fixation for acute fracture care is divided into two categories: definitive and temporary. Temporary fixation for diaphyseal and intra-articular fractures is further subdivided into "damage control" and provisional use. This chapter will address both of these short-term applications, providing information regarding the placement of these devices and how the extremity and fractures should be managed during the use of temporary fixation.

Implant and Anatomic Considerations

Temporary fixation needs to provide enough flexibility to allow manipulation of the fractures and enough strength to maintain the reduction until definitive fixation can be safely undertaken. Although bilateral and multiplanar fixation increase construct stability, they take longer and are more cumbersome to assemble, may have a higher potential for pin tract infections, and may impede the management of the surrounding soft tissues and wounds. In contrast, uniplanar fixation, although four to seven times biomechanically weaker when stressed in the plane orthogonal to the pins [9], is easier and quicker to apply and allows for greater access to the surrounding soft tissues and wounds. The three components necessary for construction of external fixator frames include pins, clamps, and rods (connecting bars).

Pins

External fixation pins come in a variety of sizes, shapes, and lengths. Half-pins are most commonly used and range in size from 2.5 mm to 6 mm in diameter, with centrally threaded (transfixion) pins also available. Although thin wire (1.5–2 mm) fixation is available, these require the use of half-rings or rings and necessitate tensioning of the thin wires to increase rigidity and obtain stability. These frames typically require more time to construct and to place than half-pin fixation.

Design, geometry, and thread lengths of half-pins vary considerably according to the manufacture(s). However, all pins act similarly with bending strength increased to the fourth power of the increase in the pin's radius [9]. Thread designs consist of tapered tips, self-drilling tips, and cancellous and cortical threads and are available in stainless steel, titanium, and titanium-copper alloy. In addition, some pins are coated with nitric oxide, chitosan, chlorhexidine and iodine, monolaurin, silver, titanium zirconium nitride, or titanium zirconium silver, in an effort to decrease the risk of pin tract infections [10–13], or with hydroxyapatite or bisphosphonate, to improve the pin-bone interface, resulting in greater extraction torque [14, 15]. However, coated pins are rarely used during temporary frames. To minimize stress risers and possible iatrogenic fractures, pin diameter should not exceed one-third of the diameter of the bone [9].

The success of external fixation is dependent on the stability at the pin-bone interface. Stability with half-pins is optimized with purchase of the far cortex. The weakest point occurs at the thread-shank (smooth) junction. This junction forms a large stress riser and, when it occurs at the same level as the pin-bone interface, can promote early fatigue and fracture of the pin. Modifications to prevent pin breakage include using a pin with longer threads, which moves the stress riser away from the bone surface; using a tapered (conical-shaped) threaded pin, which limits the magnitude of stress risers at any single point on the pin; or using a pin with a shorter thread to allow the shank to be buried below the near cortex [16]. The short-threaded pin is favored by the authors for all fixators (if possible), because by placing the shank below the cortical surface it doubles the pin's stiffness [9] and results in less irritation to the surrounding soft tissue.

Clamps

The clamp's function is to connect the pin to the rod (bar). These are classified as simple, which connect one pin to one bar, or modular, which connects several pins in a cluster to up to two bars. Modular clamps distribute the pins symmetrically and provide the best pin fixation strength [17]. However, modular clamps can produce uneven holding strength on multiple pins, a result that is avoided with the use of simple clamps. Some clamps have a ball-and-socket type joint that allows more degrees of freedom during the frame construction, and some systems have the clamp and rod form a single unit with the clamp attached to the rod through a universal ball joint. There are also clamps designed to connect two bars together to allow angular placement of the bars across a joint (e.g., knee) or increase in the length of the fixation when used to stabilize an entire extremity (Fig. 5.1).

Rods

The rods (connecting bars) allow construction of the external fixator frame. They are made of stainless steel, aluminum alloy, or carbon fiber and can have circular, elliptical, square, or multiple-faced cross-sectional geometries. Carbon fiber rods are radiolucent connectors that allow greater visualization of the bony reductions during placement with fluoroscopy and in subsequent radiographs. Although they are more expensive than stainless steel or aluminum alloy, testing has shown that they are 15% stiffer in loading to failure than stainless steel tubes [18]. Whereas stainless steel deforms at 50% of its maximum load, carbon fiber rods maintain the same stiffness until failure. However, carbon fiber rods are only 85% as stiff as a frame constructed using stainless steel. This is attributed to limitations in the ability of the clamp to tighten onto the carbon fiber rods versus tightening using stainless steel rods. If greater strength (stiffness) of a uniplanar frame is needed, it can be obtained by adding a second plane of fixation (biplane), by stacking a second rod on top of the already constructed fixation (favored by the authors), or by placing rods closer to the bone [8].

Anatomic Considerations

Prior to placing the pins, consideration must be given to the relevant soft tissue anatomy, the impact on the patient's physiology and care, and how the placement of these pins and the frame will affect definitive fixation. This also includes thinking about the neurovascular and musculotendinous structures that are at risk during and after the insertion of the pins. Consequently, one must have a good understanding of the cross-sectional anatomy of the limb, particularly at the

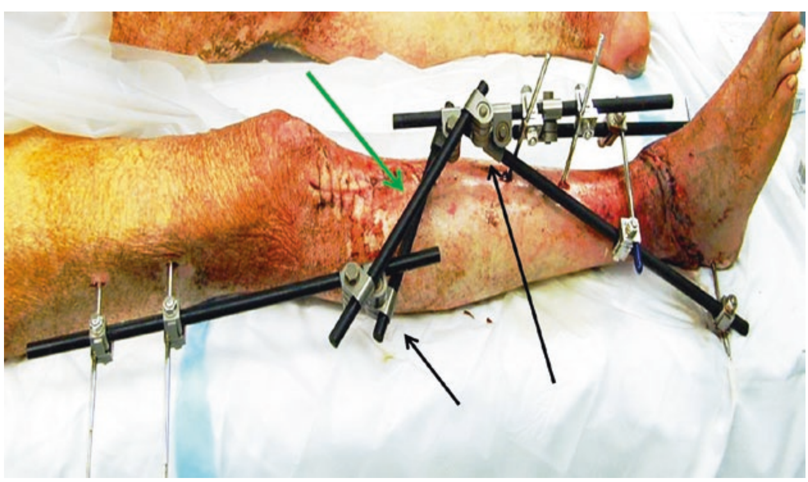


Fig. 5.1 A 39-year-old male with a proximal tibia fracture who initially managed with provisional (spanning) fixation. Note the bar-to-bar clamps (*black arrows*) that allowed for a greater length to the frame to be constructed with intercalary rods (*green arrow*)

level of pin insertion. To aid that comprehension, Behrens divided the bones of the limb into eccentric and concentric types [7]. Eccentric bones have a subcutaneous border (e.g., tibia, ulna, pelvis, metacarpals) and rarely produce any pin tract complications. Concentric bones are centrally located in the limb, are surrounded by the muscle (e.g., humerus, radius, femur), and require a deeper dissection for pin placement; and here there is greater potential for muscle damage, joint stiffness, neurovascular injuries, or pin complications. He then described three distinct areas for pin insertion in each bone as *safe, hazardous, and unsafe* corridors [7] (Table 5.1).

Safe corridors do not contain musculotendinous entities or any important neurovascular structures. They occur only in eccentric limbs and have a low incidence of pin tract infections or loosening. *Hazardous corridors* contain musculotendinous elements but no important neurovascular structures. Pins in these corridors may induce a compartment syndrome, cause permanent joint stiffness by tethering tendons or ligaments, and often result in higher rates of pin tract complications. In areas devoid of a safe corridor, this is the safest region for pin insertion. *Unsafe corridors* contain both musculotendinous and important neurovascular structures. It has the highest rate of pin tract-related complications and the greatest potential for nerve and vascular injuries. If a pin is placed in this corridor, it is essential to use an exposure that allows for open pin placement and protection of the soft tissues (Table 5.2).

The width of the safe and hazardous corridors rarely exceeds 90–140° [6, 7]. This means that transfixion (centrally threaded) pins are usually contraindicated. An exception can be made if the safe corridor exceeds 180° (e.g., the proximal tibia) or for circumstances where the advantages of pin placement exceed its potential risks (e.g., the calcaneus). It is also important to remember that even if with thin wire fixation (1.5, 2 mm pins), the dangers associated with traversing a hazardous or an unsafe zone are still present. This is why some patients managed with skinny wire fixation have excessive pain,

Table 5.1 The divisions of bones and how external fixation can be applied to eccentric and concentric segments

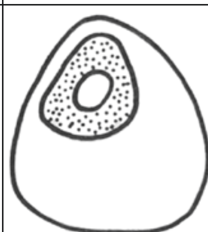
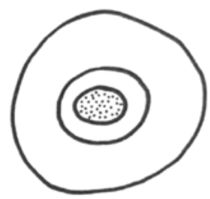
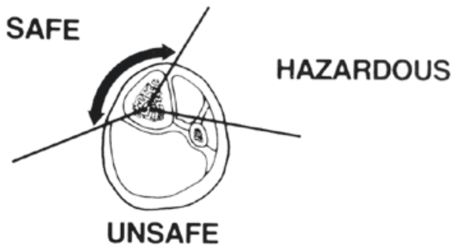
Limb segment	Eccentric	Concentric
• One segment		
Upper extremity	Scapula Ulna Metacarpals	Humerus Radius Phalanges
Lower extremity	Pelvis Tibia Metatarsals	Femur Fibula Phalanges
• Transarticular multisegmental	•←	→•
Complications	Rare	Joint stiffness Pin problems Neurovascular injuries
Frame application	Long-term	Short-term ► Internal fixation ► Braces

Table 5.2 Table showing the safe, hazardous, and unsafe corridors for pin placement
Dots represent intensity of finding:
• = present,
•• = more severe,
••• = most severe,
(•) = inconsistent



Corridor	Safe	Hazardous	Unsafe
Limb Segment			
• Eccentric	•	•	•
• Concentric		•	•
Pin insertion			
• Stab Wounds	•	•	
• Open Exposure		(•)	• • •
Soft tissues			
• Skin		•	•
• MT Units	•	•	•
• NV Structures			•
Complications			
• Pin Tracts	•	• •	• • •
• Compartments		•	•
• Joint Stiffness		(•) •	(•) •
• NV Injuries			• • •

reduced motion in distal articulations, and may sustain major and persistent neurovascular problems.

Indications for Temporary External Fixation

Temporary external fixation can be used for "damage control" or to simply act as a provisional method of fixation. For this discussion, the former is defined as external fixation applied to a polytrauma patient presenting with life- or limb-threatening conditions that need emergent skeletal stabilization. The latter is applied to a trauma patient presenting with an unstable bony injury that requires restoration of length and axial alignment with the fixator applied in a semi-elective (non-emergent) setting.

Damage Control Orthopedics

Polytrauma patients were initially viewed as too ill to tolerate the stress associated with any skeletal fixation. This is because the mortality of polytrauma patients has been identified as bimodal. This pathophysiologic concept is described as a two-hit model where death can occur almost immediately after the trauma or within a few days after the trauma has occurred [19]. The first hit begins after the initial trauma and generates a major inflammatory cascade producing hypoxemia (acute respiratory distress syndrome, or ARDS), hypotension, and multiple organ failure. Independent to the damage associated with fractures, it often leads to the patient's early death. The second hit begins after the first hit and generates a less severe inflammatory model, producing ischemic or reperfusion problems, the development of compartment syndromes, or infections and sepsis. If not adequately addressed, it leads to later deaths. With the two-hit model, it became evident that early, extensive skeletal fixation was not always desirable or recommended because it leads to greater physiologic insults and more complications [20, 21].

This leads to the concept of *damage control orthopedics* [22]. Scalea et al. credited this expression to the naval war term of *damage control* as it applied to the initial, limited treatment of managing exsanguinating, penetrating abdominal trauma [23]. The tenets include recognizing who needs damage control, doing only absolutely necessary operations, keeping the patient warm and alive, accepting the morbidity of the operative procedures, and providing definitive care at a later date. Patients who need it are those described as at risk for loss of life or limb. The absolutely necessary operations include limited open fracture debridement, performing fasciotomies, stabilization of long bone fractures and the pelvic ring, and deciding on limb salvage versus an amputation [24]. In addition to unstable pelvic ring injuries and long bone fractures, damage control orthopedics can also be considered for intra-articular fractures of the knee, for multiple open fractures, for patients with extensive burns, and for other joint injuries/dislocations.

Researchers have found that delaying major fracture surgery for a few days in unstable polytrauma patients avoided the second hit phenomenon, had a protective effect on the inflammatory response, and decreased pulmonary and hepatic dysfunction [21, 22, 25]. As a result, current recommendations are that damage control orthopedics be considered as the primary method for managing physiologically unstable/under-resuscitated polytrauma patients (Fig. 5.2).

Provisional Fixation

The indications for provisional fixation include patients with displaced peri- or intra-articular fractures, fracture-dislocations of the upper and

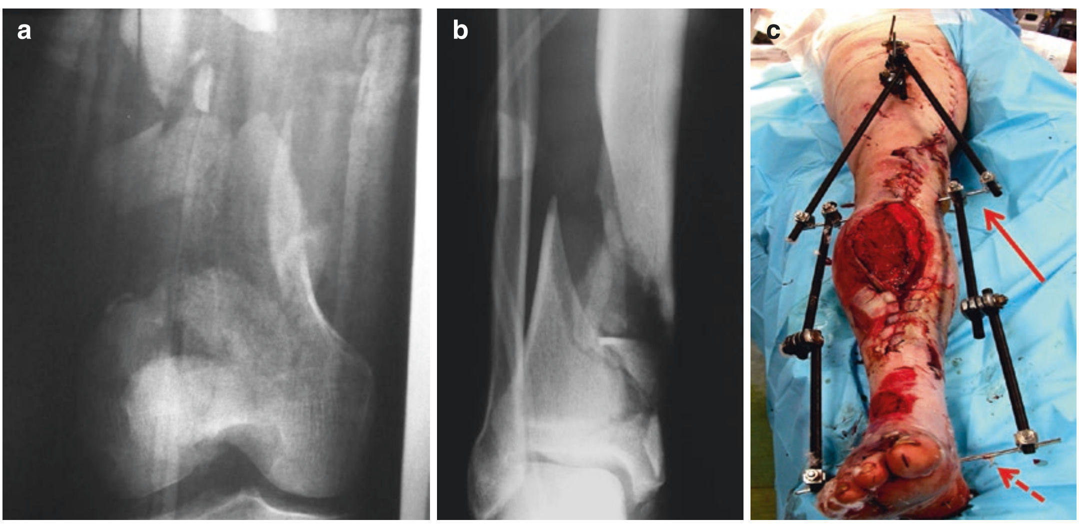


Fig. 5.2 (**a**) Anteroposterior radiograph of an open, comminuted distal femur fracture presenting in a 26-year-old polytrauma patient with a concurrent liver laceration, pulmonary contusion, and a closed head injury. (**b**) Anteroposterior radiograph demonstrating the patient having also sustained an open, comminuted, ipsilateral distal tibia fracture. (**c**) Anterior view demonstrating damage control orthopedics in which a bilateral, uniplanar frame with transfixion pins through the proximal tibia (*solid red arrow*) and through the calcaneus (*dotted red arrow*) was connected to an anterior in the proximal femur frame

lower extremity, or displaced diaphyseal fractures that are unable to undergo early definitive fixation. The term *spanning external fixation* may be a better description of this fixation, indicating that this spanning will span the zone of injury. The goals of provisional fixation are to restore length, improve alignment of the limb and fracture, and protect the surrounding soft tissues. This has been shown to decrease pain of the extremity, to avoid prolonged pressure that can lead to skin necrosis and irreversible damage of the soft tissues, and to correct the deformation of the blood vessels that occurs with displaced fractures, improving the circulation of the extremity [26–29].

The advantages of provisional fixation include placing the fixation away from the zone of injury, precluding interference with future incisions, providing length to surrounding soft tissues to help realign the fracture fragments (ligamentotaxis), allowing visualization of the soft tissues, and permitting patients to be mobile rather than bedridden (Fig. 5.3).

Construction of External Fixator Frames

External fixation has been used for trauma involving the pelvis, the upper and lower extremities, the spine [30], and the thorax [31]. Its versatility, reproducibility, and its effectiveness in maintaining reductions have contributed to making it an extremely valuable tool. The fixation should be a simple construct with the best locations determined by the fracture location(s), its relationship to the proximity of safe or hazardous corridors, and its effect on future definitive fixation.

Pin Insertion Technique

Regardless of whether one plans to use the external fixation for damage control or for provisional management, attention should first be directed toward the proper method of pin insertion. This is the most critical step and determines the overall

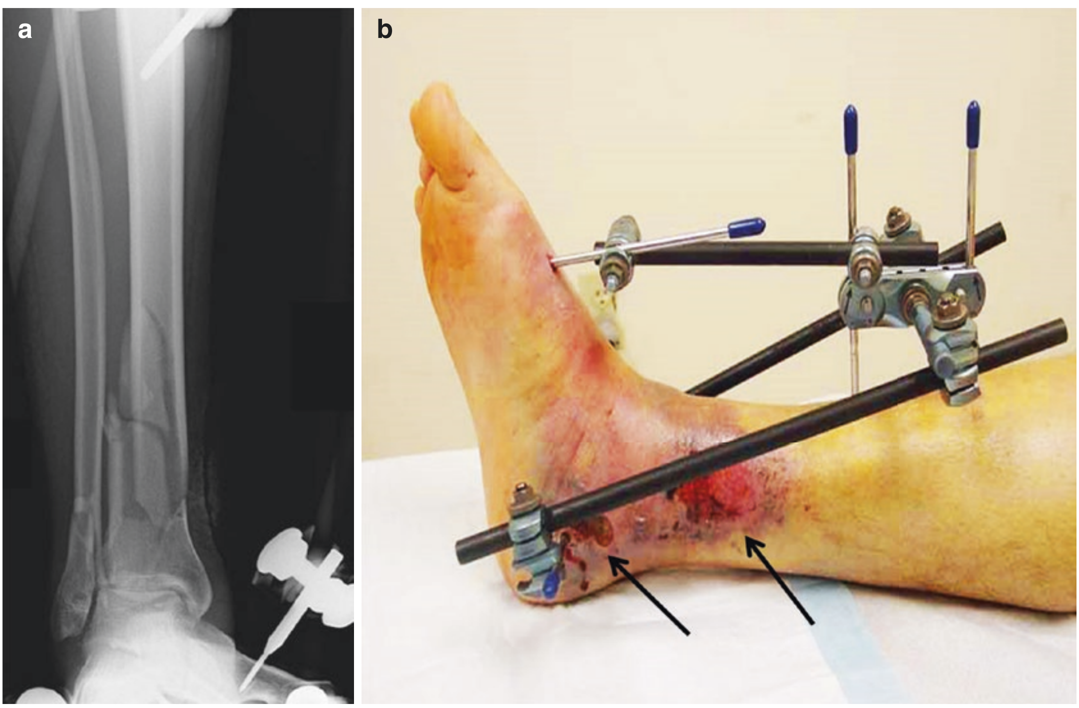


Fig. 5.3 (**a**) Mortise view of a comminuted pilon fracture aligned using provisional fixation. (**b**) The provisional fixation allowed fracture blisters to be seen and treated until re-epithelialization of the skin occurred (*black arrows*)

success, because using the correct technique improves pin torque resistance and minimizes loosening.

The approach begins with a generous skin incision followed by blunt dissection to the periosteum, which is then reflected off the bone. To minimize soft tissue damage, a cannula is placed directly onto the bone and is kept in position until after pin insertion. A bicortical pilot hole is predrilled, and the pin is inserted by hand to the correct depth. One should avoid wobbling during pin insertion as this can produce a small but significant conical deformation of the near cortex, reducing stability of the pin at the near cortex and increasing stress at the far cortex [32]. To avoid this, the authors frequently use battery-powered drills at low speeds to insert the pin until it reaches the opposite cortex and then complete the insertion by hand.

The use of self-drilling or self-tapping pins, although touted as easier and faster, can generate temperatures that can easily exceed 194 °F. This produces bony necrosis with irreversible osteocyte death and alkaline phosphate inactivation [15]. Self-drilling pins have also been shown to produce microfractures of both cortices, resulting in bony resorption and decreased pullout strengths [16]. Current modifications—incorporating drill points and flutes with a modified thread pitch that avoids stripping the near cortex when drilling the far cortex—have still demonstrated a reduction in bony purchase of 22% compared with a predrilled technique [33]. In addition, critics complain of the difficulty in feeling the tip as it penetrates through the endosteal surface on the far cortex [32]. This leads to a tendency to increase the depth of pin insertion, resulting in soft tissue invagination (i.e., pulling the surrounding tissue into the pin tract) [16, 33]. However, in life- or limb-threatening situations, and with a limited amount of time given to stabilize the patient, the authors do offer a disclaimer that even with these problems the use of self-drilling pins may be acceptable in order to provide fixation and gain rapid bony stability.

Management of Diaphyseal Fractures

In the 1970s and 1980s, external fixation was used for the definitive management of diaphyseal injuries. These frames were in position for 3–4 consecutive months, and descriptions were offered on how to increase frame stiffness and how to make them less rigid (dynamization), allowing motion at the fracture site and resulting in secondary healing by callus formation [8].

In contrast, the aim of temporary fixation of diaphyseal fractures is to hold the fracture and thus the limb in an acceptable alignment until definitive fixation is undertaken. There are some principles that can be useful for the management of *all* diaphyseal injuries.

1. The subcutaneous (eccentric) border of a bone should be used for the placement of the half-pins. In concentric bones, a generous incision with blunt dissection should be performed before pin placement.
2. Place the muscle compartment, through which the pin is passing, on stretch during insertion of the pin. Doing so maintains motion around the adjacent joints and minimizes irritation by the pin.
3. Avoid placing pins into a physis, the articular surface of a joint, or in close proximity to the articular surfaces. This is especially important around the knee joint. Studies have identified the capsule of the distal femur extending an average of 7 cm proximal to the center of anterior part of the notch and extending to an average of less than 6 mm distal to the anterior articular surface of the proximal tibia [34, 35].
4. The length of the frame should encompass almost the entire length of the long bone and should be constructed of the simplest frame possible.
5. If possible, the rods should be positioned at least a hand's width away from the surface of the skin to accommodate swelling and to allow visualization of the soft tissue structures.

6. If more stability is needed, add a second bar (stacked) to the constructed unilateral frame, or one can achieve additional stability with the use of a posterior splint.
7. Evaluate fixation location, and determine whether it can be left in position to help with (or whether it will interfere with) the reduction of the fracture during definitive fixation (Fig. 5.4).
8. It is unknown how long external fixation can safely remain in place before there is an increase in the risk of infection following definitive fixation [36]. Therefore, conversion to definitive fixation is recommended as soon as the patient and surrounding soft tissues are deemed stable.

To stabilize the diaphyseal fracture and increase the stiffness of the fixator-bone construct, a minimum of four half-pins with a unilateral frame or two transfixion pins with a bilateral (single plane) frame should be used. Other alterations that can increase frame stiffness include an anatomic reduction of the fracture, best seen in Winquist type I–III fracture patterns [37], and when interfragmentary compression is obtained through the frame [8]. For patients deemed too unstable, more pins can be added at a later date, or the implants for damage control can include placement of the entire fixator in the intensive care unit [32].

Fixator configurations are dependent on the long bone being addressed. In the *humerus* pins should be placed anterolaterally in the proximal humerus, to avoid injuring the axillary and radial nerves, and posterolateral in the distal humerus, avoiding the olecranon fossa and radial nerve. In the *forearm* the best pin placement uses smaller diameter pins (3–4 mm) through the subcutaneous border of the ulna. If stabilization of the radius

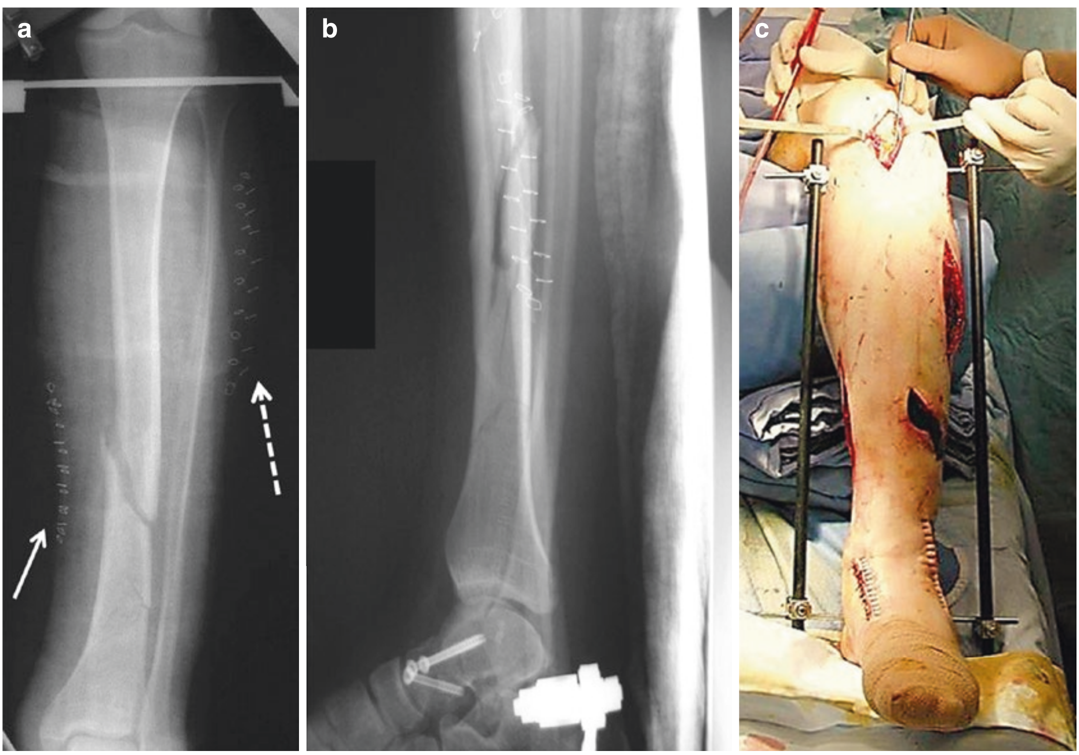


Fig. 5.4 (**a**) Anteroposterior radiograph demonstrating an open (*solid white arrow*) tibial shaft fracture treated with a bilateral, uniplanar frame and fasciotomy (*white dotted line*). (**b**) Lateral view demonstrating an associated talar neck fracture, which is why the transfixion pin was placed into the calcaneus. (**c**) The fixator was left in position during intramedullary nailing of the tibia to maintain the reduction of the fracture

is needed, an open pin placement is recommended to avoid damaging the posterior interosseous nerve proximally and the superficial radial nerve distally. For the *femur* pin placement should pass through relatively safe corridors, including direct anterior, direct lateral, or anterolateral. In the *tibia* the anteromedial subcutaneous surface is best suited to place pins perpendicular to either the anteromedial or posterior faces of the tibial cortex. In the distal-fifth, pins should be placed using an open technique to avoid injuring the anterior tibial vessels and the deep peroneal nerve.

Frame construction can be achieved using a single rod placed between the two pins in each fragment. This creates a base for the fixation and allows manipulation of the individual fracture segments, which is stabilized using a third rod (intercalary rod) connecting the two bases. An alternative, and one that the authors favor, is manipulating the fracture fragments into better position using the pins and applying a long rod that connects to all of the pins. It should be remembered that temporary fixation is not intended to achieve an anatomic reduction but rather to stabilize and improve the alignment of the fracture and limb until definitive fixation can be achieved.

Management of Intra-articular Fractures

Management goals are to improve the overall alignment of the extremity, to avoid prolonged pressure to the skin and soft tissues, to allow ligamentotaxis to help reduce some of the fracture fragments, and to improve circulation to the extremity [26–28].

Obtaining these goals requires that a spanning fixator be placed across the joint. The lone exception is in pelvic injuries where the fixator is used to decrease intra-pelvic volume and stabilize the associated bony/ligamentous injuries. In the *pelvis*, pins can be placed along the iliac crest beginning 2 cm posterior to the anterior superior iliac spine (ASIS) and directed posteromedially between the inner and outer tables of the gluteal pillar of the pelvis. If a second pin is needed, it is placed more posterior on the crest. Care should be taken to avoid penetration of the outer cortex, which may lead to pin cutout. An alternative is an open placement of supra-acetabular pins. This construct will provide better control of the hemipelvis [38] than a frame with iliac crest pins and consists of placing a pin through each anterior inferior iliac spine and directing them posteriorly and medially along the supra-acetabular region of the pelvis (Fig. 5.5).

The *elbow* is stabilized by placing pins into the posterolateral distal humeral and the subcutaneous border of the ulna. For the distal *radius,* an open technique is used to protect the superficial radial nerve by placing 3–4 mm pins through the radius, posterior to the radial artery. The authors' preferred location on the radius is on the middle third of the radial border of the radius. Three mm pins are then inserted into the dorsoradial border of the second metacarpal, using an open technique to protect the superficial radial nerve and to sharply elevate the dorsal interosseous muscle off the bone. Finger flexion problems, produced by overdistraction, are avoided by placing the wrist in a neutral position at the end of the procedure.

For injuries around the *knee*, pins can be placed laterally into the femur and connected to pins placed anteriorly in the tibia, or the entire frame can be placed anteriorly (Fig. 5.6). In both applications, a single long rod is connected to

Fig. 5.5 Anteroposterior view of a pelvic ring injury managed with a supra-acetabular placed external fixator

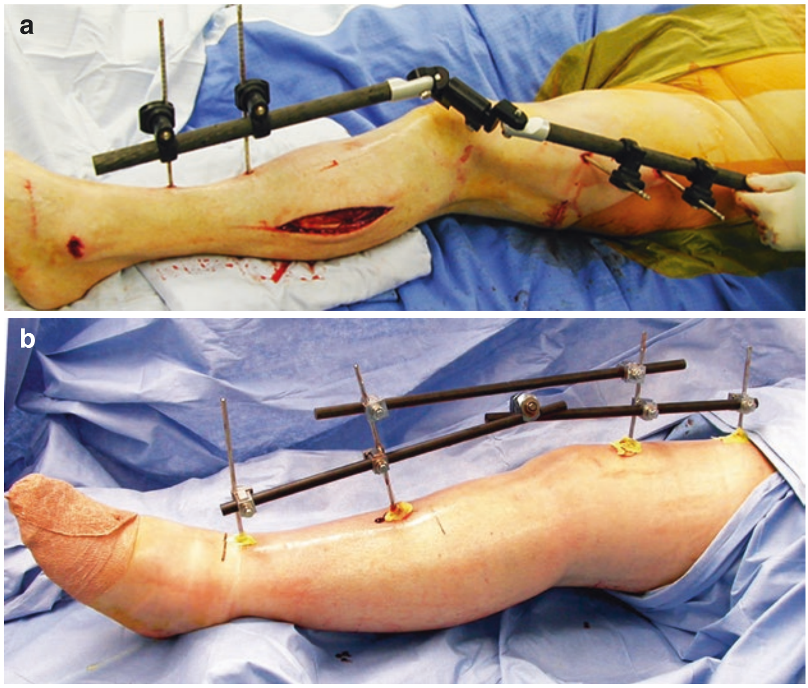


Fig. 5.6 (**a**) Provisional fixation of a proximal tibia fracture constructed with a lateral pin in the femur connected to an anterior pin in the tibia. (**b**) Patient with a comminuted proximal tibia initially managed with the entire frame placed anteriorly

each pair of pins with adequate length needed to connect to the other rod. After application, the frame as well as the knee should be placed in approximately 5–15° flexion. For *ankle* injuries, the configuration most commonly used involves the construction of a delta frame (Fig. 5.7). Pins are placed anteriorly or anteromedially in the tibia and are connected to a transfixion (centrally threaded) pin placed through the calcaneus. In the calcaneus, this is a hazardous corridor. The medial open pin insertion is placed halfway between the posteroinferior calcaneus and the inferior edge of the medial malleolus and posterior to a line drawn from the navicular tuberosity and the posteroinferior calcaneus (Fig. 5.8) [39]. As previously described, a posterior splint can be used to provide additional stability for all intra-articular fractures.

Complications and Pin Care

Localized pin tract infection is the most common complication associated with external fixation and is why some surgeons avoid its use. The incidence has been reported to occur in 0–100% of patients [40]. Pin colonization can occur immediately after insertion when membrane proteins and polysaccharides allow the bacteria to bind to the pin surface. With enough bacteria, a colony forms and secretes a protective biofilm rendering it resistant to antibiotics [41] (Fig. 5.9). A pin-site infection is defined as the presence of any signs or symptoms of infection around a pin that requires treatment with antibiotics, pin removal, or debridement [42]. Factors leading to pin problems include pin sites with large soft tissue sleeves (muscle), pin sites with motion, skin tension or irritation around a pin site, and prolonged duration of pin fixation. These result in local inflammation, leading to pin tract infections and possible osteomyelitis. To guide the management of pin site problems, two classification systems have been described [43, 44].

Pin loosening is also a common complication, resulting in failure of fixation and a loss of the reduction. Causes include thermonecrosis during insertion, excess stress at the pin-bone interface, or the development of a pin-site infection. Strategies to reduce stress at the pin-bone interface have already been discussed, but if a pin is found loose, it should be removed and replaced.

Fig. 5.7 (**a**) Anteroposterior view of a pilon fracture demonstrating improved alignment using a delta frame. Note the absence of the frame in the radiograph. (**b**) Anterior view demonstrating the delta frame configuration with pins placed into the proximal tibia and a transfixion pin placed through the calcaneus. (**c**) Lateral view of the delta frame

For pin care, an adequate release of the skin and stabilizing the soft tissues around the pin to prevent motion appear to be more important than the method or agent used to cleanse the pin. The former can include wrapping a bolster or sponge around the pin to stabilize the soft tissue, which prevents motion during activity or with mobilization of the extremity [7, 32] (Fig. 5.10). Upon discharge, patients are provided with information on how to care for the fixator and pins.

Methods of pin care range from doing nothing to washing the pin site three times per day using peroxide, alcohol, or other cleaning solutions. In two Cochrane systematic reviews and meta-

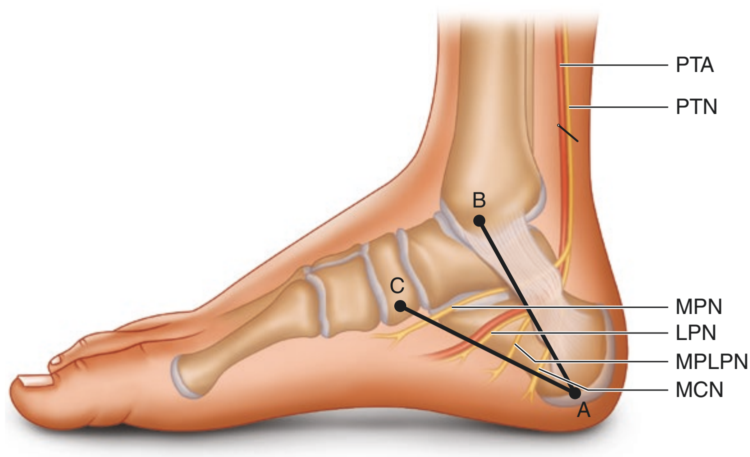


Fig. 5.8 Illustration demonstrating the neurovascular structures at risk during medial calcaneal pin placement. Note the small window for safe pin placement. A = posteroinferior calcaneus; B = inferior edge medial malleolus; C = navicular tuberosity; PTA = posterior tibial artery; PTN = posterior tibial nerve; MPN = medial plantar nerve; LPN = lateral plantar nerve; MCN = medial plantar nerve; MPLPN = most posterior lateral plantar nerve (From Casey et al. [39], with permission Wolters Kluwer)

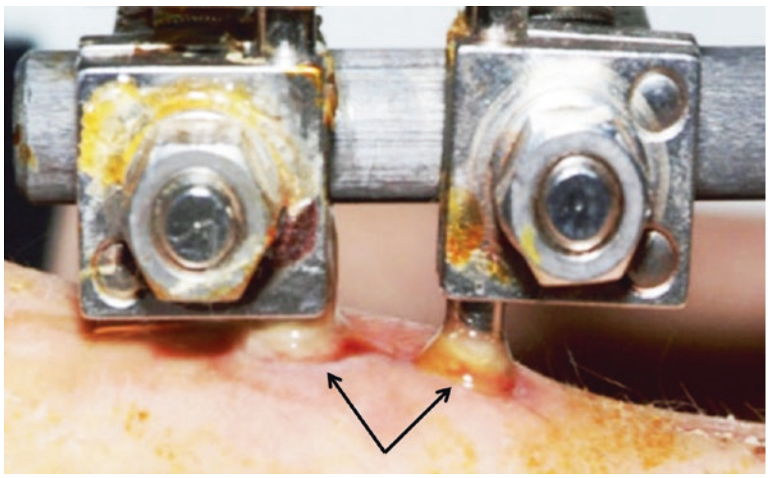


Fig. 5.9 Magnified image of external fixator demonstrating erythema and purulent drainage (*black arrows*) from multiple pin sites

Fig. 5.10 Soft tissue bolster applied to decrease motion of the skin at the proximal end of a delta frame

analysis studies comparing cleaning solutions versus no cleansing regimen, Lethaby et al. demonstrated no significant differences regarding inflammation, rates of infection, or rates of pin loosening [45, 46]. They concluded that there was insufficient evidence to promote any recommendations addressing cleansing strategies or the frequency with which pins should be cleaned in order to reduce the risk of infection.

Given this information, the authors' preference, after the index surgical dressing has been removed, is to allow the patient to shower daily and to instruct them to clean the pins and fixator with soap and water as part of their daily hygiene. Patients are told to avoid soaking the pins.

We recommend leaving the pins uncovered, but if they are to venture out of the house, we instruct them to cover the pin sites with clean dressings or bolsters if necessary.

Conclusion

In summary, temporary external fixation for the management of diaphyseal and intra-articular fractures is an extremely valuable tool for the care of trauma patients. Use of a simple uniplanar fixator will allow for quick application and greater access to the surrounding soft tissues and wounds. It is important to have a good understanding of the cross-sectional anatomy of the limb, to consider the three anatomic corridors that can be used for pin placement, and to use good technique for pin insertion. If more stability is needed, it can be increased with the addition of a second (intercalary) bar or by adding a posterior splint to the extremity. Finally, there is insufficient evidence to promote any recommendations regarding the optimal frequency for pin site care or any specific recommendations addressing cleansing strategies of the pin site, in order to reduce the risk of infection.

References

1. Malgaigne JF. Considérations cliniques sur les fractures de la rotule et leur traitment par les griffes. J des Connnaissances Med Pratiques. 1853;16:9.
2. Johnson HF, Stovall SL. External fixation of fractures. J Bone Joint Surg Am. 1950;32A(2):466–7.
3. Brooker A, Edwards C. External fixation: the current state of the art. Baltimore: Williams and Wilkins; 1979.
4. Mears DC. External skeletal fixation. Baltimore: Williams and Wilkins; 1981.
5. Behrens F, Jones RE 3rd, Fischer DA, Mears DC. External skeletal fixation. Instr Course Lect. 1981;30:112–82.
6. Behrens F. A primer of fixator devices and configurations. Clin Orthop Relat Res. 1989;(241):5–14.
7. Behrens F. General theory and principles of external fixation. Clin Orthop Relat Res. 1989;(241):15–23.
8. Behrens F, Johnson W. Unilateral external fixation. Methods to increase and reduce frame stiffness. Clin Orthop Relat Res. 1989;(241):48–56.
9. Behrens F, Johnson WD, Koch TW, Kovacevic N. Bending stiffness of unilateral and bilateral fixator frames. Clin Orthop Relat Res. 1983;(178):103–10.
10. Holt J, Hertzberg B, Weinhold P, Storm W, Schoenfisch M, Dahners L. Decreasing bacterial colonization of external fixation pins through nitric oxide release coatings. J Orthop Trauma. 2011;25(7):432–7.
11. Jennison T, McNally M, Pandit H. Prevention of infection in external fixator pin sites. Acta Biomater. 2014;10(2):595–603.
12. Gil D, Shuvaev S, Frank-Kamenetskii A, Reukov V, Gross C, Vertegel A. Novel antibacterial coating on orthopedic wires to eliminate pin tract infections. Antimicrob Agents Chemother. 2017;61(7):e00442–17. https://doi.org/10.1128/AAC.00442-17.
13. Slate AJ, Wickens DJ, El Mohtadi M, Dempsey-Hibbet N, West G, Banks CE, Whitehead KA. Antimicrobial activity of Ti-ZrN/Ag coatings for use in biomaterial applications. Sci Rep. 2018; 8:1497. https://doi.org/10.1038/s41598-018-20013-z . Author correction: Sci Rep. 2018;8(1):6251. https://doi.org/10.1038/s41598-018-24174-9.
14. Toksvig-Larsen S, Aspenberg P. Bisphosphonate-coated external fixator pins appear similar to hydroxyapatite-coated pins in the tibial metaphysis and to uncoated pins in the shaft. Acta Orthop. 2013;84(3):314–8.
15. Eriksson A, Albrektsson T, Grane B, McQueen D. Thermal injury to bone: a vital microscopic description of heat effects. Int J Oral Surg. 1982;11(2):115–21.
16. Moroni A, Vannini F, Mosca M, Giannini S. Techniques to avoid pin loosening and infection in external fixation. J Orthop Trauma. 2002; 16(3):189–95.
17. Aro HT, Hein TJ, Chao EY. Mechanical performance of pin clamps in external fixators. Clin Orthop Relat Res. 1989;(248):246–53.
18. Kowalski M, Schemitsch EH, Harrington RM, Chapman JR, Swiontkowski MF. Comparative biomechanical evaluation of different external fixation sidebars: stainless- steel tubes versus carbon fiber rods. J Orthop Trauma. 1996;10(7):470–5.
19. Faist E, Baue AE, Dittmer H, Heberer G. Multiple organ failure in polytrauma patients. J Trauma. 1983;23(9):775–87.
20. Pape HC, Rixen D, Morley J, Husebye EE, Mueller M, Dumont C, et al.; EPOFF Study Group. Impact of the method of initial stabilization for femoral shaft fractures in patients with multiple injuries at risk for complications (borderline patients). Ann Surg. 2007;246(3):491–9; discussion 499–501.
21. Morshed S, Miclau T III, Benbom O, Cohen M, Knudson MM, Colford JM Jr. Delayed internal fixation of femoral shaft fracture reduces mortality among patients with multisystem trauma. J Bone Joint Surg Am. 2009;91(1):3–13.
22. Scalea TM, Boswell SA, Scott JD, Mitchell KA, Kramer ME, Pollak AN. External fixation as a bridge

to intramedullary nailing for patients with multiple injuries and with femur fractures: damage control orthopedics. J Trauma. 2000;48(4):613–21; discussion 621–3.
23. Rotondo MF, Schwab CW, McGonigal MD, Phillips GR 3rd, Fruchterman TM, Kauder DR, et al. ‘Damage control’: an approach for improved survival in exsanguinating penetrating abdominal injury. J Trauma. 1993;35(3):375–82.
24. Tscherne H, Regel G, Pape HC, Pohlemann T, Krettek C. Internal fixation of multiple fractures in patients with polytrauma. Clin Orthop Relat Res. 1998;(347):62–78.
25. Tuttle MS, Smith WR, Williams AE, Agudelo JF, Hartshorn CJ, Moore EE, Morgan SJ. Safety and efficacy of damage control external fixation versus definitive stabilization for femoral shaft fractures in the multiple-injured patient. J Trauma. 2009;67(3):602–5.
26. Haidukewych GJ. Temporary external fixation for the management of complex intra- and periarticular fractures of the lower extremity. J Orthop Trauma. 2002;16(9):678–85.
27. Sirkin M, Sanders R, DiPasquale T, Herscovici D Jr. A staged protocol for soft tissue management in the treatment of complex pilon fractures. J Orthop Trauma. 1999;13(2):78–84.
28. Egol KA, Tejwani NC, Capla EL, Wolinsky PL, Koval KJ. Staged management of high-energy proximal tibia fractures (OTA Types 41). J Orthop Trauma. 2005;19(7):448–55.
29. Min W, Ding BC, Tejwani NC. Staged versus acute definitive management of open distal humeral fractures. J Trauma. 2011;71(4):944–7.
30. Wang W, Yao N, Song X, Yan Y, Wang C. External spinal skeletal fixation combination with percutaneous injury vertebra bone grafting in the treatment of thoracolumbar fractures. Spine (Phila Pa 1976). 2011;36(9):E606–11.
31. Henley MB, Peter RE, Benirschke SK, Ashbaugh D. External fixation of the sternum for thoracic trauma. J Orthop Trauma. 1991;5(4):493–7.
32. Ziran BH, Smith WR, Anglen JO, Tornetta P 3rd. External fixation: how to make it work. J Bone Joint Surg Am. 2007;89(7):1620–32.
33. Seitz WH Jr, Fromison AI, Brooks DB, Postak P, Polando G, Greenwald AS. External fixator pin insertion techniques; biomechanical analysis and clinical relevance. J Hand Surg Am. 1991;16(3):560–3.
34. Lowery K, Dearden P, Sherman K, Mahadevan V, Sharma H. Cadaveric analysis of capsular attachments of the distal femur related to pin and wire placement. Injury. 2015;46(6):970–4.
35. DeCoster TA, Crawford MK, Kraut MA. Safe extracapsular placement of proximal tibial transfixation pins. J Orthop Trauma. 2004;18(Suppl 8):43–7.
36. Nowotarski PJ, Turen CH, Brumback RJ, Scarboro JM. Conversion of external fixation to intramedullary nailing for fractures of the shaft of the femur in multiply injured patients. J Bone Joint Surg Am. 2000;82(6):781–8.
37. Winquist R, Hansen S. Comminuted fractures of the femoral shaft treated by intramedullary nailing. Orthop Clin N Am. 1980;11(3):633–47.
38. Stahel PF, Mauffrey C, Smith WR, McKean J, Hao J, Burlew CC, Moore EE. External fixation for acute pelvic ring injuries: decision making and technical options. J Trauma Acute Care Surg. 2013;75(5):882–7.
39. Casey D, McConnell T, Parekh S, Tornetta P 3rd. Percutaneous pin placement in the medial calcaneus: is anywhere safe? J Orthop Trauma. 2004;18(8 Suppl):S39–42.
40. Kazmers NH, Fragomen AT, Rozbruch SR. Prevention of pin site infection in external fixation: a review of the literature. Strategies Trauma Limb Reconstr. 2016;11(2):75–85.
41. Parameswaran AD, Roberts CS, Seligson D, Voor M. Pin tract infection with contemporary external fixation: how much of a problem? J Orthop Trauma. 2003;17(7):503–7.
42. Ceroni D, Grumetz C, Desvachez O, Pusateri S, Dunand P, Samara E. From prevention of pin-tract infection to treatment of osteomyelitis during paediatric external fixation. J Child Orthop. 2016;10:605–12.
43. Dahl MT, Gulli B, Berg T. Complications of limb lengthening. A learning curve. Clin Orthop Relat Res. 1994;(301):10–8.
44. Checketts RG, MacEachem AG, Otterbum M. Pin tract infection and the principles of pin site care. In: DeBastiani G, Apley AG, Goldberg AA, editors. Orthofix external fixation in trauma and orthopaedics. London: Springer-Verlag London Ltd; 2000. p. 97–103.
45. Lethaby A, Temple J, Santy J. Pin site care for preventing infections associated with external bone fixators and pins. Cochrane Database Syst Rev. 2008;(4):CD004551. https://doi.org/10.1002/14651858.CD004551.pub2.
46. Lethaby A, Temple J, Santy-Tomlinson J. Pin site care for preventing infections associated with external bone fixators and pins. Cochrane Database Syst Rev. 2013;(12):CD004551. https://doi.org/10.1002/14651858.CD004551.pub3.

6 Periarticular Fractures

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Introduction

External fixation as temporary or definitive treatment of periarticular fractures has been regularly utilized for less than three decades [1–6]. During this time, popularity of these techniques has varied based on surgeon preferences, current literature, and the availability and safety of alternative techniques of internal fixation. External fixation is generally considered an optimal technique to minimize soft tissue complications. It frequently has been paired in philosophy and technique with less extensile or percutaneous approaches for reducing and fixing periarticular fractures. Compared to internal fixation, definitive treatment of periarticular fractures with external fixation is more time-consuming and labor intensive for the surgeon in the postoperative period. External fixation is also more difficult for the patient who wears and maintains the frame as an outpatient. There are issues of cost-effectiveness, particularly when expensive frames are briefly used for temporary stabilization prior to definitive internal fixation. In addition, the effectiveness of external fixation, particularly when used as definitive treatment, remains a question [7].

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In current practice, external fixation has several roles in the treatment of periarticular fractures. These roles are joint specific, depending on the local soft tissues, other available treatment options, and the local anatomy. External fixators are used in periarticular fractures either for temporary or definitive treatment. The biomechanical principles that are present depend on the goals of the treatment. This chapter will focus on each of these indications separately.

External Fixation for Temporary Stabilization

External fixation has an important role in *temporary treatment* of periarticular fractures [3, 8]. The purpose is to restore the length and gross alignment, provide some degree of joint stability, protect injured soft tissues, and mobilize the patient while awaiting definitive treatment of the fracture. The time between temporary external fixation and definitive treatment can be used to (1) improve the general condition of the patient, (2) monitor recovery of local soft tissue injury, (3) arrange for optimal resources and or elective surgical time, and (4) coordinate definitive fixation and soft tissue coverage. When used for temporary treatment, external fixation almost always spans the joint and the injured articular fragments. Intact soft tissues on displaced articular fragments allow for ligamentotaxis to reduce the

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articular surface. The fixator is usually removed at the time of definitive treatment days or weeks after it was applied.

External Fixation for Definitive Treatment

In some joints, external fixation can be used as the major stabilizing device for *definitive treatment* of the articular fracture [1, 5–7]. With this treatment strategy, the frame is worn until the fracture is healed or nearly healed. The use of external fixation for definitive treatment introduces more complex mechanical and anatomical issues as the frame must provide stability for months until the fracture heals [9]. During this therapeutic timeframe, the joint may be spanned definitively or for a limited period of time. Joint-spanning monolateral frames most commonly require joint spanning until fracture healing, while tensioned thin wires in ring external fixation can stabilize the articular block allowing for removal of the joint-spanning portion of the frame. Regardless of the device, the elements of the external fixation construct will typically be used to directly stabilize articular fragments. Screw fixation of large metaphyseal or articular fragments often accompanies this type of external fixation strategy.

Technical Issues with Periarticular External Fixation

External fixation of periarticular fractures presents several problems that are impacted by the mechanical stability of the external fixation. There are mechanical challenges that are not present for diaphyseal external fixation. Periarticular external fixation is an art of compromise and alternative techniques, as best practice mechanical principles are impossible due to factors related to soft tissue anatomy and injury.

The local anatomy and injury patterns of different joints present joint-specific challenges preventing a one-size-fits-all philosophy. For example, the proximal and distal tibia capsular reflection limits fixation options adjacent to the joint [10–12]. The joint capsule inserts at variable distances onto metaphyseal segments, depending on the joint and location of the fixation. Septic arthritis is a concern with intracapsular placement of external fixation pins or tensioned wires [13]. In the proximal tibia, risk for this complication can be reduced by placing fixation a minimum of 14 mm distal to the subchondral bone. If closer fixation is required, pins can be placed in the anterior half of the joint at least 6 mm from the subchondral bone [14]. Capsular reflection of the ankle joint to the distal tibia extends an average of 32 mm from the tip of the medial malleolus and 21 mm from the anterolateral joint line [11]. Cadaveric studies have defined the "safe zones" of pin placement are in the distal femur [12]. Fixation can be safely placed in the distal femur, proximal to the adductor tubercle and in the posterior three-fourths of the femur to avoid intracapsular placement of fixation. High-energy periarticular fractures frequently produce significant associated injury to the surrounding soft tissues, which may limit otherwise optimal paths for pin or wire insertion. Cross joint external fixation may transfix muscle tendon units that bridge a joint, binding joint motion.

As a result of some fractures being close to the articular surface, there is generally a short segment on one side of the construct. Standard mechanical principles indicate that pin spread is optimal, but this is not possible with a short periarticular segment. In addition, optimal stability requires a fixation element close to the fracture, but severe soft tissue injury, presence of the joint, and the need to avoid compromise of future approaches for internal fixation often prevent this optimal construct.

The fracture itself presents further challenges. The short segment is typically fractured into a variable amount of comminution. The articular block must be reconstructed by reducing and fixing the fragments prior to placing external fixation pins or wires. These pins or wires then must be placed directly across fracture lines, a technique that typically would be avoided in other injured areas. To make matters more challenging, the articular block that must support the external fixation pins or wires is in the metaphysis, which provides less robust fixation than cortical bone in diaphyseal segments.

Pin and wire infections are frequent occurrences with all external fixators. They are usually easily treated with local antiseptic, antibiotics, or pin removal, but when they occur near joints or when pins or wires become loose, infection and local pain can become a significant issue [15].

While not all of these challenges are directly biomechanical, they all need to be considered in designing frames for the diverse mechanical challenges presented by high-energy periarticular fractures.

Advantages of External Fixation in Periarticular Fractures

The issues and problems with external fixation of periarticular fractures are at least partially offset by the many advantages that are afforded by this technique. In frame construction, there is tremendous flexibility based on the needs of the fracture. Easy rapidly applied frames are the workhorse of temporary treatment. However, more robust frames needed for highly unstable fractures or for fractures where prolonged definitive treatment is planned can be constructed with experience. External fixation with ring fixators can come close to providing circumferential stabilization of an injured joint.

Since external fixation elements are applied percutaneously, treatment with external fixators is amenable to limited approaches or percutaneous reduction approaches. This is well suited to severe soft tissue injures. Treatment with external fixators is optimal for minimizing the risk of wound breakdown and deep infections [1, 5–7, 16]. External fixators can cross joints in ways not possible with internal fixation. This is ideal for temporary fixation but can also be used to support across a joint, adding additional stability as part of definitive treatment. Also, a cross joint frame can be used to assist with fracture reduction. External fixation frames can be used to adjust and fine-tune reductions, assuring optimal limb alignment [17]. Finally, external fixation allows for controlled loading of a fracture by varying the stability of the frame during treatment.

Current Indications

All indications for external fixation are relative—there are no absolute indications. External fixation use is highly surgeon dependent. Some surgeons have specific indications for definitive treatment of periarticular fractures in external fixation. Others rarely use this technique, preferring temporary frames followed by delayed internal fixation. Clinical use may be limited in part by the cost as a single articular fracture requires two surgical procedures instead of one, thus significantly increasing total cost for a patient [18].

The more severely injured a patient, with one or more periarticular fractures, the greater the indication for the use of a temporary *joint-spanning external fixation.* The rapid application; the ability to avoid splints or casts, to visualize soft tissues circumferentially, and to allow soft tissue procedures; and the relative ease of mobilizing the patient are all advantages in polytrauma patients.

The more severe the fracture, the more useful the temporary joint-spanning frame. Severe comminution, shortening displacement, open wounds, and severe closed soft tissue injury are all characteristics of the injury benefit from joint-spanning frames (Table 6.1).

The most common joint to be spanned is the ankle, followed by the knee. Some severe foot fractures/dislocations may benefit from spanning

Table 6.1 Principles of temporary joint-spanning external fixation

Restore length
Restore alignment
Stabilize the fracture
Provide easy access to the soft tissues/wound
Avoid overlapping external fixation pins with future internal fixation
Safe pin placement to avoid injury to neurovascular structures and binding tendons
Minimize frame cost

external fixation [19]. In the upper extremity the wrist is frequently spanned and, occasionally, the elbow. Spanning fixation of the hip or the shoulder is rare because proximal fixation in either of these areas is problematic. Subsequent definitive treatment options include definitive external fixation or internal fixation with plates and screws. Modality of definitive treatment is surgeon dependent. External fixations of the distal tibia, proximal tibia, and distal radius are the most commonly utilized areas.

External fixation may be used to provide cross joint support and stabilization, which may reduce the required amount of internal fixation and provide additional joint stability. An external fixator may be used to provide joint distraction as means to minimize loads on articular cartilage and to stimulate fracture healing. Articulated cross joint external fixators allow joint mobility while still providing fracture stabilization, with a theoretical benefit of improved synovial fluid circulation and improved range of motion. The hinge of the frame must approximate the axis of rotation of the joint [20, 21].

Biomechanics of Periarticular External Fixation—General Principles

The mechanical demands and principles for construct stability are very different for temporary joint-spanning frames than for definitive same side of the joint frames, so they will be discussed separately in Chap. 4.

Temporary joint-spanning external fixators are used for complex periarticular fractures. Fixation is typically in diaphyseal cortical bone, with notable exception in the foot. The frame provides stability for days up to 2–3 weeks. Modern pin-to-bar frames allow a wide variety of frame constructs to be built based on the needs of the injury and patient. The goals in the short term are simple: (1) Restore length and alignment, (2) allow for planning of soft tissue management and later complex fracture management, and (3) mobilize the patient until definitive treatment. Table 6.1 identifies some basic principles of temporary joint-spanning external fixation. A mechanical principle of joint-spanning frames is that spanning the joint comes with the price of decreasing frame stability. However, the fixation is typically designed to be provisional. Thus, spanning the joint with the expense of losing frame stability is acceptable. Further explanation of mechanical principles of external fixation can be found in Chap. 4.

Frames are made up of half-pins, bars, and clamps. Half pins are the workhorses of temporary joint-spanning frames, as such most of the discussion in this section will focus on pins. Specifically, we will discuss how pin diameter, spread (distance between pins), size, and location elements are all important factors mediating frame stiffness. For standard applications, two proximal and two distal pins or a transfixion pin generally suffice. Pin spread increases stability [22]. In the lower extremity 6 mm pins are more stable than 5 mm pins. Pin-to-bar and bar-to-bar couplings provide a basic construct to distract the limb and restore and secure length and alignment. Care must be taken to avoid placing external fixation in the path of planned future internal fixation incisions, as this has been shown to increase risk of deep infection [23]. Additional pins, bars, and planes of fixation can be used to increase the stability of the frame. Half pins and transfixion pins are the preferred fixation devices for temporary joint-spanning frames. Many easy-to-use pin-to-bar external fixation devices are on the market for joint-spanning external fixation in periarticular fractures. These systems have simple pin clamps and pin-to-bar/bar-to-bar connectors that can be rapidly applied in an expedited fashion that is often required when managing polytrauma patients.

Hydroxyapatite-coated pins are not used for temporary joint-spanning external fixation. Less expensive stainless steel and titanium pins are used in most systems.

Bicortical fixation, especially in the diaphysis, is important for stability. Centrally threaded transfixion pins used most commonly in the calcaneus provide balanced medial and lateral stability when spanning frames are placed across the ankle joint.

It is important to understand pin characteristics that contribute to frame stiffness. Six millimeter pins are recommended in the femur and tibia for added stability. Pin stiffness is proportional to the radius of the pin to the fourth power.

In pediatric patients or adults with atypically narrow long bones, the maximum pin diameter should be less than one-third the width of the bone in narrowest diameter to prevent fracture through the pin tract. The pin is at risk of failing at the near cortex when the frame is used for early weight-bearing. Self-drilling pins have disadvantages that include increased required depth of pin insertion to engage threads in the far cortex and stripping of the near cortex threads when the pin initially engages the far cortex. Advantages of self-drilling pins, including quick insertion and simpler technique, make these pins adequate for temporary joint-spanning frames. Other techniques for improved external fixation stability are discussed in other chapters but are also worth mentioning here including increasing pin spread to capture the most proximal and distal aspects of the bone segments, adding more pins, stacking bars, placing out-of-plane fixation, placing pin parallel to the plane of motion, and adding pin fixation closer to the fracture.

Ring external fixation and thin tensioned wires are not typically chosen for temporary periarticular fracture stabilization, but in some cases when the treating surgeon is planning to use this fixation for definitive fixation, these implant devices can be used. Typically, simple pin-to-bar spanning fixation is used to allow for easy access to soft tissue and more complex soft tissue reconstruction procedures in the operating room before ring external fixation is applied. The ring frame can make complex soft tissue reconstruction challenging, and we prefer not to place the ring external fixator until completion of soft tissue coverage procedures.

Generally, temporary joint-spanning external fixation is used for complex periarticular fractures. The frame provides stability for days up to 2–3 weeks. When choosing a frame, a surgeon must consider potential complications related to that frame and the fiscal cost incurred by a patient. A few notable examples are the following. Care must be taken to avoid placing external fixation in the path of planned future internal fixation incisions, as this has been shown to increase risk of deep infection [23]. The goals in the short term are simple: to restore length and alignment, to allow for planning of soft tissue management and later complex fracture management, and to allow the patient to mobilize until definitive treatment.

Definitive Frames

High-energy periarticular fractures in the lower extremity need 3–4 months of external support to heal. Given the challenges of a short broken unstable segment, achieving this can be difficult and requires thoughtful planning to optimize mechanics. All of the challenges will be in the segment near the joint, since diaphyseal fixation above or below the joint will be easily obtained. There is not one construct that is always optimal because fractures, degree of instability, and frame-type choices are all variable. Table 6.2 identifies some basic mechanical and other principles of definitive periarticular frames.

Table 6.2 Principles of definitive treatment of periarticular fractures with external fixation

Adequate fixation of the articular block to allow for fracture stability
Place thin wire fixation with maximum angular and translational spread
Restore alignment
Prevent tethering of tendon units to allow for free joint motion during fracture healing
Prevent soft tissue impingement on external fixation (can be a problem with late swelling)
Safe wire/pin placement to avoid injury to neurovascular structures

Options for definitive periarticular fracture fixation using external fixation include (1) ring external fixation with thin wires, (2) hybrid external fixation, and (3) monolateral frames. Examples of these types of external fixation are described below. The decision to use each of these devices for fixation depends on experience of the surgeon and injury characteristics. Ring external fixation can be performed with threaded rods or hexapod-type struts between the levels. Using threaded rods requires more preoperative planning and attention to detail when placing the ring perpendicular to the axis of the extremity. Hexapod struts are more expensive but provide more stability than threaded rods and allow for simple correction of deformity after the definitive surgery.

In the long diaphyseal bone, fixation considerations are straightforward compared to options in the short periarticular segment. Half pins are the preferred fixation method in the diaphysis because of the ease of use, stability, and less risk of binding soft tissues. Self-drilling pins are described above for temporary external fixation. More precautions should be taken to prevent infection, thermal necrosis, and pin loosening when placing half pins used for definitive fixation. A longitudinal incision is made in the skin large enough to accommodate the sleeve for drilling and pin placement. Too small of an incision will cause the drill and pin to bind the soft tissues. Large incisions heal more slowly around the half pin. Soft tissue should be spread down to the bone. A sharp drill is used to prevent thermal necrosis, and a sleeve is used for ease of pin insertion and for soft tissue protection. Irrigation can be used to cool the drill as necessary in hard, cortical bone. Bone debris is then irrigated from the drilled hole and the pin is inserted with a hand driver. Care is taken when attaching the pin to the frame to prevent bending/preloading of the pin when connected to the frame. Again, 6 mm pins are preferred in the diaphysis of the tibia and femur for stability. Typically, three half pins are placed in each segment of bone. Figure 6.1 demonstrates osteolysis that can occur with thermal necrosis or pin loosening resulting in chronic infection or even fracture. Hydroxyapatite-coated half pins have recently become popular for external fixation that is needed for definitive treatment of fractures. Such pins have been shown to have increased pullout strength, lower infection rates, and less incidence of loosening [24–27].

Fig. 6.1 Osteolysis seen with thermal necrosis or loosening after half pin removal with chronic wound and osteomyelitis

Thin tensioned wires in ring external fixation allow for optimum stabilization of short segment periarticular fractures. Near circumferential stabilization of the metaphysis can be achieved with ring external fixation. Wires should be placed as near to 90° as possible to optimize stability. Typically, in the distal tibia, a transfibular wire and a medial face wire are placed as near perpendicular as allowed by anterior and posterior soft tissues. A third wire is placed traversing the anterolateral fragment typically seen in intra-articular distal tibia fractures. Elevating the wire above and below the ring adds stability by increasing the superior to inferior spread of the wires. Olive wires are used for added medial to

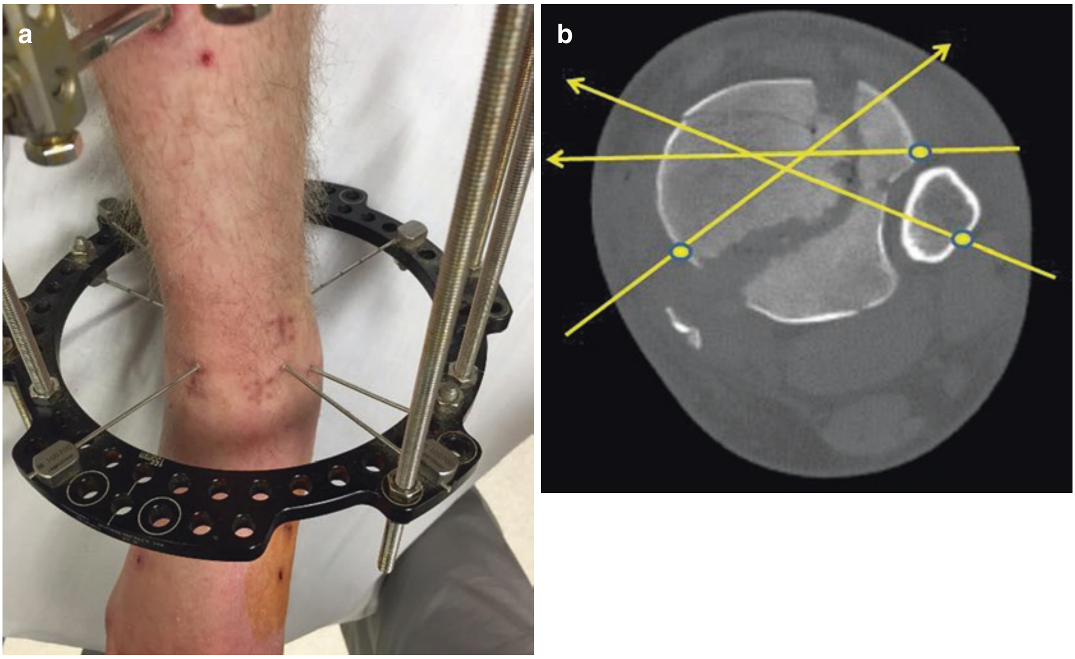

Fig. 6.2 Typical thin wire configuration in the distal tibia. Attempt is made to place wires perpendicular while avoiding neurovascular structures and tendon pathways (**a**) Clinical photograph (**b**) typical wire paths superimposed on a axial CT cut

lateral translational stability and bending stiffness [28]. Figure 6.2 demonstrates thin wire configuration in the distal tibia. In some cases, with a longer periarticular segment, a half pin can be placed for added stability.

Often fixation in the short articular segment is not adequate for early joint motion and weight-bearing, making it necessary to span the joint with temporary fixation. In the distal tibia, this can be achieved with a U-ring and tension wire fixation in the calcaneus and forefoot. Joint-spanning external fixation can be useful for ligamentotaxis and indirect fracture reduction at the time of definitive articular fracture reduction. Spanning the joint allows for immediate weight-bearing on articular fractures. After a short period of time protecting the articular segment fixation with the joint-spanning portion of the frame, the frame is removed from the far side of the joint in 3–6 weeks after definitive fixation to allow for joint motion. An example of this frame removal and early weight-bearing is demonstrated in section "Case 1: Temporary Joint-Spanning External Fixation".

In most cases external fixation is removed in the clinic after fracture healing [29]. In rare instances anesthesia is required in the operating room for pediatric patients, more complex frames, or when additional surgical procedures are indicated.

Definitive joint-spanning external fixation is used commonly for distal tibia and distal radius fractures with monolateral frames. This technique is less common but can be also used in the knee and elbow. In fractures with a very short periarticular segment, where same-side external fixation is impossible or dangerous for intracapsular fixation, joint-spanning external fixation keeps fixation out of the zone of injury. Again, the joint-spanning frame allows for distraction with ligamentotaxis assisting with articular fracture reduction. Because this technique immobilizes the joint for an extended time, residual joint stiffness is a concern.

Case 1: Temporary Joint-Spanning External Fixation

Temporary joint-spanning external fixation allows for correction of deformity and stabilization of fracture fragments and local soft tissues. This can be achieved with relatively simple and inexpensive external fixation. This first case example is a healthy 22-year-old female who sustained a motor vehicle collision that resulted in an open intra-articular distal tibia fracture. She had a stellate medial wound with significant deformity (Fig. 6.3). Simple pin-to-bar ankle-spanning external fixation was performed with self-drilling 6 mm pins in the tibia and a centrally threaded 5 mm transfixion pin in the calcaneus (Fig. 6.4). Care was taken to avoid overlapping with definitive internal plate fixation. A posterior splint was used to prevent equinus while the soft tissues healed over 2 weeks (Fig. 6.5).

Figure 6.6 shows similar case of temporary ankle-spanning external fixation for a pilon fracture, the least expensive configuration in our system. If more stability was required in this case, an out-of-plane half pin could be placed closer to the fracture site in the tibia diaphysis. Additionally, half pins could be placed in the talar neck, midfoot (cuneiforms), or forefoot to prevent equinus and improved stability.

In our practice, a dilute chlorohexidine-soaked gauze dressing is placed over the pin sites in the operating room. For temporary frames, patients leave this dressing in place and do not perform pin site cares [30]. Dressing is changed daily over

Fig. 6.3 A 22-year-old female with an open, intra-articular distal tibia fracture

Fig. 6.4 Patient shown in Fig. 6.3 after reduction and temporary stabilization in joint-spanning external fixation

Fig. 6.5 Medial soft tissue wound of patient shown in Fig. 6.3 that is easily monitored with simple joint-spanning external fixation

Fig. 6.6 Another patient with a distal tibia fracture that demonstrates simple pin-to-bar ankle-spanning external fixation. The proximal pin clamp is the least expensive construct in our healthcare system. A posterior slab splint or brace can be used to help control equinus

the open wound and monitored for infection. After allowing time for soft tissue healing to have occurred (generally 14 days or so), definitive fracture fixation is performed with a lateral approach to the distal tibia (Fig. 6.7). This case demonstrates typical temporary joint-spanning external fixation of periarticular distal tibia fractures followed by definitive open reduction and internal fixation.

Case 2: Same-Side Definitive External Fixation

Definitive same-side external fixation is a reliable option to treat patients with a high risk of soft tissue complications and deep surgical site infection. The second case example is a 31-year-old man with poorly controlled type 1 diabetes. His hemoglobin A1C was 10.8% on presentation, and he had a chronic wound over his left anterior mid-tibia. He sustained a fall while rock climbing and had bilateral closed, intra-articular distal tibia fracture dislocations (Fig. 6.8). We elected to perform definitive fixation of the fractures with ring external fixation because of his poorly controlled diabetes and chronic distal tibial wound. Small incisions and indirect, fluoroscopic techniques were used for reduction and screw fixation of the articular fracture fragments. Typically, 4.0 mm partially threaded cancellous screws are used for fixation, but 3.5 or 2.7 mm screws can be used for smaller fracture fragments. Half pins were placed in the tibia diaphysis, and tensioned olive wires were used for fixation in the articular block. The ankle was spanned to protect the distal tibia fixation for 6 weeks after definitive fixation (Fig. 6.9). He remained weight-bearing as tolerated throughout the treatment course but was only able to stand to transfer to a wheelchair while in the ankle-spanning portion of the frame. Figure 6.10 demonstrates the patient standing at 12 weeks after

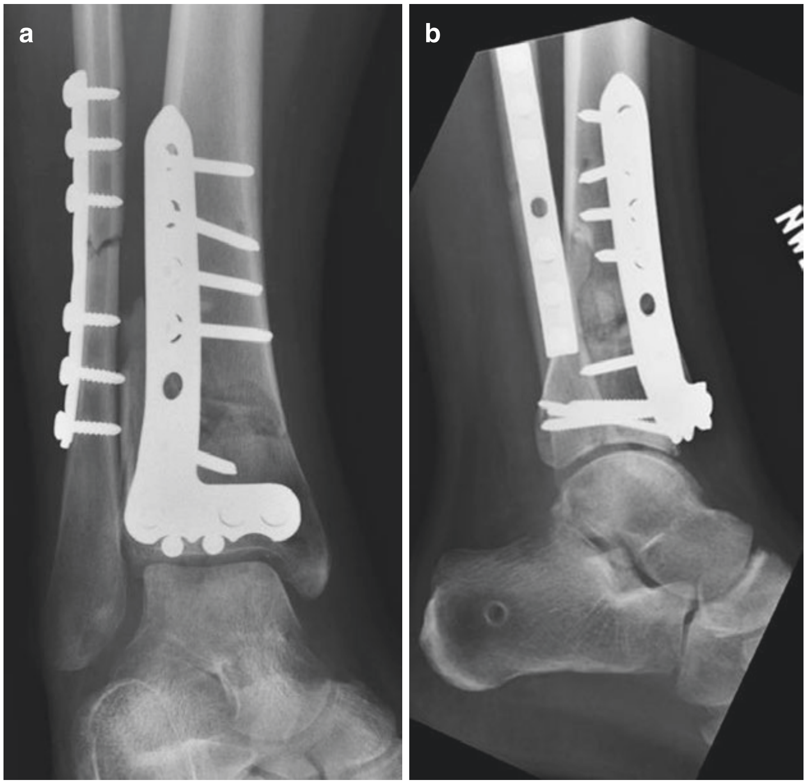

Fig. 6.7 Patient in Fig. 6.3 who underwent temporizing ankle-spanning external fixation with definitive fracture fixation 2 weeks after the initial injury; medial soft tissues had healed (**a**) AP and (**b**) lateral radiographs

surgery, and standing radiographs are shown at 15 months (Fig. 6.11) after fracture.

Patients who are in external fixation for an extended period of time in our practice do minimal pin site care. The dilute chlorohexidine dressing is removed 2 weeks after surgery. This is left in place initially to allow for skin healing around the pins. After the initial 2 weeks, no dressing is placed over the pins. Sterile saline and a cotton tip applicator are used to clean exudate from pins that have minor drainage. External fixation remained in place for 5 months until fracture had completely healed. The patient was placed in a cast for 2 weeks after removal and then resumed weight-bearing. This case demonstrates typical definitive ring external fixation of distal tibia fractures. Small incisions and indirect techniques were used for articular fracture fragment reduction and fixation, leading to imperfect articular reduction, but this technique is felt to result in lower risk of soft tissue complications including chronic infections in these high-risk patients.

Case 3: Articulated External Fixation

Attempting to maintain joint motion while stabilizing an unstable fracture with a spanning external fixation is challenging. Articulated external fixation has been utilized in multiple joints with varying success [31, 32]. Most commonly, articulated external fixation is performed in the elbow joint [33]. The third case example is a 40-year-old female who presented 4 weeks after sustaining an elbow dislocation as part of polytrauma (Fig. 6.12). Her elbow initially was reduced and splinted, but she was found to have multiple redislocations while immobilized and was referred for further management. She was found to have a small coronoid tip fracture best characterized on computed tomography (CT) scan but had primarily ligamentous instability of the elbow. She underwent lateral collateral ligament repair with a suture anchor. Hinged elbow external fixation

Fig. 6.8 A 31-year-old man with bilateral intra-articular distal tibia fracture dislocations

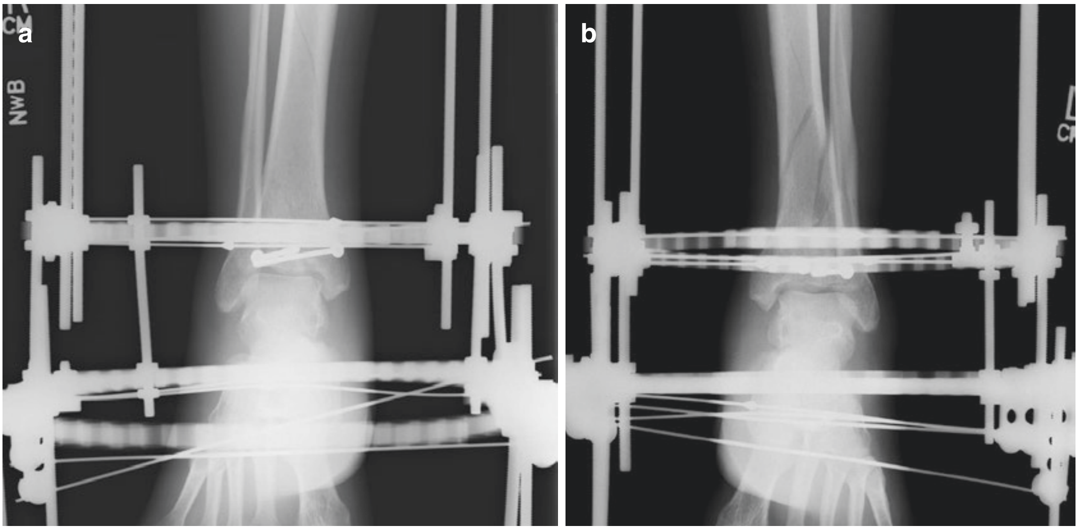

Fig. 6.9 Patient shown in Fig. 6.8 after definitive treatment of bilateral intra-articular distal tibia fractures treated with ring external fixation. AP radiographs are shown (**a**) right side (**b**) left side

Fig. 6.10 Weight-bearing 12 weeks after same-side ring external fixation of bilateral pilon fractures

Fig. 6.11 15-month follow-up standing radiographs of the patient shown in Fig. 6.8 after bilateral pilon fractures treated in ring external fixation

Fig. 6.12 A 40-year-old female with subacute elbow instability after a polytrauma

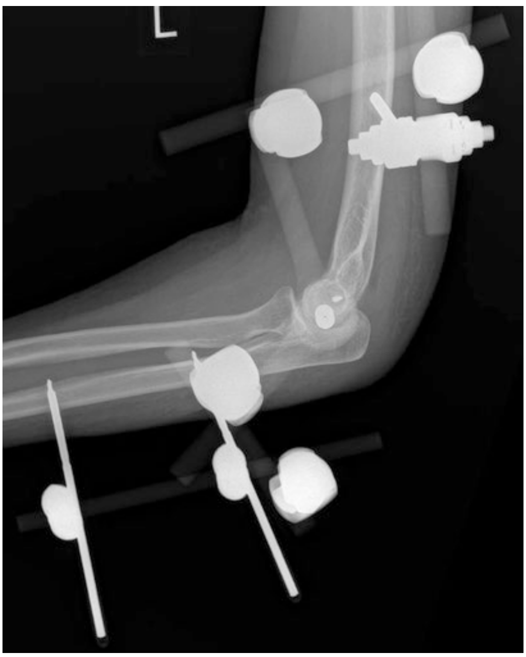

Fig. 6.13 The patient shown in Fig. 6.12 after repair of the lateral collateral ligament and placement of hinged external fixation

was used to stabilize the elbow while maintaining a limited arc of motion (Fig. 6.13). Aligning the external fixation hinged with the axis of rotation at the joint remains the most challenging part of this procedure [34]. The external fixation was removed at 6 weeks, and the elbow

Fig. 6.14 The external fixation was removed at 6 weeks and the elbow remained stable but had limitations to range of motion

remained reduced but had significant limitations to range of motion (Fig. 6.14). Retrospective studies have demonstrated similar range of motion with less device-related complications when comparing cross pinning of the elbow joint to hinged external fixation [35], making this a controversial clinical problem.

Conclusions

The biomechanics of external fixation for periarticular fractures can be challenging. For temporary treatment, a fairly simple frame can be rapidly applied that restores length and alignment with multiple different possible constructs. Standard biomechanical principles of external fixation apply, but a long-spanned segment with a short segment articular block will always have some degree of instability at the fracture site. Stably fixing periarticular fractures for definitive treatment with frames is challenging because of the severity of the injury and the difficult local anatomy and fracture location near a joint. A variety of strategies are possible, which need to be individualized for each case. Often the joint is spanned for a short period of time to allow weight-bearing while the articular block heals. Achieving mechanical stability of complex periarticular fractures with a frame and maintaining it until the fracture is healed need careful planning and attention to the local circumstances presented by the injury.

References

1. Marsh JL, Smith ST, Do TT. External fixation and limited internal fixation for complex fractures of the tibial plateau. J Bone Joint Surg Am. 1995;77(5):661–73.
2. Bone L, Stegemann P, McNamara K, Seibel R. External fixation of severely comminuted and open tibial pilon fractures. Clin Orthop Relat Res. 1993;292:101–7.
3. Sirkin M, Sanders R, DiPasquale T, Herscovici D Jr. A staged protocol for soft tissue management in the treatment of complex pilon fractures. J Orthop Trauma. 2004;18(8 Suppl):S32–8.
4. Ketz J, Sanders R. Staged posterior tibial plating for the treatment of Orthopaedic Trauma Association 43C2 and 43C3 tibial pilon fractures. J Orthop Trauma. 2012;26(6):341–7.
5. McDonald MG, Burgess RC, Bolano LE, Nicholls PJ. Ilizarov treatment of pilon fractures. Clin Orthop Relat Res. 1996;325:232–8.
6. Bonar SK, Marsh JL. Unilateral external fixation for severe pilon fractures. Foot Ankle. 1993;14(2):57–64.
7. Tornetta P 3rd, Weiner L, Bergman M, Watnik N, Steuer J, Kelley M, et al. Pilon fractures: treatment with combined internal and external fixation. J Orthop Trauma. 1993;7(6):489–96.
8. Molina CS, Stinner DJ, Fras AR, Evans JM. Risk factors of deep infection in operatively treated pilon fractures (AO/OTA: 43). J Orthop. 2015;12(Suppl 1):S7–S13.
9. Kapoor SK, Kataria H, Patra SR, Boruah T. Capsuloligamentotaxis and definitive fixation by an ankle-spanning Ilizarov fixator in high-energy pilon fractures. J Bone Joint Surg Br. 2010;92(8):1100–6.
10. Reid JS, Van Slyke MA, Moulton MJ, Mann TA. Safe placement of proximal tibial transfixation wires with respect to intracapsular penetration. J Orthop Trauma. 2001;15(1):10–7.
11. Vives MJ, Abidi NA, Ishikawa SN, Taliwal RV, Sharkey PF. Soft tissue injuries with the use of safe corridors for transfixion wire placement during external fixation of distal tibia fractures: an anatomic study. J Orthop Trauma. 2001;15(8):555–9.
12. Lowery K, Dearden P, Sherman K, Mahadevan V, Sharma H. Cadaveric analysis of capsular attachments of the distal femur related to pin and wire placement. Injury. 2015;46(6):970–4.
13. Hutson JJ Jr, Zych GA. Infections in periarticular fractures of the lower extremity treated with tensioned wire hybrid fixators. J Orthop Trauma. 1998;12(3):214–8.
14. DeCoster TA, Crawford MK, Kraut MA. Safe extracapsular placement of proximal tibia transfixation pins. J Orthop Trauma. 1999;13(4):236–40.
15. Mahan J, Seligson D, Henry SL, Hynes P, Dobbins J. Factors in pin tract infections. Orthopedics. 1991;14(3):305–8.
16. Bacon S, Smith WR, Morgan SJ, Hasenboehler E, Philips G, Williams A, et al. A retrospective analysis of comminuted intra-articular fractures of the tibial plafond: open reduction and internal fixation versus external Ilizarov fixation. Injury. 2008;39(2):196–202.
17. Moed BR, Watson JT. Intramedullary nailing of the tibia without a fracture table: the transfixion pin distractor technique. J Orthop Trauma. 1994;8(3):195–202.
18. Wetzel RJ, Kempton LB, Lee ES, Zlowodzki M, McKinley TO, Virkus WW. Wide variation of surgical cost in the treatment of periarticular lower extremity injuries between 6 fellowship-trained trauma surgeons. J Orthop Trauma. 2016;30(12):e377–e83.
19. Kadow TR, Siska PA, Evans AR, Sands SS, Tarkin IS. Staged treatment of high energy midfoot fracture dislocations. Foot Ankle Int. 2014;35(12):1287–91.
20. Fitzpatrick DC, Sommers MB, Kam BC, Marsh JL, Bottlang M. Knee stability after articulated external fixation. Am J Sports Med. 2005;33(11):1735–41.
21. Fitzpatrick DC, Marsh JL, Brown TD. Articulated external fixation of pilon fractures: the effects on ankle joint kinematics. J Orthop Trauma. 1995;9(1):76–82.
22. Roberts CS, Dodds JC, Perry K, Beck D, Seligson D, Voor MJ. Hybrid external fixation of the proximal tibia: strategies to improve frame stability. J Orthop Trauma. 2003;17(6):415–20.
23. Shah CM, Babb PE, McAndrew CM, Brimmo O, Badarudeen S, Tornetta P 3rd, et al. Definitive plates overlapping provisional external fixator pin sites: is the infection risk increased? J Orthop Trauma. 2014;28(9):518–22.
24. Moroni A, Toksvig-Larsen S, Maltarello MC, Orienti L, Stea S, Giannini S. A comparison of hydroxyapatite-coated, titanium-coated, and uncoated tapered external-fixation pins. An in vivo study in sheep. J Bone Joint Surg Am. 1998;80(4):547–54.
25. Moroni A, Aspenberg P, Toksvig-Larsen S, Falzarano G, Giannini S. Enhanced fixation with hydroxyapatite coated pins. Clin Orthop Relat Res. 1998;346:171–7.
26. Piza G, Caja VL, Gonzalez-Viejo MA, Navarro A. Hydroxyapatite-coated external-fixation pins. The effect on pin loosening and pin-track infection in leg lengthening for short stature. J Bone Joint Surg Br. 2004;86(6):892–7.

27. Pommer A, Muhr G, David A. Hydroxyapatite-coated Schanz pins in external fixators used for distraction osteogenesis: a randomized, controlled trial. J Bone Joint Surg Am. 2002;84-A(7):1162–6.
28. Antoci V, Voor MJ, Antoci V Jr, Roberts CS. Biomechanics of olive wire positioning and tensioning characteristics. J Pediatr Orthop. 2005;25(6):798–803.
29. Ryder S, Gorczyca JT. Routine removal of external fixators without anesthesia. J Orthop Trauma. 2007;21(8):571–3.
30. Egol KA, Paksima N, Puopolo S, Klugman J, Hiebert R, Koval KJ. Treatment of external fixation pins about the wrist: a prospective, randomized trial. J Bone Joint Surg Am. 2006;88(2):349–54.
31. Marsh JL, Muehling V, Dirschl D, Hurwitz S, Brown TD, Nepola J. Tibial plafond fractures treated by articulated external fixation: a randomized trial of postoperative motion versus nonmotion. J Orthop Trauma. 2006;20(8):536–41.
32. Sommers MB, Fitzpatrick DC, Kahn KM, Marsh JL, Bottlang M. Hinged external fixation of the knee: intrinsic factors influencing passive joint motion. J Orthop Trauma. 2004;18(3):163–9.
33. Madey SM, Bottlang M, Steyers CM, Marsh JL, Brown TD. Hinged external fixation of the elbow: optimal axis alignment to minimize motion resistance. J Orthop Trauma. 2000;14(1):41–7.
34. von Knoch F, Marsh JL, Steyers C, McKinley T, O'Rourke M, Bottlang M. A new articulated elbow external fixation technique for difficult elbow trauma. Iowa Orthop J. 2001;21:13–9.
35. Ring D, Bruinsma WE, Jupiter JB. Complications of hinged external fixation compared with cross-pinning of the elbow for acute and subacute instability. Clin Orthop Relat Res. 2014;472(7):2044–8.

7 External Fixators for Limb Lengthening

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General Principles

Distraction osteogenesis (DO) refers to the process of forming new bone at the site of a corticotomy/osteotomy undergoing gradual distraction [1]. The new bone that forms during the process of DO is termed "regenerate." Regenerate formation begins as bony fracture callus at the site of the cut in the bone and, with the application of gradual distraction, forms a column of new bone extending from this site primarily through a process of intramembranous ossification [2, 3]. When DO is used to make new bone to treat a segment of bone loss, it is called bone transport. When DO is used to lengthen an extremity, it is often termed limb lengthening or distraction histogenesis. The term distraction histogenesis is preferred in this scenario because it emphasizes that in addition to new bone formation there is also generation of vascular, nerve, and other soft tissue structures.

DO is most often performed with external fixation. The process begins with application of the external fixator. An Ilizarov circular external fixator is the most classic method, but many types of external fixators can be used, including other varieties of circular fixation, monolateral rails, hexapods, and cable constructs. The chosen construct is then used to achieve angular correction, lengthening of the limb, and/or transportation of the bone. Once the external fixator is applied, the next step is to cut the bone with either a corticotomy or osteotomy. Following completion of the operation, DO progresses through three phases; latency, distraction, and consolidation. The latency period is usually 3–7 days during which early bony callus forms and neovascularization of the bone at the corticotomy site occurs. The distraction phase then begins, usually at a rate of 1 mm per day, until the desired length and angular correction is obtained. The consolidation phase follows during which calcification and maturation of the regenerate bone occur.

Applying an external fixation construct that is mechanically sound and stable throughout the process of DO and performing an appropriate bone cut are critical to the success of the procedure. Early descriptions of DO paid a great deal of attention to the concept of a corticotomy in which the periosteum and endosteal bone along with their blood supply were preserved in their entirety [4]. This method therefore aimed to cut just the bony cortex whether performed with a drill, osteotome, or Gigli saw. The importance of the true corticotomy has been challenged over time as being both impractical and unnecessary

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to achieve a successful result. It has been demonstrated that an osteotomy in which the cancellous bone is also cut and the periosteum separated can also form excellent regenerate. However, the core concept embodied in the original description of a corticotomy—that performing a low-energy bone cut preserves local vascularity and minimizes damage to the periosteum—remains essential to a successful result. The modern concept of a corticotomy/osteotomy is thus focused primarily on the critical aspects of minimal energy, minimizing damage to local blood supply and preventing thermal injury. There are a number of methods that have been shown to achieve this successfully, including a multiple drill hole osteotomy completed with methods such as rotational osteoclasis, inserting an osteotome at the corticotomy site and rotating it 90°, or using a Gigli saw [5]. Although no human studies are available, animal models show delayed consolidation when using higher-energy techniques more prone to burning the bone (such as an oscillating saw) to perform the osteotomy, and therefore this technique is highly discouraged [6]. The metaphyseal region is an ideal site to perform a corticotomy because of the large trabecular surface and robust vasculature that often leads to a large amount of regenerate, although other regions of the bone can also be used when necessary [7].

Prior to starting the distraction phase, a postoperative latency period is advocated [4]. This is usually 3–7 days during which early callus formation and local neovascularization occur. The exact length of the latency period should be individualized for each patient based on physiologic factors. Once begun, distraction at the osteotomy site typically proceeds at 1 mm/day. This rate was established by the work of Ilizarov, who found that in dog studies 0.5 mm/day of distraction can result in premature consolidation, while 2.0 mm/day produced poor regenerate [7]. Although 1 mm is most common, the rate may need to be altered due to patient factors. For example, young children may require a faster rate to prevent premature consolidation. In contrast, patients with multiple comorbidities such as diabetes and smoking may require a slower rate to allow for good regenerate formation that does not outpace the neovascularization occurring at the scene of the regenerate.

In addition to rate, rhythm is also an important aspect. Ilizarov demonstrated that more frequent and shorter distance distractions lead to improved regenerate formation. However, dividing the distraction into a large number of separate distractions is impractical, and so the recommendation is made to use a rhythm of four separate one-fourth millimeter turns per day, which is practical and also achieves excellent bone formation. It is notable that comparisons between regenerate formed using Taylor Spatial Frame (TSF) (Smith & Nephew, London, UK) distraction at 1 mm/day in one increment and that with an Ilizarov at one-fourth mm 4 × per day have failed to demonstrate a difference in quality of regenerate bone formation. This is encouraging when it is necessary to use the less frequent protocol, but it is best to respect the more scientifically rigorous data from the basic science studies and use the more frequent rhythm when feasible. Following completion of distraction, the new bone calcifies and remodels to form a cortex and medullary canal [8–10].

Distraction also induces morphologic changes in surrounding soft tissues. Muscle tissue undergoes hypertrophy and hyperplasia. Neoangiogenesis occurs in the direction of the tension vector. Nerves to innervate the growing and new tissue develop as well. Different tissues have different biologic compositions, and thus the "optimal" distraction rate is different for each tissue than it is for the bone. This difference is one reason for nerve palsies and joint contractures, which will be discussed later.

Biology of Distraction

The classic experiments done by Ilizarov provide helpful insight into the biochemical, mechanical, and biophysical processes that are involved in DO. After corticotomy, a local inflammatory response facilitates new bone formation during the latency period. This response is multifactorial but primarily consists of migration of pluripotential cells and the secretion of cytokines and growth factors to guide osteogenesis. During distraction, regenerate has a characteristic histologic appearance with five zones that resemble a growth plate [11]. The central portion is a growth zone, with fibroblast-like cells that secrete collagen. These collagen fibers align parallel to the distraction force being applied. This zone is bordered on either side by a mineralization front, with osteoblasts producing osteoid in a manner that resembles intramembranous ossification. This occurs without any endochondral ossification when a stable, rigid construct is used. If there is some instability, the process is slowed and more closely resembles endochondral ossification or even pseudoarthrosis if gross instability is present [7]. Between the mineralization front and the surface of the native corticotomized bone lies a zone of microcolumn formation. Primary bone is mineralized in this zone, which later in the consolidation phase continues to cross-link and remodel all zones of the regenerate. By this mechanism, the distraction gap is replaced by mature, remodeled bone with distinct medullary canal and cortices, in accordance with Wolff's law [11].

Several signaling molecules have been identified to play an important role in the process of DO. They are categorized as (1) pro-inflammatory cytokines, (2) transforming growth factor-β (TGF-β) and bone morphogenetic family of proteins, and (3) angiogenic factors [12]. Pro-inflammatory cytokines initiate the repair cascade after corticotomy. Interleukin-1 (IL-1), tumor necrosis factor-α (TNF-α), and IL-6 are elevated in the latency and distraction phases and play a significant role in the process of intramembranous bone formation and remodeling [13, 14]. Insulin growth factor-1 (IGF-1) is elevated early in the distraction phase, and levels decrease once distraction stops, suggesting a key role in osteogenesis [15]. TGF-β has been found to support new bone formation, and its levels are elevated during the early distraction phase [16, 17]. Bone morphogenetic proteins (BMPs) are also upregulated, and high levels are maintained throughout the distraction phase [18–20]. BMP-2 has been shown to accelerate the rate of bone formation in rabbits [20]. Vascular endothelial growth factor (VEGF) is recognized as an important stimulator for angiogenesis, and levels are increased during the distraction phase. Angiogenesis facilitates the diffusion of signaling molecules during DO.

Mechanotransduction plays an important role in osteogenesis by triggering cell signaling and gene expression [21, 22]. Integrins are key in cell signaling and are the primary pathway by which mechanotransduction induces stem cell differentiation [23, 24]. Mechanical load by the distractor also stimulates the osteoblastic production of extracellular matrix proteins, such as collagen type I and osteocalcin [17].

Finally, the fluid flow theory helps to explain how mechanical load translates into bone remodeling. The theory proposes that load forces interstitial fluid to flow around the bone microarchitecture, creating shear strain. This initiates the downstream cascade of cell signaling, mainly through the activity of nitric oxide (NO), prostaglandins, and Wnt [25, 26]. Wnt is upregulated by shear stress, which leads to osteoblastic bone formation and inhibition of osteoclast formation [27, 28].

External Fixator Construction

The goals of using external fixation in bone transport are to maintain stable bone alignment and to allow adequate compression of bone at the docking site to encourage healing [1, 29]. The construct should be stable enough to permit weight-bearing and to allow as normal as possible functioning of the limb and adjacent joints. Weight-bearing and limb use help support local neovascularity and facilitate bone healing [30]. Stability of the construct is multifactorial [30, 31], and general concepts of the biomechanics of external fixator constructions and fixation block composition are discussed in earlier chapters. These concepts apply equally when DO is being performed, but in this circumstance, there are a number of additional considerations relevant to creating a construct that will achieve the reconstructive goals. DO can be used for limb lengthening, bone transport, or a combination of both with osteotomies proximal, distal, or both. The construct employed depends primarily on the goals of the procedure in terms of the location of malangulation, bone defect, planned lengthening, and planned osteotomy(s).

There are many external fixator configurations capable of achieving stable fixation and successful DO for limb lengthening. Circular fixators are most often used because they are mechanically the soundest and allow great flexibility in obtaining and maintaining stability. Ilizarov discovered DO and pioneered the use of ring fixation to apply this method. He used a frame constructed of threaded rods attached to stainless steel or carbon fiber rings. These rings were fixed to the bone with high-tension wires both proximal and distal to the zone of injury and/or osteotomy site (Fig. 7.1a). Limb lengthening can be performed at the site of an osteotomy by distracting the rings using telescopic rods such as seen in Fig. 7.1a, b.

Another circular fixation construct well suited to performing limb lengthening is a hexapod fixator such as seen in Fig. 7.1c. This construct uses six struts instead of four threaded rods to stabilize across the distraction site. The advantage of this method is that it allows for a simple method of deformity correction and/or correction of angulation that develops during lengthening caused by either an imperfectly mounted fixator or a drift of the transport segment away from its original alignment during transport caused by instability of the transport segment fixation and/or uneven soft tissue tensions. Hexapods use a computer program generated from information about the frame and osteotomy that informs the patient about which struts to turn and how often. Hexpods

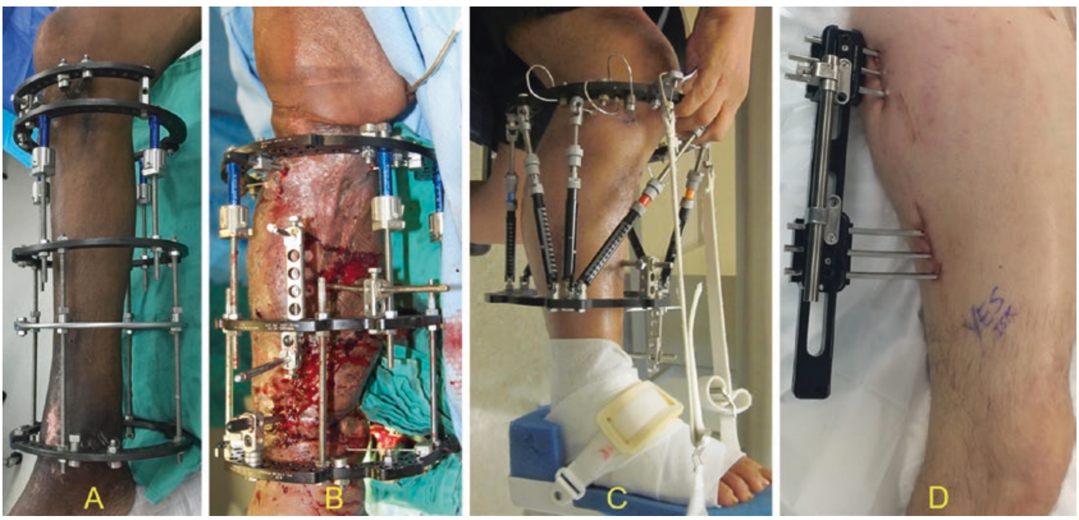

Fig. 7.1 Frame constructs used for lengthening. (**a**) Ilizarov external fixator constructed as both a lengthening and bone transport frame. Fixation rings are carbon fiber and the transport ring is stainless steel. The rings are fixed to the bone with high tension wires and hydroxyapatite-coated half pins. Lengthening is motored by the telescopic rods (often referred to as "clickers"). The clickers are designed to motor the lengthening with the patient making turns in one-fourth mm increments. (**b**) Ilizarov-type external fixator with stable block of two rings distally connected by threaded rods and connection of top two rings with telescopic rods. Distraction occurs between the proximal and middle rings driven by the telescopic rods. (**c**) Hexapod external fixator (specifically, a Taylor Spatial Frame (TSF) in this example) being used as a lengthening construct. There are proximal and distal fixation blocks each built off of a single ring with fixation widely spread. Distraction is motored by the TSF struts between the rings. The struts move in 1 mm increments when turned by the patient. This construct allows for simultaneous correction of malalignment while performing limb lengthening. (**d**) Monolateral rail external fixator construct. Limb lengthening is driven by the distraction rod placed between the two stable bases. The distraction rod turns in one-fourth mm increments when turned by the patient using a special wrench. This construct is mechanically disadvantageous because the pins are all in the same plane, but is much better tolerated in the femur than ring fixation

can be highly advantageous in certain circumstances, but it is especially important to assure sound biomechanics when using these devices.

Another type of external fixator that is commonly used for limb lengthening is a monolateral rail (Fig. 7.1d). Rails have the mechanical disadvantage that all pins are parallel and in the same plane and are therefore a less stable construct than a multiplanar construct. In addition, because all of the fixation is performed with half pins, there is a cantilever effect with a tendency toward deformity away from the pins (e.g., tendency toward varus in the femur as lateral pins bow apart). However, the pins and frame are manufactured to be especially robust in order to help prevent this. This tendency of the monolateral rail frames having been recognized, many of these systems have the ability to angulate either at the beginning or at the end of distraction to compensate for this tendency. The major advantage of a monolateral rail is that it is much easier for the patient to tolerate than a complete ring around the limb, especially in the femur [32–34].

DO for bone transport or lengthening combined with bone transport requires additional consideration in regard to frame construct [35]. Transport with a traditional Ilizarov fixator occurs at an intermediate ring, traditionally stainless steel, with distraction driven by square nuts such as seen in Fig. 7.2a. The intermediate ring moves along the threaded rods "rails" and drags the transport segment with it. A column of new regenerate bone forms behind the transport segment, and eventually the transport segment crosses the defect to meet the opposite bone end. Half pins can also be used to fix the transport segment to the bone and have the advantages of traversing less soft tissue and application with greater crossing angles than wires (Fig. 7.2b). However, half pins have larger dimensions and cut a larger path through soft tissue during transport and are therefore less soft-tissue-friendly in this circumstance.

Many types of rings are available with variable thicknesses and made of differing materials. These rings can be connected to threaded rods and function the same as an Ilizarov fixator. For this reason, the author refers to a construct of rings connected with threaded rods as an Ilizarov-type construct and then names the type of rings, for example, "Ilizarov-type construct with Taylor Spatial Frame rings" (Fig. 7.2c). Transport using an Ilizarov-type construct with hexapod rings is straightforward with progression along the threaded rods, but there is a big advantage in that the threaded rod segments crossing the docking site can be changed to struts at the time of docking. The struts then allow for easy adjustment of bone end alignment at the docking site without the need for strut adjustments or changes during transport.

An alternative construct that allows for bone transport and limb lengthening is the bifocal frame. Fundamentally, a bifocal transport frame distracts an osteotomy at one location and compresses the gap to bring bone ends together at another site. The lengthening is typically motored by either telescopic rods or square nuts, and compression is performed with either square nuts or struts (Fig. 7.2d). The bifocal frame with telescopic rods at the distraction site and struts at the docking site is convenient because it allows for biologically friendly distraction with flexibility to adjust docking site alignment without frame modification at the time of docking. This is a powerful construct but requires many adjustments at two levels by the patient and surgeon during the reconstruction.

A special type of bifocal frame is the "double-stacked" hexapod (Fig. 7.2e). The double-stacked frame is advantageous because there is maximum adjustability of both the regenerate bone segment and the docking site. However, this method of transport requires the greatest number of adjustments by the patient and strut changes by the surgeon, is by far the most expensive, has the most hardware obscuring radiographic evaluation, and is mechanically less rigid. For these reasons, the authors generally reserve this construct for special situations that require additional flexibility in alignment such as soft tissue coverage, deformity correction with multiple CORAs (center of rotation of angulation), or malalignment between the segments across regenerate column at the end of transport. An alternate construct is the cable transport frame. Figure 7.2f shows an example of


A
B
C
D
E
F

Fig. 7.2 Ilizarov frame variations for lengthening and transport. (**a**) Traditional Ilizarov external fixator frame with rings attached along long threaded rods that run the full length of the construct. Note the transport ring is stainless steel even when the fixation rings are carbon fiber. Square nuts are used as motors. Typically, there would be two carbon fiber rings distally or the addition of a foot plate. A foot plate was originally attached but was removed in clinic 6 weeks after transport docking. The patient has a typical dorsiflexion splint attached to the frame. (**b**) Ilizarov transport with half pins fixing the bone to the transport ring with square nuts as the motor. Distal fixation with metaphyseal wire cluster after staged foot plate removal in clinic. (**c**) Ilizarov-style transport frame with long threaded rods attached to the rings. This construct uses Taylor Spatial Frame (TSF) hexapod rings instead of Ilizarov rings. This construct allows for adjustability at the end of transport because the threaded rods can be cut and struts applied across the docking segment. (**d**) Bifocal transport frame with telescopic rods "clickers" proximal and hexapod TSF struts distally. This allows for biologically optimal cadence of 4 × 0.25 mm movements per day but great flexibility in controlling alignment at docking. This construct allows for adjustability without revising frame components but requires many more daily adjustments than in (**c**). (**e**) Double-stacked hexapod with TSF struts. Maximizes adjustability of alignment for both transport and docking segments. (**f**) Balanced cable transport external fixator frame. Allows bone transport with no pins or wires dragging through the skin. TSF struts with the shoulder bolt removed are used as motors in this example

a balanced cable transport frame with internal cables pulling the transport segment [36]. Note the absence of pins and wires in the transport segment and the attachment to a strut proximally used to motor the transport. The chosen construct for any given patient should be tailored to the specifics of the bone available for fixation, soft tissue constraints, where the osteotomy is planned, and whether lengthening and/or transport is planned.

Mechanical Modulation to Encourage Bone Formation

Altering the mechanical load on the affected limb can modulate the regenerate. Early weight-bearing has been a mainstay for encouraging better bone formation and remains a cornerstone of treatment. As discussed earlier, increasing the frequency of distraction while decreasing the amount of lengthening at each interval may shorten the external fixation index [31, 37]. However, currently available methods make greater than four incremental turns per day impractical and have not been clinically demonstrated to be of significant benefit to justify the added difficulty. Techniques such as compression after over-distraction, "pumping the regenerate," have been described but have not demonstrated clear benefit in increasing the rate of regenerate healing. In contrast, "pumping of the regenerate" can be a useful method of salvage when poor regenerate is formed early in the distraction phase. In this scenario, the transport segment is compressed back to or near its original position and then gradually distracted again. This can often encourage a greatly improved regenerate to salvage a poor start.

Dynamization, as classically described, has been used since the original descriptions by Ilizarov in order to encourage fracture healing and regenerate consolidation. In its original form, dynamization meant that the nuts holding the stable ring on one side of the fracture or regenerate were made loose and backed up by a small amount (~2 mm). This had the effect of loosening the frame and allowing a small amount of dynamic compression at the fracture site. Dynamization was performed to encourage additional callus formation or as a final stage prior to fixator removal. The process also acted as a clinical test to see how the patient felt with an unstable fixator. If they could walk without pain, then it likely meant it was safe to remove the fixator. This method is still commonly used today as is a process of dynamization where frame components are gradually removed in order to shift weight-bearing forces from the fixator to the bone. The introduction of the TSF as the first hexapod complicated the ability to dynamize the external fixator. It was no longer possible to back up and stabilize the nuts as had been possible with an Ilizarov-type fixator. However, *dynamization* continued to be a highly employed concept but with a new method of application. The hexapod could by dynamized by either removing fixation components to provide more flexibility or by unlocking the struts, which completely destabilizes the fixator across the fracture site. *Dynamization* performed by unlocking the struts is a good test of fracture and regenerate healing but is not helpful for encouraging bone formation during the consolidation process. To address this problem, there are reports of special shoulder bolts designed to allow a true axial dynamization of hexapod external fixators in the same manner that an Ilizarov frame could be dynamized, but to date these are not widely available [31]. Consequently, the exact meaning of the word *dynamization* has become somewhat confused, as the same word is used to describe very different mechanical processes. However, the principle of fixator destabilization late in the reconstruction process remains a common element of the treatment process.

More recently, there has been compelling basic science evidence that challenges the usefulness of dynamization as a method to encourage final healing. This evidence supports a new paradigm called "reverse dynamization" [38, 39]. Reverse dynamization relies on the principle that early on in fracture healing there is a soft and flexible hematoma that is converted to a cartilaginous callus. Callus formation during these early stages is encouraged by fracture micromotion, and larger amounts of relative motion of the bone ends are well tolerated. Later stages of fracture healing occur as softer bone is replaced by more rigid organized mature bone formation. This stage is sensitive to relative motion of the bone ends and is harmed by larger amounts of motion and is thus aided by greater construct stability. The reverse dynamization concept therefore advocates for making the fixator construct more stable during the consolidation phase and after the end of the initial phases of callus formation in order to optimize the speed of bony healing. Therefore, instead of removing components in late healing, the surgeon would add threaded rods or attach additional points of fixation after the initial healing stages in order to encourage final healing. Reverse dynamization is a relatively new concept and is awaiting validation from clinical data but has shown anecdotal success in the authors' experience.

The use of noninvasive physical modalities has become a popular adjuvant to encourage bone healing. One such intervention is the use of low-intensity pulsed ultrasound (US). US is theorized to modulate signal transduction at the cellular level by inducing a pressure wave [40]. US has been shown to increase callus formation during fracture healing [41]. This potential has led researchers to investigate its use during DO. A recent meta-analysis suggests that US could possibly reduce the healing index of DO by 15 days/cm in tibia defects, and it is more effective when used during distraction and early consolidation phases [42]. However, a more recent study did not show a statistical difference in reduction in treatment time, radiographic or histologic fill length, or bone density increase [43]. The limitation in interpreting efficacy of US results from the heterogeneity of patients reported, publication and selection bias, and other confounding factors.

Biological Adjuvants

The role of BMPs in osteogenesis has been previously described. Recombinant BMP-2 and BMP-7 have been used in adults as adjuvants or substitutes for bone graft. Although not approved by the US Food and Drug Administration (FDA) for DO, off-label applications have been reported for patients with poor regenerate and persistent nonunion [44].

Platelet-rich plasma (PRP) contains osteoinductive growth factors and has been investigated in combination with bone marrow grafting for bone formation in DO [45, 46]. The results of these investigations showed increased cellular activity in rats, but there was no difference in osteoblast activity. There are also no clinical data to support the use of PRP as an adjunct to improve regenerate bone formation. Anticatabolic agents (i.e., calcitonin, diphosphonates) have also been used in off-label cases in pediatric patients with poor-quality regenerate with eventual healing [47, 48]. However, there are limited data to support the efficacy of these agents, and in fact the use of an agent that retards bone turnover seems counterproductive given that callus and regenerate maturation rely on bone turnover as part of the natural healing process.

Augmentation with bone marrow aspirate concentrate (BMAC) has also been proposed. Percutaneous insertion of marrow cells has been shown to be a safe and effective approach and to accelerate bone regeneration during DO [49, 50]. This technique has also been used as an adjuvant to treat segmental long-bone defects [51]. Another study reported the use of BMAC in femur and tibia lengthening, with faster femoral than tibial healing, with no difference in the number of cells present in the concentrate. These results suggest that the effect of BMAC on the bone regenerate may be multifactorial and probably related to the local milieu at the transplanted site and not the actual number of cells. However, more studies are needed to optimize this technique.

Complications

DO with external fixation provides a reliable tool for lengthening of an extremity and treating even large bone defects [52]. However, there are significant challenges to consider. The related problems of superficial cellulitis, deep pin infection, and loosening have improved with hydroxyapatite-coated (HA) pins but remain the most common problem for both surgeon and patient [53, 54]. Pin site cellulitis causes increased pain and the need for additional clinic visits and infrequently may require hospital admission for IV antibiotics or pin removal/exchange. Most cases of pin site cellulitis are successfully treated with a short course of oral antibiotics and do not compromise the final outcome of reconstruction, but the short-term burden for both patient and surgeon is significant. Apart from cellulitis, HA half pins can mature to be painless, but discomfort around wire sites generally persists to some degree until their removal. This discomfort can lead to greater pain medication use during treatment [55].

A related concern is that irritation from points of fixation may lead to discomfort that discourages joint range of motion and may lead to joint contractures [56]. Joint contracture can also occur because of the pull on muscle-tendon units and the translocation of muscular origins that can occur during the process of DO. Joint contracture can be one of the most difficult problems to deal with during limb lengthening and bone transport. In fact, loss of motion and joint stiffness are the most likely cause for long-term problems following DO. Great care must be taken during treatment to encourage range of motion and physical therapy. In addition, early recognition and intervention for a developing contracture is an important part of the treatment.

The weight of the external fixator can be a challenge for some patients, such as the elderly, with limited strength reserve. Therefore, the weight of the external fixator construct should be considered carefully in this patient population and construct choice modified as needed.

Shortening and angulation of the regenerate is a significant complication. This occurs when the fixator is removed prior to complete consolidation of the regenerate. When this occurs, it is almost always impossible to acutely correct without an osteotomy, as the regenerate tends to rapidly consolidate in this scenario. Correction requires a return to the operating room for an osteotomy and surgical correction of angulation. This is best prevented by assuring adequate regenerate healing prior to frame removal by obtaining radiographic confirmation of healing, waiting an adequate and expected time for healing (generally no less than 1.5 months/cm in an adult), and testing with frame dynamization prior to removal.

Nonunion of the regenerate typically occurs when there has been poor compliance with the distraction process. This can be treated with bone grafting and other methods described above. Another alternative is to consider conversion to internal fixation, but it should be emphasized that this must be undertaken with great care and respect for contaminated pin and wire sites. Multiple means such as a pin holiday and antibiotic cement-coated implants can be used to help moderate this risk when this approach is necessary. However, generally speaking, conversion to internal fixation at the conclusion of limb lengthening or bone transport should be considered a salvage procedure with significant attendant risks. The exception to this is when the initial construct was applied to avoid contamination from the fixator components in the path of the staged internal fixation, in which case routine conversion has been shown to carry low risks.

Integrated Techniques

To address some of the challenges of DO with external fixation alone, methods that integrate the use of internal fixation have been proposed. Techniques include lengthening over an intramedullary nail (LON) [57–60], lengthening and then nailing (LATN) [61], transport and then nailing (TATN) [36], and lengthening over a plate (LOP). LOP has had mixed results and is generally not preferred. LON, LATN, and TATN have all proven to significantly decrease external fixation index or days in ex-fix/cm new bone (EFI). LATN and TATN have also substantially decreased the bone healing index or months/cm new bone (BHI). Disadvantages of using internal hardware include the potential for deep infection, increased surgical time, blood loss, added cost, and the added technical difficulty.

Lengthening Over a Nail (LON)

With this technique, an intramedullary nail is inserted after the corticotomy is performed. A frame is then applied after the nail is inserted with care taken to keep fixation points remote from the deep hardware. The external fixator is used to lengthen over the nail (Fig. 7.3). When the desired length is achieved, the nail is locked,

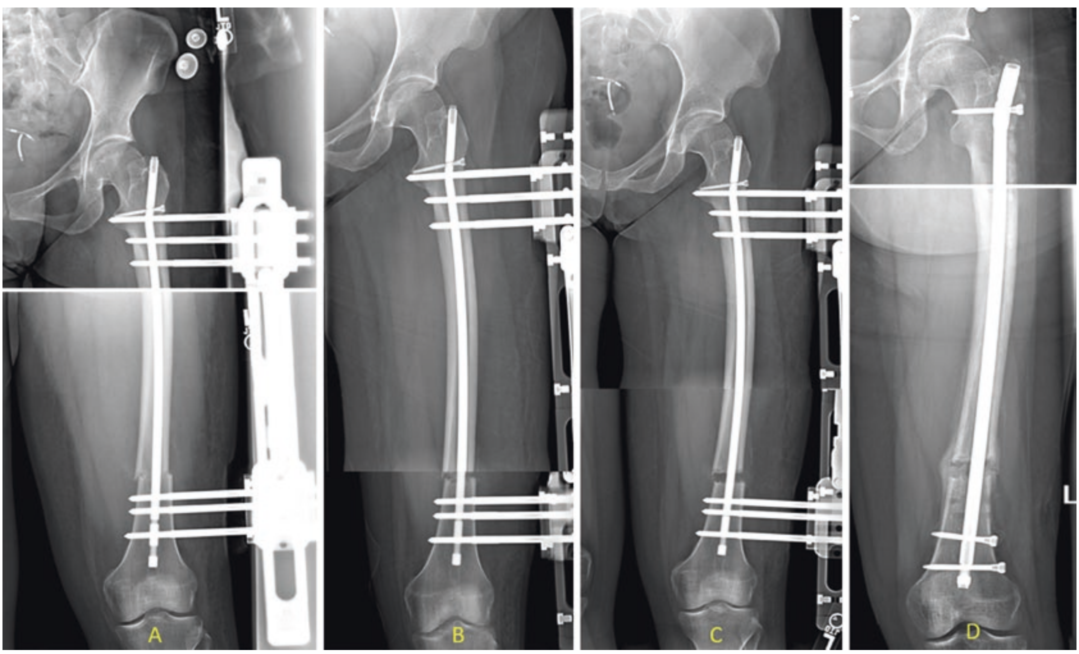

Fig. 7.3 This is an example of lengthening over a nail. This patient had residual limb length discrepancy after being treated for Perthes as a child. She had failed orthotic treatment and had persistent back pain and a limp. (**a–d**) An antegrade nail was placed in the femur with a distal corticotomy in the diaphyseal-metaphyseal junction. The limb was subsequently lengthened with a monolateral frame

and the fixator removed [57, 59, 60, 62]. EFI is decreased while providing the regenerated bone support in the consolidation phase, although the BHI is not significantly different than the classic Ilizarov technique [60, 63]. The deep infection risk must be considered, as the rate has been reported to be 14%. Another disadvantage is the need to use smaller-diameter nails to allow sliding of the bone and to allow concomitant placement of an external fixator. This may lead to suboptimal stability. Any deformity must be corrected acutely with this technique, which may compromise bone healing.

Lengthening and Then Nailing (LATN) and Transport and Then Nailing (TATN)

LATN is the technique of using a ring fixator to perform limb lengthening followed by placement of an intramedullary nail at the conclusion of the distraction phase with removal of the external fixator. The initial external fixator is constructed in such a manner that it avoids placing contaminated pins and wires in the path of the intramedullary nail that is placed later on. The regenerated bone is supported by the nail during the consolidation phase. The EFI is decreased from 45–60 days to approximately 14 days/cm, and the BHI is decreased from 1.5–2.0 to 0.9. The time in frame is therefore 75% less, with healing in 50% less time. Both LATN (Fig. 7.4) and TATN (Fig. 7.5) have shown identical results in terms of effect on EFI and BHI. One concern of using an intramedullary device after prolonged time in external fixation is the risk of deep infection. This risk, however, has been reported to be lower than 5% and as low as 0% in some studies [36, 61, 64]. This technique can be used for pure lengthening, transport, or combined cases.

Meta-analysis of the results of bone defect management indicates that integrated methods appear to be the most effective treatment for bone loss and limb length discrepancy, with LATN and TATN having significant advantages over all other methods. Because there are far more data on traditional methods, additional data on integrated methods are necessary before any solid conclusions can be reached.

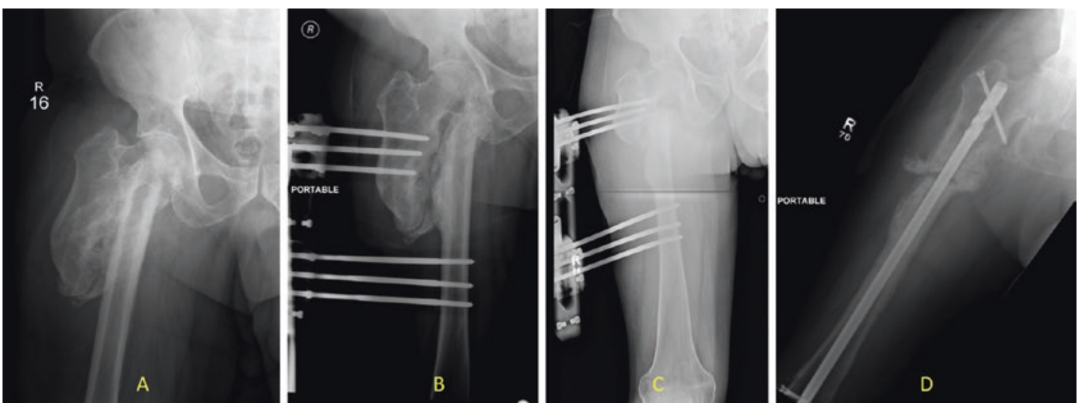

Fig. 7.4 This patient had suffered a right femur fracture treated without surgery in another country (**a**). The length of the femur was re-established using a monolateral external fixator (**b**, **c**). After length was restored, the frame was removed, and an intramedullary nail was placed (**d**)

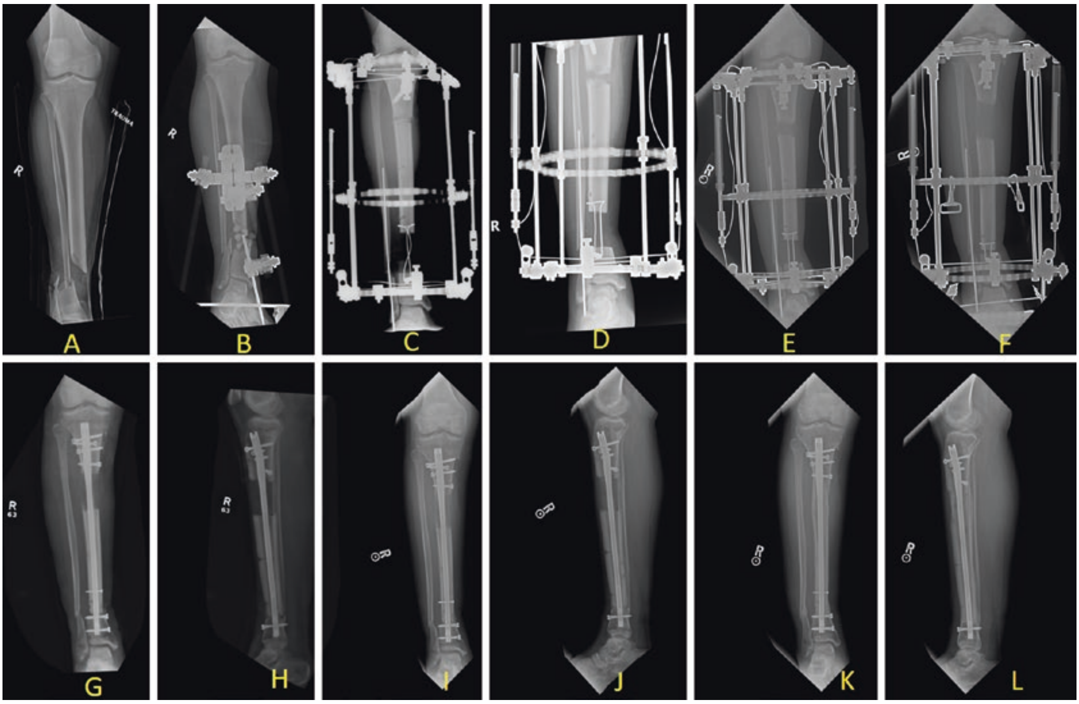

Fig. 7.5 This patient sustained a high-energy grade III B tibia fracture that underwent serial debridement (**a**, **b**). After soft tissue stabilization, a complex transport using a cable frame was used to reconstruct the residual segmental defect of the bone (**c**, **d**, **e**). After the transport segment reached the docking site (**f**), docking with placement of an intramedullary antibiotic cement-coated nail was performed, and the frame removed (**g**, **h**). Consolidation occured rapidly as seen at one month (**i**, **j**) and with final healing at 3 months after frame removal (**k**, **l**)

References

1. Aronson J, Harrison BH, Stewart CL, Harp JH Jr. The histology of distraction osteogenesis using different external fixators. Clin Orthop Relat Res. 1989;241:106–16.
2. Kojimoto H, Yasui N, Goto T, Matsuda S, Shimomura Y. Bone lengthening in rabbits by callus distraction. The role of periosteum and endosteum. J Bone Joint Surg Br. 1988;70(4):543–9.
3. Kusec V, Jelic M, Borovecki F, Kos J, Vukicevic S, Korzinek K. Distraction osteogenesis by Ilizarov and unilateral external fixators in a canine model. Int Orthop. 2003;27(1):47–52.
4. De Bastiani G, Aldegheri R, Renzi-Brivio L, Trivella G. Limb lengthening by callus distraction (callotasis). J Pediatr Orthop. 1987;7(2):129–34.
5. Dabis J, Templeton-Ward O, Lacey AE, Narayan B, Trompeter A. The history, evolution and basic science of osteotomy techniques. Strategies Trauma Limb Reconstr. 2017;12(3):169–80.
6. Yasui N, Kojimoto H, Sasaki K, Kitada A, Shimizu H, Shimomura Y. Factors affecting callus distraction in limb lengthening. Clin Orthop Relat Res. 1993;293:55–60.
7. Ilizarov GA. The tension-stress effect on the genesis and growth of tissues. Part I. The influence of stability of fixation and soft-tissue preservation. Clin Orthop Relat Res. 1989;238:249–81.
8. Codivilla A. On the means of lengthening, in the lower limbs, the muscles and tissues which are shortened through deformity. Clin Orthop Relat Res. 1904;1994(301):4–9.
9. Monticelli G, Spinelli R, Bonucci E. Distraction epiphysiolysis as a method of limb lengthening. II. Morphologic investigations. Clin Orthop Relat Res. 1981;154:262–73.
10. De Bastiani G, Aldegheri R, Renzi Brivio L, Trivella G. Limb lengthening by distraction of the epiphyseal plate. A comparison of two techniques in the rabbit. J Bone Joint Surg Br. 1986;68(4):545–9.
11. Fischgrund J, Paley D, Suter C. Variables affecting time to bone healing during limb lengthening. Clin Orthop Relat Res. 1994;301:31–7.
12. Dimitriou R, Tsiridis E, Giannoudis PV. Current concepts of molecular aspects of bone healing. Injury. 2005;36(12):1392–404.
13. Cho TJ, Kim JA, Chung CY, Yoo WJ, Gerstenfeld LC, Einhorn TA, et al. Expression and role of interleukin-6 in distraction osteogenesis. Calcif Tissue Int. 2007;80(3):192–200.

14. Kon T, Cho TJ, Aizawa T, Yamazaki M, Nooh N, Graves D, et al. Expression of osteoprotegerin, receptor activator of NF-kappaB ligand (osteoprotegerin ligand) and related proinflammatory cytokines during fracture healing. J Bone Miner Res. 2001;16(6):1004–14.
15. Liu Z, Luyten FP, Lammens J, Dequeker J. Molecular signaling in bone fracture healing and distraction osteogenesis. Histol Histopathol. 1999;14(2):587–95.
16. Farhadieh RD, Dickinson R, Yu Y, Gianoutsos MP, Walsh WR. The role of transforming growth factor-beta, insulin-like growth factor I, and basic fibroblast growth factor in distraction osteogenesis of the mandible. J Craniofac Surg. 1999;10(1):80–6.
17. Mehrara BJ, Rowe NM, Steinbrech DS, Dudziak ME, Saadeh PB, McCarthy JG, et al. Rat mandibular distraction osteogenesis: II. Molecular analysis of transforming growth factor beta-1 and osteocalcin gene expression. Plast Reconstr Surg. 1999;103(2):536–47.
18. Onishi T, Ishidou Y, Nagamine T, Yone K, Imamura T, Kato M, et al. Distinct and overlapping patterns of localization of bone morphogenetic protein (BMP) family members and a BMP type II receptor during fracture healing in rats. Bone. 1998;22(6):605–12.
19. Weiss S, Baumgart R, Jochum M, Strasburger CJ, Bidlingmaier M. Systemic regulation of distraction osteogenesis: a cascade of biochemical factors. J Bone Miner Res. 2002;17(7):1280–9.
20. Yonezawa H, Harada K, Ikebe T, Shinohara M, Enomoto S. Effect of recombinant human bone morphogenetic protein-2 (rhBMP-2) on bone consolidation on distraction osteogenesis: a preliminary study in rabbit mandibles. J Craniomaxillofac Surg. 2006;34(5):270–6.
21. Carter DR, Beaupre GS, Giori NJ, Helms JA. Mechanobiology of skeletal regeneration. Clin Orthop Relat Res. 1998;(355 Suppl):S41–55.
22. Pavalko FM, Norvell SM, Burr DB, Turner CH, Duncan RL, Bidwell JP. A model for mechanotransduction in bone cells: the load-bearing mechanosomes. J Cell Biochem. 2003;88(1):104–12.
23. D'Angelo F, Tiribuzi R, Armentano I, Kenny JM, Martino S, Orlacchio A. Mechanotransduction: tuning stem cells fate. J Funct Biomater. 2011;2(2):67–87.
24. Gjorevski N, Nelson CM. Bidirectional extracellular matrix signaling during tissue morphogenesis. Cytokine Growth Factor Rev. 2009;20(5-6):459–65.
25. Burger EH, Klein-Nulen J. Responses of bone cells to biomechanical forces in vitro. Adv Dent Res. 1999;13:93–8.
26. Klein-Nulend J, Bakker AD, Bacabac RG, Vatsa A, Weinbaum S. Mechanosensation and transduction in osteocytes. Bone. 2013;54(2):182–90.
27. Tan SD, de Vries TJ, Kuijpers-Jagtman AM, Semeins CM, Everts V, Klein-Nulend J. Osteocytes subjected to fluid flow inhibit osteoclast formation and bone resorption. Bone. 2007;41(5):745–51.
28. Vezeridis PS, Semeins CM, Chen Q, Klein-Nulend J. Osteocytes subjected to pulsating fluid flow regulate osteoblast proliferation and differentiation. Biochem Biophys Res Commun. 2006;348(3):1082–8.
29. Aronson J, Johnson E, Harp JH. Local bone transportation for treatment of intercalary defects by the Ilizarov technique. Biomechanical and clinical considerations. Clin Orthop Relat Res. 1989;243:71–9.
30. Ilizarov GA. Clinical application of the tension-stress effect for limb lengthening. Clin Orthop Relat Res. 1990;250:8–26.
31. Ilizarov GA. The tension-stress effect on the genesis and growth of tissues: part II. The influence of the rate and frequency of distraction. Clin Orthop Relat Res. 1989;239:263–85.
32. Prince DE, Herzenberg JE, Standard SC, Paley D. Lengthening with external fixation is effective in congenital femoral deficiency. Clin Orthop Relat Res. 2015;473(10):3261–71.
33. Szymczuk VL, Hammouda AI, Gesheff MG, Standard SC, Herzenberg JE. Lengthening with monolateral external fixation versus magnetically motorized intramedullary nail in congenital femoral deficiency. J Pediatr Orthop. 2019;39(9):458–65.
34. Arora S, Batra S, Gupta V, Goyal A. Distraction osteogenesis using a monolateral external fixator for infected non-union of the femur with bone loss. J Orthop Surg (Hong Kong). 2012;20(2):185–90.
35. Quinnan SM. Segmental bone loss reconstruction using ring fixation. J Orthop Trauma. 2017;31(Suppl 5):S42–S6.
36. Quinnan SM, Lawrie C. Optimizing bone defect reconstruction-balanced cable transport with circular external fixation. J Orthop Trauma. 2017;31(10):e347–e55.
37. Mizuta H, Nakamura E, Kudo S, Maeda T, Takagi K. Greater frequency of distraction accelerates bone formation in open-wedge proximal tibial osteotomy with hemicallotasis. Acta Orthop Scand. 2004;75(5):588–93.
38. Glatt V, Tepic S, Evans C. Reverse dynamization: a novel approach to bone healing. J Am Acad Orthop Surg. 2016;24(7):e60–1.
39. Glatt V, Evans CH, Tetsworth K. A concert between biology and biomechanics: the influence of the mechanical environment on bone healing. Front Physiol. 2016;7:678.
40. Harrison A, Lin S, Pounder N, Mikuni-Takagaki Y. Mode & mechanism of low intensity pulsed ultrasound (LIPUS) in fracture repair. Ultrasonics. 2016;70:45–52.
41. Claes L, Willie B. The enhancement of bone regeneration by ultrasound. Prog Biophys Mol Biol. 2007;93(1-3):384–98.
42. Raza H, Saltaji H, Kaur H, Flores-Mir C, El-Bialy T. Effect of low-intensity pulsed ultrasound on distraction osteogenesis treatment time: a meta-analysis of randomized clinical trials. J Ultrasound Med. 2016;35(2):349–58.
43. Lou S, Lv H, Li Z, Tang P, Wang Y. Effect of low-intensity pulsed ultrasound on distraction osteogenesis: a systematic review and meta-analysis of

randomized controlled trials. J Orthop Surg Res. 2018;13(1):205.
44. Burkhart KJ, Rommens PM. Intramedullary application of bone morphogenetic protein in the management of a major bone defect after an Ilizarov procedure. J Bone Joint Surg Br. 2008;90(6):806–9.
45. Kawasumi M, Kitoh H, Siwicka KA, Ishiguro N. The effect of the platelet concentration in platelet-rich plasma gel on the regeneration of bone. J Bone Joint Surg Br. 2008;90(7):966–72.
46. Kitoh H, Kitakoji T, Tsuchiya H, Mitsuyama H, Nakamura H, Katoh M, et al. Transplantation of marrow-derived mesenchymal stem cells and platelet-rich plasma during distraction osteogenesis--a preliminary result of three cases. Bone. 2004;35(4):892–8.
47. Kiely P, Ward K, Bellemore CM, Briody J, Cowell CT, Little DG. Bisphosphonate rescue in distraction osteogenesis: a case series. J Pediatr Orthop. 2007;27(4):467–71.
48. Kokoroghiannis C, Papaioannou N, Lyritis G, Katsiri M, Kalogera P. Calcitonin administration in a rabbit distraction osteogenesis model. Clin Orthop Relat Res. 2003;415:286–92.
49. Kitoh H, Kawasumi M, Kaneko H, Ishiguro N. Differential effects of culture-expanded bone marrow cells on the regeneration of bone between the femoral and the tibial lengthenings. J Pediatr Orthop. 2009;29(6):643–9.
50. Gessmann J, Koller M, Godry H, Schildhauer TA, Seybold D. Regenerate augmentation with bone marrow concentrate after traumatic bone loss. Orthop Rev (Pavia). 2012;4(1):e14.
51. Petri M, Namazian A, Wilke F, Ettinger M, Stubig T, Brand S, et al. Repair of segmental long-bone defects by stem cell concentrate augmented scaffolds: a clinical and positron emission tomography--computed tomography analysis. Int Orthop. 2013;37(11):2231–7.
52. Kadhim M, Holmes L Jr, Gesheff MG, Conway JD. Treatment options for nonunion with segmental bone defects: systematic review and quantitative evidence synthesis. J Orthop Trauma. 2017;31(2):111–9.
53. Moroni A, Faldini C, Marchetti S, Manca M, Consoli V, Giannini S. Improvement of the bone-pin interface strength in osteoporotic bone with use of hydroxyapatite-coated tapered external-fixation pins. A prospective, randomized clinical study of wrist fractures. J Bone Joint Surg Am. 2001;83-A(5):717–21.
54. Moroni A, Orienti L, Stea S, Visentin M. Improvement of the bone-pin interface with hydroxyapatite coating: an in vivo long-term experimental study. J Orthop Trauma. 1996;10(4):236–42.
55. Quinnan SM. Definitive management of distal tibia and simple plafond fractures with circular external fixation. J Orthop Trauma. 2016;30(Suppl 4):S26–32.
56. Pettine KA, Chao EY, Kelly PJ. Analysis of the external fixator pin-bone interface. Clin Orthop Relat Res. 1993;293:18–27.
57. Kocaoglu M, Eralp L, Kilicoglu O, Burc H, Cakmak M. Complications encountered during lengthening over an intramedullary nail. J Bone Joint Surg Am. 2004;86-A(11):2406–11.
58. Kristiansen LP, Steen H. Lengthening of the tibia over an intramedullary nail, using the Ilizarov external fixator. Major complications and slow consolidation in 9 lengthenings. Acta Orthop Scand. 1999;70(3):271–4.
59. Paley D, Herzenberg JE, Paremain G, Bhave A. Femoral lengthening over an intramedullary nail. A matched-case comparison with Ilizarov femoral lengthening. J Bone Joint Surg Am. 1997;79(10):1464–80.
60. Watanabe K, Tsuchiya H, Sakurakichi K, Yamamoto N, Kabata T, Tomita K. Tibial lengthening over an intramedullary nail. J Orthop Sci. 2005;10(5):480–5.
61. Rozbruch SR, Kleinman D, Fragomen AT, Ilizarov S. Limb lengthening and then insertion of an intramedullary nail: a case-matched comparison. Clin Orthop Relat Res. 2008;466(12):2923–32.
62. Song HR, Oh CW, Mattoo R, Park BC, Kim SJ, Park IH, et al. Femoral lengthening over an intramedullary nail using the external fixator: risk of infection and knee problems in 22 patients with a follow-up of 2 years or more. Acta Orthop. 2005;76(2):245–52.
63. Alrabai HM, Gesheff MG, Conway JD. Use of internal lengthening nails in post-traumatic sequelae. Int Orthop. 2017;41(9):1915–23.
64. Siebenrock KA, Schillig B, Jakob RP. Treatment of complex tibial shaft fractures. Arguments for early secondary intramedullary nailing. Clin Orthop Relat Res. 1993;290:269–74.

External Fixators for Deformity Correction

8

Austin T. Fragomen, Kristin S. Livingston, and Sanjeev Sabharwal

Introduction

External fixators have transformed the art of deformity correction by incorporating the 4D technology of gradual correction over time into the operative strategy. Computer navigation through the use of hexapod mechanics has further advanced the surgeon's ability to realign malunited fractures and nonunions safely and reproducibly. External fixator (frame) stability is paramount to successfully controlling the fixated bone fragments and performing accurate deformity correction with reliable healing. The biomechanics begin with a thorough evaluation of the patient and the radiographs to generate a strong preoperative plan. A stable frame is applied in the operating room, and minimally invasive surgery is performed when possible. Careful follow-up is done to check the frame integrity, the adjacent soft tissues, and the radiographs. Problems and obstacles are addressed.

Several studies have been cited in this chapter but only represent a fraction of the work that has been done in this field, particularly at the Ilizarov Scientific Center in Kurgan, Russia, where extensive research has taken place for decades. The field of deformity correction continues to evolve as we better understand which deformities can be corrected acutely and which require a more gradual approach. While internal lengthening technology has dominated femur limb reconstruction in recent years, there will always be a role for circular frames in the armamentarium of limb deformity surgeons.

Why Circular Fixation?

The field of limb lengthening and deformity correction has entered into an era of rapid advancement evidenced by the emergence of multiple new hexapod external fixators and the magnetic internal lengthening and compression nail. Several major orthopedic equipment companies have committed considerable time and resources to improve upon the Taylor Spatial Frame (TSF) (Smith & Nephew, Memphis, TN, USA), itself an evolution from the traditional, all-wire, "Ilizarov apparatus" [1]. Any discussion about the biomechanics of external fixation as it applies to

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post-traumatic deformity correction needs to first address the following question: why use circular fixation for deformity correction? The benefits of circular external fixators are numerous and were best described by Dr. Gavriil Ilizarov himself:

> Deformities of the long bones are often accompanied by limb shortening. The traditional methods for eliminating severe deformities are traumatic because they do not include the gradual stretching of the shortened soft tissues on the concave side of the deformity. When correcting such a deformity the surgeon must resect a wedge-shaped segment of bone to avoid excessive traction on the soft tissues, vessels, and nerves. This resection can lead to even greater limb length inequality. We have developed a therapeutic strategy whereby the deformity is eliminated by the gradual correction of angulation and malrotation of the bone after corticotomy combined with slow elongation of the shortened soft tissues on the deformity's concave side. The surgical intervention is usually performed in a percutaneous manner, lessening trauma to soft tissues and bone. [2]

The gradual correction of a long bone deformity was a revolutionary concept that swept through Russia and Italy and has recently gained momentum in the AO dominated orthopedic trauma community in the United States. Gradual correction utilizes the power of distraction osteogenesis, avoiding the need for supplemental bone grafting and additional procedures to equalize limb length discrepancies, and provides the opportunity for "fine-tuning" bony alignment during deformity correction (postoperative adjustability). Previously traumatized tissues are spared large incisions and periosteal stripping, and the endosteal blood supply of the underlying bone is safeguarded from reaming. Transfixion elements (half pins [Schanz screws] and wires) can bypass infected zones, avoiding biofilms, while still providing stable limb fixation. Large deformities (>12° angulation) can be safely rectified with this approach [3]. Nonunions can be treated with sustained controlled compression applied across the bone ends at regular intervals postoperatively. Early weight-bearing is encouraged and helps with osteogenesis [1]. Circular external fixators are very versatile and are particularly well suited to deformity correction [1], achieving high union rates and restoration of limb length [4–8] (Tables 8.1 and 8.2) [3, 9–25].

Deformity Assessment and Strategy

Although this article will review many details of circular external fixator assembly and management, every single frame applied should be customized to the particular needs of the patient being treated. The most fundamental part of this assessment is the radiological analysis. This requires knowledge of the normal alignment parameters that have been well outlined by Paley and Tetsworth [26, 27]. The apex (or apices) of the deformity is localized, and the magnitude and direction of each deformity is quantified. The current limb length discrepancy (LLD) is measured, and the expected length gained from the angular correction alone is calculated. Residual LLD (need for additional lengthening) is then calculated. Distraction osteogenesis applied to angular correction requires some lengthening of the bone (even when using a dome osteotomy), so this method is best suited for deformities associated with shortening of the involved extremity [2]. Osteopenia from disuse or metabolic etiology is often noted in patients with long-standing malunions or nonunions and may require additional bony fixation (Fig. 8.1a–c) [28, 29]. Physical exam will help assess the rotational profile, and a computed tomography (CT) version study can be obtained to reveal a more accurate torsional measurement of the femur and tibia. This powerful combination of physical exam and imaging is required for several other aspects of the planning. The osteotomy site needs to be selected and should be done as near to the apex of deformity as possible, through less sclerotic bone (adjacent to the malunion), and at a site where the surrounding soft tissues are healthy or the underlying bone has been covered by a local or free tissue transfer [30, 31] (Fig. 8.2a–e). Sagittal correction of the tibia and femur needs to compliment the range of motion of the knee joint so that the final result is restoration of full extension (without hyperextension) and as much flexion as possible. Stabilization of varus-valgus knee

Table 8.1 Circular external fixation for deformity correction

Study	Limbs (*n*)	Frame type	Osteotomy/ nonunion repair for deformity	Initial union (%)	Major complication/ sequela (no.)	Conclusion
Paley [9] 1989	25	Ilizarov	Tibia nonunion with defect	68	1	Final union 100%. A good bone result does not guarantee good function
Tetsworth [10] 1994	28	Ilizarov	Tibia and femur	100	2	Accuracy of correction improves with experience
Shtarker [11] 2002	14	Ilizarov	Tibia (PTO) and femur (DFO)	100	0	Accurate one-stage treatment of rotation and varus/valgus deformity
Sen [12] 2003	11	Ilizarov	Tibia (SMO)	100	1	Great solution for poor soft tissue and LLD
Chaudhary [13] 2007	27	TSF	Tibia, femur, knee, ankle	97	0	TSF has simplified deformity correction
Marangoz [14] 2008	22	TSF	Femur	91	0	Final union 100%. Monitor for knee stiffness and subluxation
Rozbruch [15] 2008	38	TSF	Tibia nonunion (50% confirmed FRI)	71	2	Final union 95%. Infection correlated with failure
Rozbruch [16] 2010	122	TSF	Tibia	100	1	Accurate and reliable method
Horn [17] 2011	52	TSF	Tibia (SMO)	96	2	Final union 100% Accurate and useful for poor soft tissue. Simplifies future fusion or replacement
Sokucu [18] 2013	50	TSF	Tibia (PTO) and femur (DFO)	96	2	Hexapod very accurate in translation and rotation
Arvesen [19] 2017	37	TSF	Tibia nonunion (distal)	86	2	Final union 94%. Accurate and safe
Fragomen [3] 2018	138	TSF and monolateral	Tibia (PTO)	100	0	TSF highly accurate for all varus and torsion. Monolateral accurate for small varus

TSF Taylor Spatial Frame (Smith & Nephew, Memphis, TN, USA), *PTO* proximal tibial osteotomy, *DFO* distal femoral osteotomy, *SMO* supramalleolar osteotomy, *LLD* limb length discrepancy, *FRI* fracture-related infection

instability through tensioning of the LCL, hemiepiphyseal elevation, or correction of knee recurvatum should be considered.

The Ilizarov method requires careful patient selection; one needs to take patient's physiologic age and comorbidities into account [2]. Older patients will heal more slowly and may be better candidates for the lengthening and then nailing (LATN) procedure following deformity correction whereby an intramedullary nail is inserted through the immature regenerate rapidly advancing the mineralization rate (lower bone healing index) [32]. Advanced age typically necessitates additional wires to maintain stability for a protracted treatment course. Aging bone, with its decreased cortical thickness and increased porosity, suffers higher strain with equivalent loads and thus has increased yielding volume at the pin-bone/wire-bone interface and in "finite element analysis" showed that although adding more pins decreases pin site yielding in young or middle-aged patients, adding a pin did not decrease pin site yielding in old age suggesting that half pins are uniquely predisposed to failure in patients with osteoporosis [29, 33] (Fig. 8.3a–c). Patients with osteoporosis may be candidates for the administration of teriparatide (recombinant parathyroid hormone) which may enhance bony union

Table 8.2 Clinical comparisons of external fixators for deformity correction

Study	Limbs (*n*)	Frame type	Osteotomy/ nonunion repair for deformity	Initial union (%)	Major complication/ sequela (frame, % or no.)	Conclusion
Manner [20] 2007	208	TSF vs. Ilizarov	Tibia and femur	–	–	Hexapod more accurate and has advantage in multiplanar deformity
Dammerer [21] 2011	135	TSF (H) vs. Ilizarov (I) vs. monolateral (M)	Tibia and femur	100	H, 3.7% I, 5.7% M, 8.7%	Hexapod more accurate, faster correction, less complications
Eren [22] 2013	171	Hexapod (H) vs. Ilizarov (I) type	Mixed	–	H, 7 I, 5	Hexapod more accurate and rapid correction. Ilizarov better BHI
Lark [23] 2013	54	TSF vs. Ilizarov	Tibia malunion, nonunion	93	1	No difference in radiographic outcome
Solomin [24] 2014	123	Ilizarov vs. Ortho-SUV	Femur	100	–	Hexapod more accurate and lower EFI. No difference in BHI
Reitenbach [25] 2016	53	TSF (H) vs. Ilizarov (I)	Femur and tibia	100	H, 2 I, 2	Hexapod lower EFI, less final LLD, higher SF-36. Similar BHI

Ortho-SUV (S.H. Pitkar Orthotools Pvt. Ltd., Pune, India)
TSF Taylor Spatial Frame (Smith & Nephew, Memphis, TN, USA), *H* hexapod frame, *I* Ilizarov apparatus, *BHI* bone healing index, *LLD* limb length discrepancy, *EFI* external fixation index

while also treating the underlying bone pathology [34]. Vascular compromise may be a relative contraindication to external fixation whether from venous congestion with severe chronic lymphedema or from arterial insufficiency which can predispose to chronic pin infection and osteomyelitis (Fig. 8.4). Uncontrolled diabetes is a contraindication until safe blood glucose levels can be obtained. Neuropathy is often associated with compromised bone healing [35, 36] and requires additional fixation for frame stability.

The soft tissue condition will affect the ability of the skin to resist infection and its capacity to tolerate gradual stretching. Scarred soft tissues can entrap nerves that lie in the concavity of the deformity which are particularly susceptible to traction injury. Consideration needs to be given to prophylactic neurolysis at the time of frame application to prevent neurologic compromise. Tarsal tunnel and peroneal nerve releases are common prophylactic procedures in a limb deformity practice [37, 38]. Patients who have suffered from compartment syndrome may have residual necrotic muscle in the compartments; great caution is needed to avoid traversing these tissues with wires, because a simple pin infection can rapidly engulf the entire compartment in suppuration.

Type of Circular Frame

This chapter will primarily focus on deformity correction using circular fixators. Ilizarov popularized the use of circular external fixators with tensioned fine wires to solve a multitude of osseous dilemmas that presented to his clinic. The institute in Kurgan performed dozens of animal experiments documenting the mechanics and biology of his technique [1]. The Ilizarov apparatus was received with much enthusiasm, and as its popularity spread throughout the world, the technique and the hardware morphed with the inclusion of half pins and thicker rings, changing the biomechanics of the frames. Today, most circular fixators include both half pins and tensioned wires and are referred to as hybrid fixation frames by many authors. Beyond the distinction by fixation elements, frames can be further separated by ring connection elements into classic (utilizing

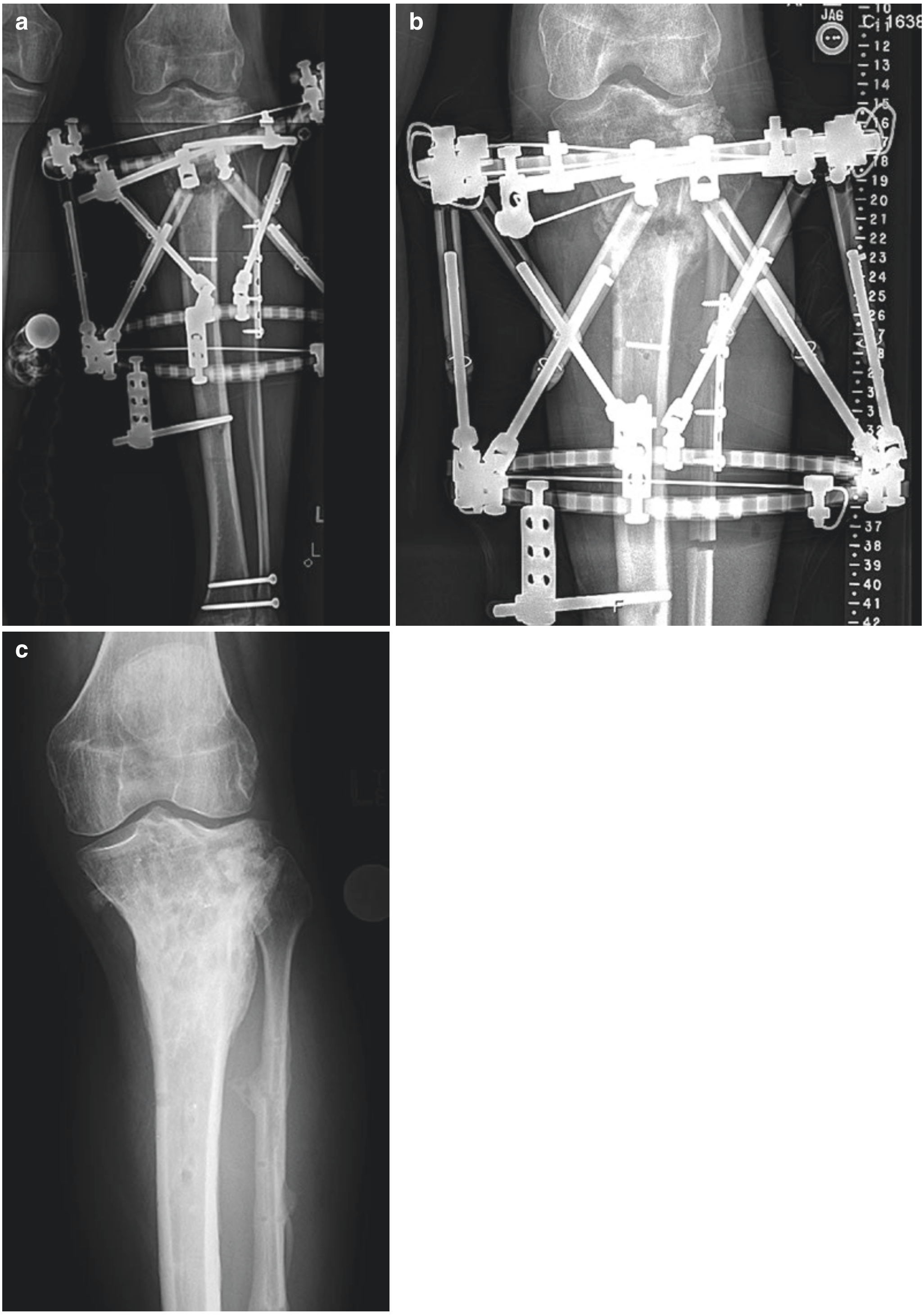

Fig. 8.1 (**a**) This 50-year-old male with a post-traumatic valgus deformity after plateau fracture was treated with circular fixation and an osteotomy. A typical construct of one tensioned wire and two half pins was used [28], but the fixation loosened in the bone and lost control of the proximal fragment. (**b**) The fixation was revised to four tensioned wires and two new half pins with excellent stability [29]. (**c**) Union was achieved with the more rigid frame

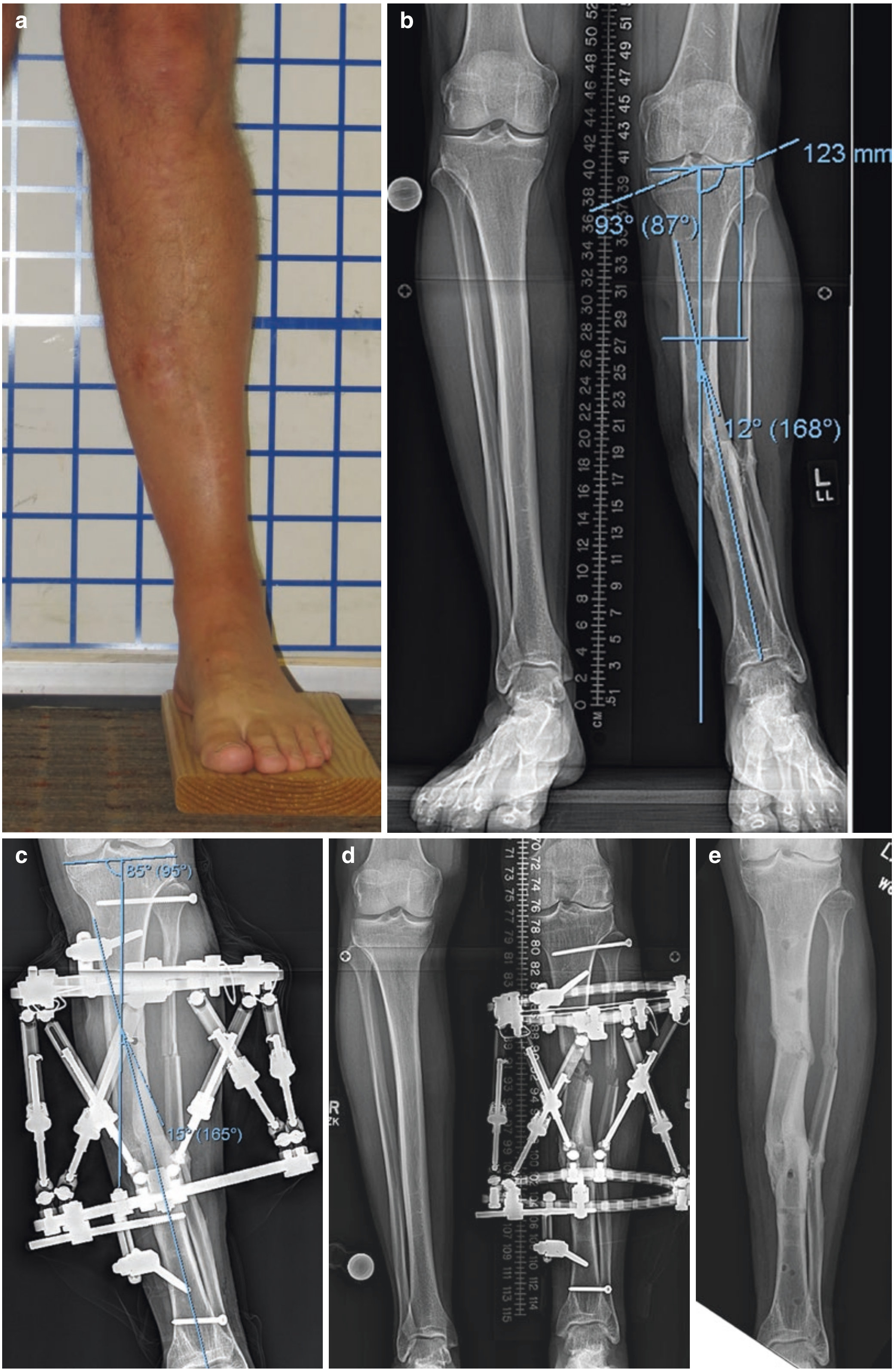
a
b
123 mm
93° (87°)
12° (168°)
c
85° (95°)
15° (165°)
d
e

Fig. 8.2 (**a**) The mid-leg area has atrophic skin with little resilience and is a poor site for osteotomy. (**b**) The malunion is seen in the mid-diaphysis where the bone is sclerotic and the soft tissue is compromised. The osteotomy site is selected proximal to the malunion, 123 mm distal to the joint line. (**c**) The initial hexapod frame is seen with two wires and two half pins proximally and three half pins distally. The fixation elements are positioned distal to the site of a future knee replacement to prevent contamination. The proximal and distal fibula are stabilized with screws across all four cortices. (**d**) At the end of the deformity correction and lengthening stage, the immature regenerate is seen clearly, and the lower limb alignment is ideal. (**e**) At final consolidation the patient is full weight-bearing and pain-free. The proximal two pin sites show evidence of loosening suggesting that more stability, in the form of another ring or fixation further away from the ring, would have been biomechanically ideal

rigid threaded rods) and hexapod (using telescopic struts based on the Gough-Stewart platform). Monolateral frames were popular for addressing simple femoral deformity and for lengthening over a nail [39, 40], but the highly accurate internal lengthening nails have come to dominate that indication. To add to the confusion, the term "hybrid frame" in the United States came to refer to a monolateral frame with a ring connected to it. These frames have been mostly abandoned and will not be discussed.

Circular External Fixator Stability

Stable fixation will improve the efficiency of the external fixator and the bone's ability to mirror the frame's movements, thus improving the accuracy of the final correction. Frame stability is associated with superior osteogenesis, pain reduction, infection eradication, and improved weight-bearing [41]. The optimal amount of stability in external fixation is not known, but this chapter will review practical frame configurations that have proven successful at straightening deformity and achieving bony union, reproducibly.

Limb stability is achieved through controlling two variables: frame stability and bone contact at the osteotomy/nonunion site. The rigidity of a circular frame is affected by many factors: ring size, ring connections, and bone fixation elements. Smaller diameter rings are more rigid than larger ones [42, 43]. Ilizarov explains that, "A ring should be 1.5–2 cm larger than the maximum diameter of the limb at each level of fixation. A ring that exceeds this amount decreases rigidity of fixation and diminishes ambulatory capacity" [1]. If swelling is expected, then a larger ring diameter can be employed. In large deformities, hexapod frames may require bigger diameter rings to avoid contact between the struts and the skin. One must consider not only the initial position of the struts but the final position as well to prevent impingement (Fig. 8.5a–c). Ring thickness also affects stability with the thinner classic Ilizarov rings being susceptible to undesirable deflection than the thicker hexapod rings. Open rings (two-third rings), which are helpful around the knee joint and the foot, are far less stable than closed rings of the same diameter [44, 45]. As wires are tensioned on the open rings, the ring contracts (deflects), and sequential tensioning of additional wires will loosen the wire that was tensioned first. The rigidity of open rings can be greatly improved by stacking two open rings [46] (Fig. 8.6). In cases where small periarticular fragments preclude the use of stacked rings or where fixation is felt to be poor, the adjacent joint can be spanned with an additional ring providing much improved stability [47] (Fig. 8.7a–c).

Ring connections vary quite a bit in this current era. The classic stainless steel threaded rods perform best when they are shorter and more numerous with at least four connections providing adequate stability. For deformity correction, hinges can be used but are subject to connecting rod bending when stiff deformities are being pushed and pulled. Although hexapod struts are available from many vendors, the TSF struts have been most widely tested and are significantly stiffer than threaded rods particularly in torsion and bending [48, 49]. Most deformity correction is carried out with hexapod struts due to their enhanced accuracy and ease of use when compared with the classic Ilizarov frame elements

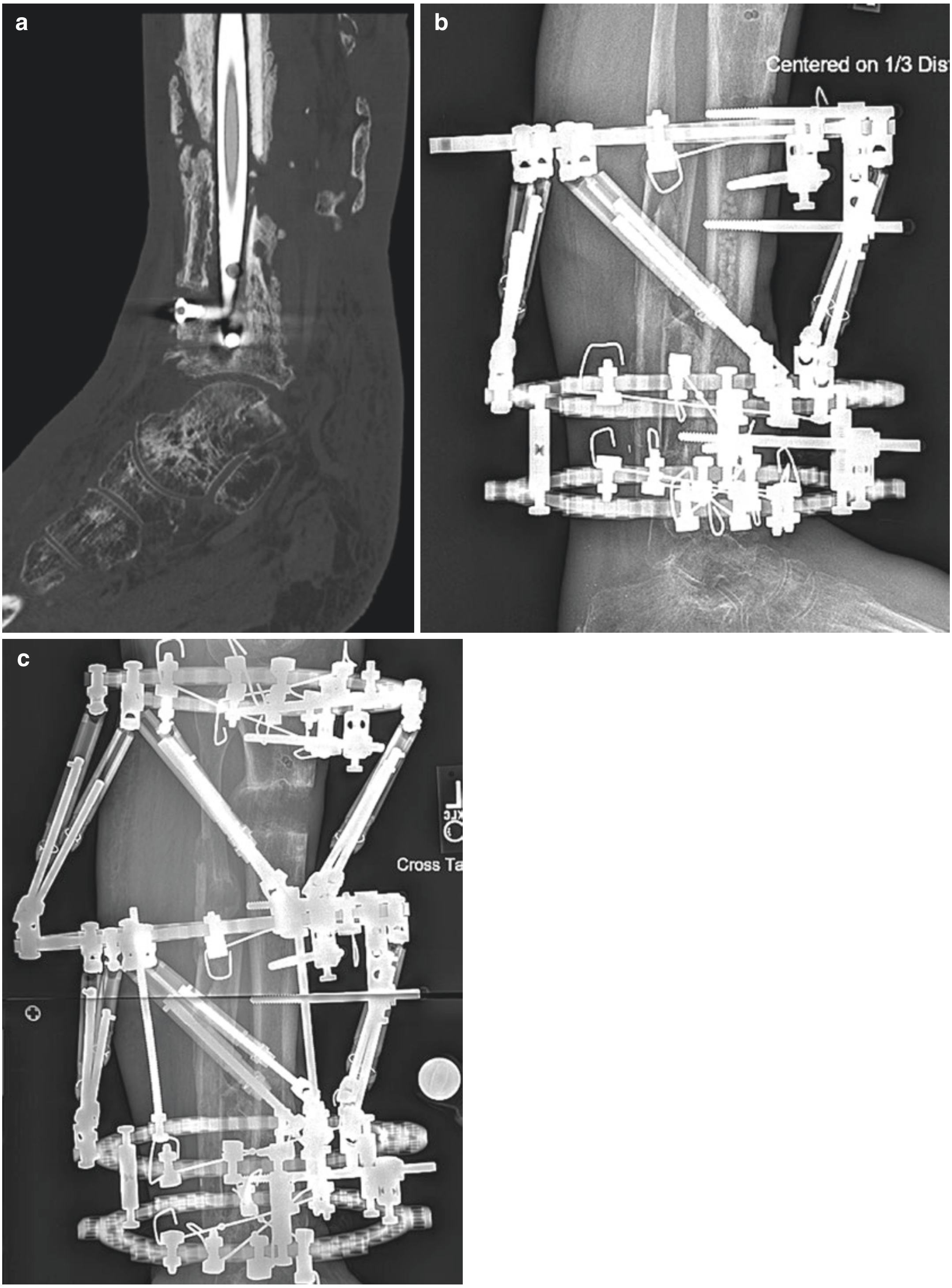

Fig. 8.3 (**a**) This computed tomography of an infected tibial nonunion shows the poor bone quality of the distal fragment. Half pin fixation is typically compromised in this type of bone with rapid loosening a common occurrence. (**b**) Strong distal fixation was established with multiple tensioned wires off of a two-ring, ring block using only one-half pin. (**c**) After completing docking and compression at the nonunion site, the rings were locked with two connecting rods. The proximal tibial osteotomy was performed with extra fixation (four wires, two half pins) to accommodate the osteopenic bone and complete this bone transport

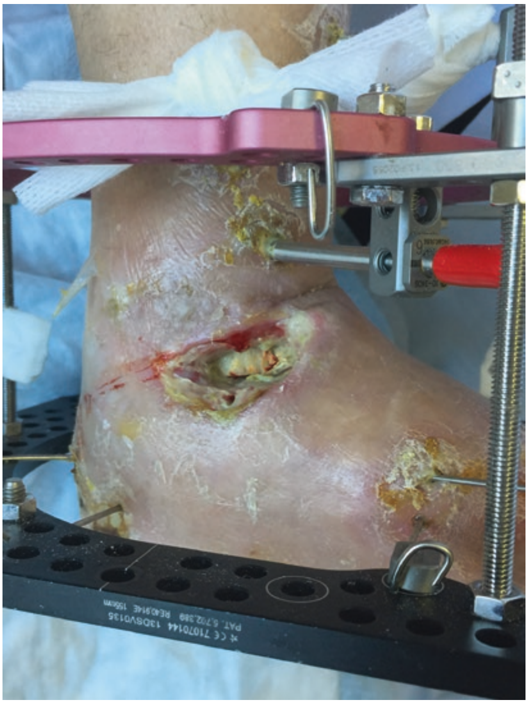

Fig. 8.4 This patient suffered from diabetes and peripheral vascular disease that had been optimized preoperatively. Her incisions broke down with necrotic bone visible through the wound. She went on to a transtibial amputation

[22, 25] (see Table 8.2). TSF struts have been found to exhibit a "shuck" or laxity when not loaded in either compression or tension [48], the significance of which is unknown with respect to bone healing. The perception among many surgeons is that locking the frame with additional rigid rods, which eliminates the excess motion in the universal hinges, will improve bone healing (Fig. 8.8a–e) [50]. This "reverse dynamization" has been shown to improve osteosynthesis in animal studies [51] (Table 8.3) [48, 49, 52–58]. Frame dynamization can still be employed later in the consolidation period to prepare for fixator removal.

The manipulation of the bone fixation elements has the greatest impact on controlling motion at the bone interface [49]. The Ilizarov frame exhibits a nonlinear stress-strain curve on axial load testing [43, 59]. This finding simply means that as the bone is loaded evenly (stressed), the tension increases in the wires and they get stiffer (less strain) and give a nonlinear or sloped increase in resistance. This nonlinearity is more profound when the wires start at a lower tension as they have more room to stiffen before reaching the point of failure or plastic deformation [60]. The nonlinearity is referred to as the "trampoline effect" and is thought to improve osteogenesis. By contrast, half pins give a linear resistance (strain) when stressed. Most hexapod frames utilize a combination of both tensioned wires and half pins. While the half pins eliminate the trampoline effect, they improve frame stability significantly [28, 61], making the larger working distances required to fit struts between the rings more secure (see Fig. 8.5a). The diameter of the pin is another very important factor in frame construction [62]. Materials matter, as a stainless steel half pin has almost double the rigidity of an equivalent titanium pin [62]. Furthermore, finite element analysis has shown that titanium half pins have significantly larger volumes of bone yielding compared to stainless steel pins, thus making stainless steel half pins much less likely to loosen [29]. The mechanics of hybrid fixation hexapod frames differ from all-wire Ilizarov frames [63], but it is not clear which is better for bone healing; while one study condemns the classic all-wire frame for allowing increased shear [48], another shows it yields a better regenerate [22]. All studies agree that when additional half-pins or "drop" wires are suspended off of the ring using cubes or connectors, the overall frame stability greatly improves [64, 65] although not nearly as much as adding an additional ring (ring-block) [66] (see Figs. 8.3b and 8.8c). The distribution of pins and wires should be multiplanar and should include fixation in the plane of the deformity correction. For instance, a frame correcting a varus deformity should have pins in the concavity on either side of the osteotomy for maximum efficiency (see Fig. 8.5b). One of the greatest advantages of the half pins over wires is the ability to place fixation from anterior to posterior thus controlling the sagittal plane. This is particularly helpful for the correction of procurvatum deformity or the prevention of deformity

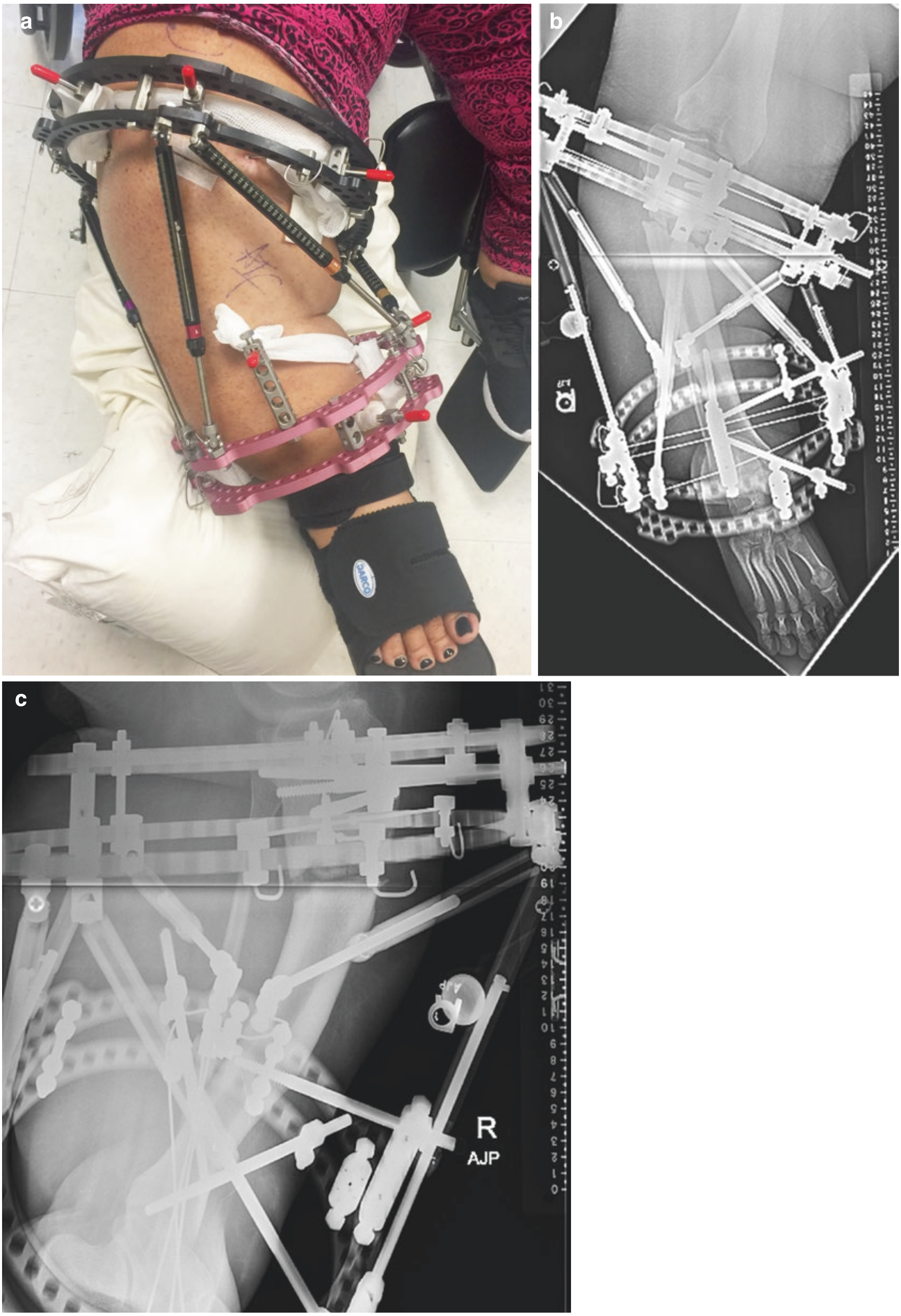
a
DARCO
b
c
R
AJP

Fig. 8.5 (**a**) This severe deformity in a large-sized patient required extra fixation including double-stacked rings and three pins-three wires proximally and distally. These were the smallest possible rings that would minimize contact between the struts and the skin. The distance between the two ring blocks is the “working distance” and is long in this case. (**b**) A radiograph of the same patient pre-correction shows an open ring bolted to a closed ring proximally to improve stability. Half pins have been inserted from medially on both the proximal and distal ring blocks to control the concavity of the deformity. (**c**) This lateral radiograph shows the anterior to posterior direction of the half pins to gain control of the sagittal plane for correction of procurvatum

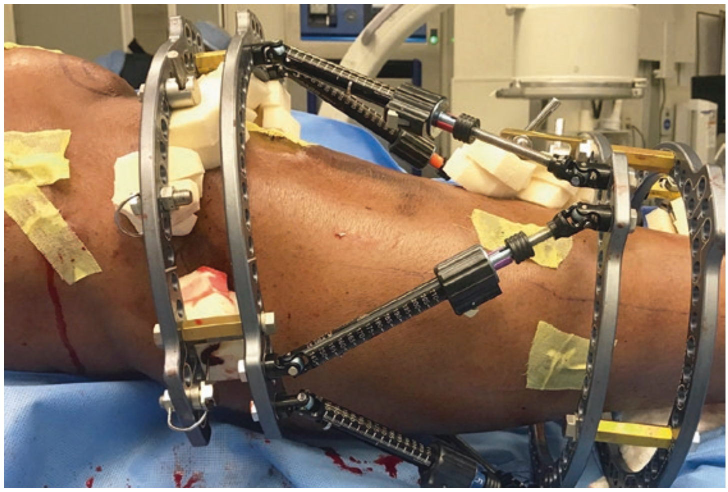

Fig. 8.6 Note the two proximal open rings bolted together to augment rigidity while allowing knee flexion

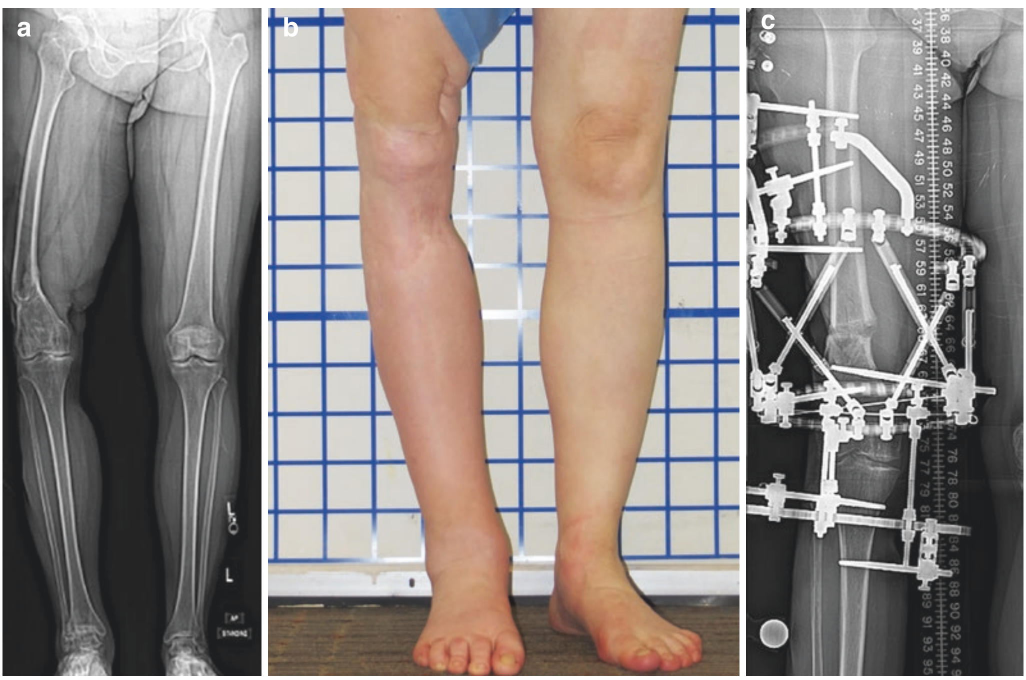

Fig. 8.7 (**a**) This patient suffers from a varus and external rotation malunion of the distal femur. (**b**) The soft tissue over the malunited zone is compromised and cannot withstand any incision precluding a standard distal osteotomy with plate fixation. (**c**) This post-correction radiograph shows a femoral arch used to extend the proximal fixation and a supplementary ring spanned across the knee to gain better control of the distal femur fragment. The distal femur bone quality was poor, indicating the need to cross the knee joint

Fig. 8.8 (**a**) This distal tibial nonunion was treated with removal of hardware, debridement, and routine cultures. (**b**) This intraoperative lateral shows the prodigious use of tensioned wires in this small distal tibial segment [50]. (**c**) This computed tomography scout film shows three connecting rods locking the frame and preventing strut-shuck. The proximal ring is seen with two four-hole Rancho cubes extending fixation off of either side of the ring. (**d**) The final AP shows successful union after several months of fixation. (**e**) This lateral demonstrates final union

during lengthening or while compressing fragments (see Fig. 8.5c). Zenios et al. noted that a procurvatum was more common likely to develop after tibial lengthening with the TSF than with the Ilizarov frame and surmised that the anterior pins underwent cantilever bending and thus contributed to the iatrogenic sagittal plane deformity, which can be addressed with a residual correction using the software [55]. While classic half pins were found to have higher infection rates than wires [67], hydroxyapatite (HA)-coated half pins offer superior bonding to the bone interface [68] as well as better control of deformity [69] and resistance to infection [70]. When a periarticular bone segment is too short for the safe placement of a half pin, then at least five tensioned wires should traverse the fragment [50] ideally including olive wires [71] (see Fig. 8.8b). Olives wires have been very helpful in gaining stability in

Table 8.3 Hexapod biomechanics

Study	Frame	Design	Conclusions
Rodl [52] 2003	TSF vs. Ilizarov	TSF and Ilizarov mounted to bone models. No MTS	TSF can correct 23° of angulation, 36 mm shortening, 71 mm translation, 43° of rotation without a strut change. Ilizarov can correct 90° of angulation, 100 mm of shortening, 25 mm translation, 12.5° of rotation without remounting
Henderson [53] 2008	TSF	MTS mounted to frame directly	Less than 30° ring-strut angle was unstable in compression and bending especially with shorter struts
Lenarz [54] 2008	TSF and Ilizarov	MTS mounted to model	Perpendicular half pins and diverging pins had similar rigidity. Diverging pins can be used with the TSF to save space and avoid strut impingement
Tan [49] 2014	TSF vs. Ilizarov	MTS mounted to model and MTS mounted to frame directly	TSF greater torsional and bending stiffness and similar axial stiffness
Zenios [55] 2014	TSF vs. Ilizarov	Clinical and MTS	TSF with anterior half pins caused the proximal fragment to bend into flexion (through cantilever) more so than the Ilizarov frame
Skomoroshko [56] 2015	Ortho-SUV vs. Ilizarov	MTS mounted to model	Ortho-SUV provided greater rigidity in all planes of loading
Faschingbauer [57] 2015	Precision hexapod	IR tracking system	Hexapod technology corrections have average accuracy of 0.3 mm (−0.5–0.5) and 0.2° (−1.0–0.9)
Birkholtz [58] 2016	TSF	MTS	Fast Fx strut can collapse destabilizing the bone
Henderson [48] 2017	TSF vs. Ilizarov	MTS mounted to model	TSF greater torsional and bending stiffness, but less axial stiffness. Half pins equalized axial stiffness. TSF had a laxity in neutral loading

Ortho-SUV (S.H. Pitkar Orthotools Pvt. Ltd., Pune, India)
Fast Fx strut (Smith & Nephew, Memphis, TN, USA)
TSF Taylor Spatial Frame (Smith & Nephew, Memphis, TN, USA), *MTS* material testing system, *IR* infrared

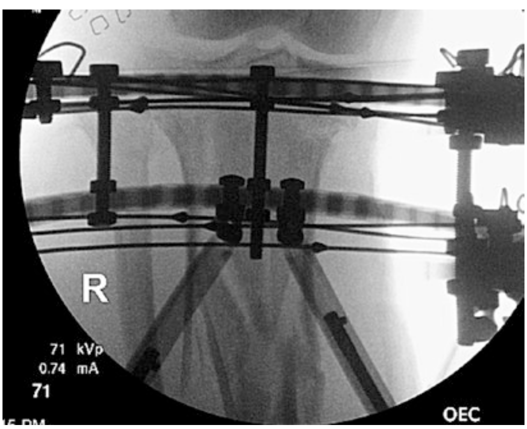

Fig. 8.9 The use of an all-wire, two-ring block construct with multiple opposing olive wires gave excellent control of the proximal fragment in this comminuted, displaced fracture with severe soft tissue compromise. The olives can be used to compress the bone fragments together or to buttress them in place

comminuted fractures and in nonunions [72] (Fig. 8.9). Periarticular fixation must be done meticulously to avoid intracapsular placement of wires and septic arthritis. One should avoid placing a wire within 1.4 cm of the proximal tibial joint surface to avoid intracapsular placement, understanding that the capsular reflection is less distal anteriorly than posteriorly, and in 10–50% of knees, there may be a connection between the knee joint and proximal tibiofibular joint [73]. In the distal tibia, there is a smaller capsular reflection, with the anterolateral capsule inserting 9–12 mm above the joint, while the anteromedial synovial reflection tends to be 3.3–5.5 mm above the ankle joint. Posterior synovial reflections tend to be <2 mm from the joint [74].

Solomin summarized the entirety of the collection of mechanical testing data eloquently, stating that:

> Most clinical studies of the biomechanics of external fixation involved stand tests of external fixation models. The interpretation of the data and their use in practice emphasize the fact that there is no single commonly accepted method for carrying out the stand test. Therefore, to compare the results of studies by different authors objectively is hardly possible, and the number of such studies grows yearly. One should also recognize the fact that no unanimous opinion exists of what the bone fragment fixation rigidity should be at all stages of the bone anatomy restoration. [75]

Fibular stabilization is often required when correcting valgus or procurvatum and when lengthening the tibia. In general, the fibula should be captured at either end to prevent migration which at the knee creates a flexion contracture and at the ankle may disrupt the ankle mortise. Stabilization can be achieved with either a tensioned wire attached to a ring or an internal stabilization screw (see Fig. 8.2c).

The nature and amount of bone contact at the osteotomy or nonunion site will affect bone stability. A fresh osteotomy that is not under compression or distraction forces will provide little stability to the system [48], while a nonunion under strong compression will impart tremendous rigidity to the bone [41]. "Width and contact surface of the bone fragment ends will affect stability" [1], such that Ilizarov recommended shaping the bone ends at a nonunion site into a dome or a lock-and-key configuration to improve contact surface area and stability for improved healing [9] (Fig. 8.10a–d). Another strategy to improve the stability at a nonunion site after deformity correction is to create an adjacent osteotomy [31]. Although this seems counterintuitive, it follows Perren's laws in that the second fracture divides the stresses between the two sites halving the strain that occurs at the nonunion. This immediately improves the relative rigidity of the same fixation at the nonunion site and speeds union [76]. This approach may be responsible for the great success of bone transport in nonunion healing [30, 77, 78] and two-level osteotomies in deformity correction. Although the regenerate bone provides no stability to the limb initially, once the consolidation phase is entered, the osteotomy site mineralizes imparting increasing stiffness over time. The shape of this regenerate bone affects stability with wider callus demonstrating less strain than thinner callus or incomplete columns of new bone [33]. Another peculiarity of the Ilizarov method is the ability to treat a stiff, hypertrophic nonunion in a closed fashion by simply applying the frame and pulling through the fracture site. These inherently rigid deformities require more fixation ("increased mechanical advantage" [2]) and respond to closed angular correction and lengthening with osteogenesis [8, 79, 80].

Need for Neurolysis

Ilizarov wrote that "during deformity correction, the soft tissues on the deformity's concave side lengthen. Many years of clinical practice have led us to the conclusion that correction of angulation of bone fragments with the apparatus must be made gradually, elongating the regenerate bone at a rate of 0.8–1.2 mm per day, while lengthening the soft tissues not more than 3 mm per day" [1]. This prescription was successful in avoiding neurologic complications at the time of the original publication. Modern techniques from the same institution utilizing automated distraction which divides the 1 mm into dozens of micromovements demonstrated even less neurologic compromise [81]. Apart from rate and rhythm considerations, risk factors for nerve entrapment must be considered when correcting deformities. A history of previous distraction osteogenesis (and distraction histogenesis) predisposes the patient to subclinical nerve damage from traction [82]. These traumatized nerves are ripe for further injury with any subsequent lengthening procedure, and any acute deformity correction with nerve stretching ("double crush") is to be avoided [83]. Visible surgical or traumatic scars over the fibular neck or posterior-medial ankle are signs of cicatricial nerve entrapment. The magnitude and location of the deformity will impact the risk of nerve injury with distal tibial combined varus and equinus frequently requiring a tarsal tunnel

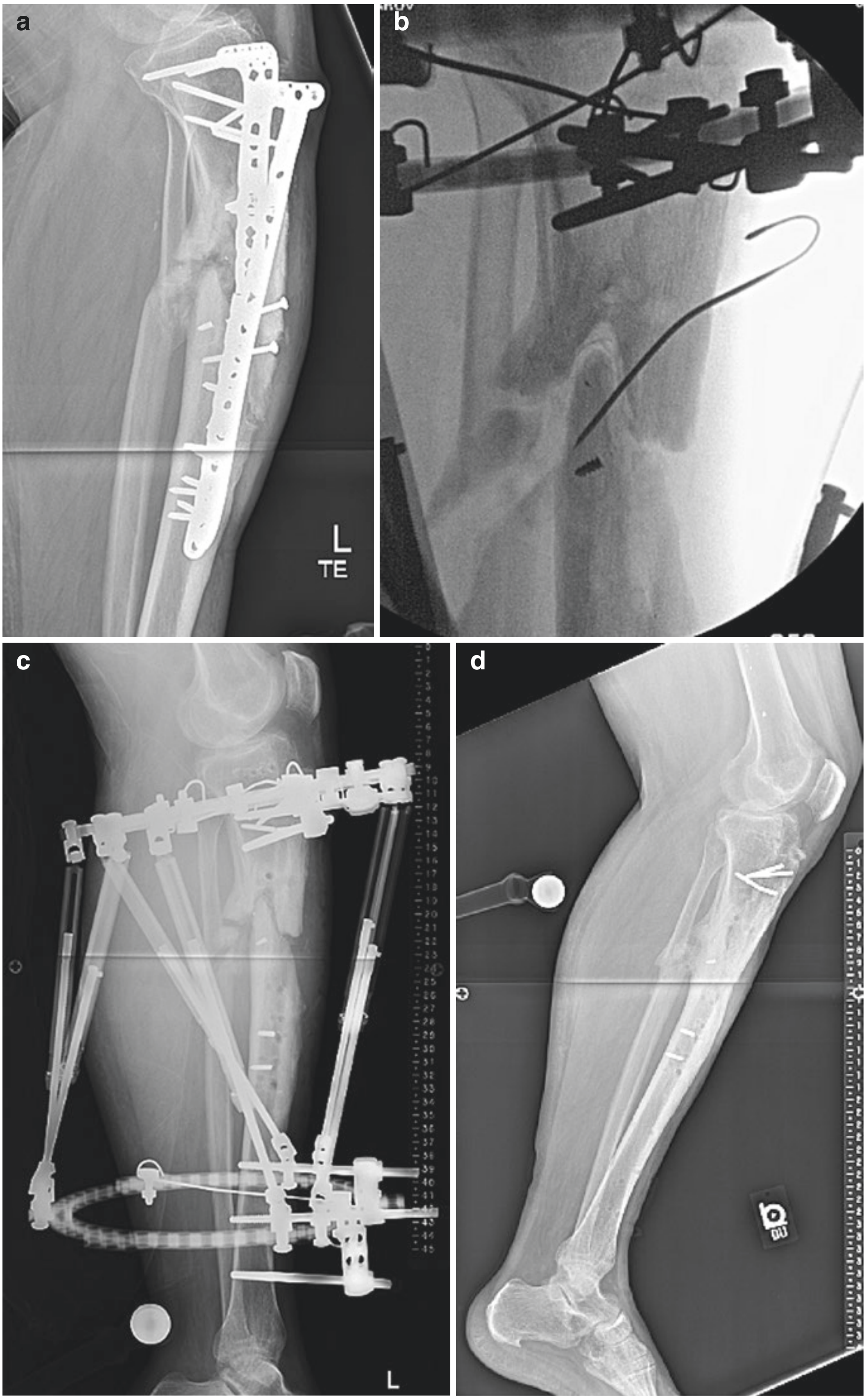

Fig. 8.10 (**a**) This normotrophic nonunion with broken hardware was treated with open hardware removal and nonunion repair. (**b**) The bone ends were fashioned to interlock improving stability during compression. The reduction was temporarily pinned during frame application. (**c**) The rigidity imparted to the limb from having such a stable nonunion site under strong compression allowed for relatively sparse fixation. This two-ring construct spans a long distance of diaphysis without any additional fixation. (**d**) Final images show a successful nonunion repair

release for safe correction. The power of a gradual correction method lies in the ability to react to a changing clinical landscape, so that when a patient begins to complain of new onset tingling in a nerve distribution, the rate of correction can be slowed, paused, or even reversed to allow for nerve recovery. An acute deformity correction can be "undone" or reversed in the early post-op period if an external fixator has been used, immediately relieving the nerve tension and often avoiding the need for nerve decompression.

When to Remove the Frame?

The consensus among limb lengthening surgeons is that the optimal time to remove the external fixator after deformity correction should be made using plain radiographs and identifying bridging callus with a continuous cortical line on at least three of four cortices [83], ensuring that the frame has been in place a perceived adequate length of time, and confirming that the patient no longer has pain with unrestricted weight-bearing [84, 85]. Longer lengthening sites and associated fibular nonunion are associated with a higher risk of the regenerate bending into valgus after frame removal [86]. Dynamization is often invoked to prepare the bone for frame removal [1, 75, 84]. This can be accomplished by removing connecting rods (not struts), removing wires, or adding a spring element to the rod/strut connection to the ring. Thin-sliced CT scan with sagittal and coronal reconstructions provides valuable information of the percentage of bridging the bone at a nonunion site and is also helpful for timing fixator removal in select cases [35].

Complications

Complications that accompany circular external fixation-assisted deformity correction vary in severity and frequency [87]. The Paley categorization of the difficulties encountered during deformity correction into *problems, obstacles, and complications* [88] is the most appropriate way of considering these events. *Problems,* such as low-grade pin infections and transient joint stiffness, are both common [89, 90] and easily remedied with no long-term sequelae. Pins and wires that are closer to the osteotomy in the concavity of the deformity can be expected to become irritated or infected during the correction. Slowing the rate of correction and limited weight-bearing helps reduce pin problems. Wires that lose tension can be re-tensioned in the office, often resolving pain and skin irritation. To prevent the problem of wire slippage at the ring fixation attachment site [91], special grooved/ruffled wire fixation bolts were designed. While the wires no longer slipped, they instead stretched out (plastic deformation) with a similar loss in tension [92, 93]. *Obstacles* require operative intervention to resolve, such as replacing critical broken fixation elements (or rings), neurolysis, gastrocnemius recession, fasciotomies for compartment syndrome recognized early, correction of residual deformity after frame removal, and bone grafting an inadequate regenerate or nonunion site (Fig. 8.11).True *complications* are rare and include the missed compartment syndrome, permanent joint contracture, knee subluxation, and missed septic arthritis cases, all of which are associated with permanent sequelae. Frequent follow-up and anticipation of these issues will assist in their early detection and resolution while still obstacles.

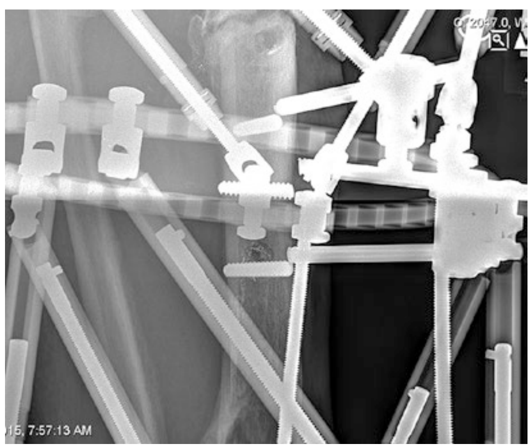

Fig. 8.11 This patient was undergoing bone transport with full weight-bearing when three fractured half pins were discovered at a routine visit. This *obstacle* was resolved with adding additional pins in the operating room

Summary

External fixators have provided the foundation for limb deformity surgery and remain vital to the field. Since external fixation systems are modular, a working knowledge of proper biomechanical principles is important in order to construct a stable frame that will accomplish the goals of the surgeon. The mechanics need to be tempered by patient factors that include bone quality, soft tissue integrity, and anatomic location in order to achieve a successful clinical result. Preoperative planning provides the opportunity to consider the whole patient, to choose an implant, and to anticipate obstacles. The ability to stretch the tissues gradually has broadened the surgeon's ability to treat complex cases. The optimal time for frame removal will likely be more accurate as technology for assessing bone healing advances. Most complications can be anticipated, occur slowly, need to be recognized, and require quick management that typically results in no permanent sequelae.

While this chapter reviews the clinical and laboratory research behind building a biomechanically stable frame and covers a clinical approach to deformity correction, this text should not be mistaken for a comprehensive instruction manual. Its purpose is to introduce the reader to the biomechanical principles that underpin successful limb deformity surgery and provide references for further study. External fixators and the Ilizarov method have created a new field in orthopedic surgery: *limb lengthening and reconstruction.* Dedicated to the correction of limb deformities, limb lengthening surgeons are highly trained and growing in number internationally. Deformity correction surgery using external fixation is a skill best acquired through extra training with an expert. Industry-sponsored courses are very helpful for learning the nuances of particular hexapod systems, but basic surgical technique is best obtained through observerships and limb deformity clinical fellowship programs. In both the office and the operating room, the apprentice will appreciate the dedicated team approach and specialized equipment that is integral to this subspecialty. We hope that this review will inspire further interest in this exciting field.

References

1. Ilizarov GA. The apparatus: components and biomechanical principles of application. In: Ilizarov GA, Green SA, editors. Transosseous osteosynthesis. Heidelberg: Springer; 1992. p. 62–136.
2. Ilizarov GA. Correction of deformities of long tubular bones with simultaneous limb lengthening. In: Ilizarov GA, Green SA, editors. Transosseous osteosynthesis. Heidelberg: Springer; 1992. p. 329–63.
3. Fragomen AT, Meade M, Borst E, Nguyen J, Rozbruch SR. Does the surgical correction of tibial torsion with genu varum produce outcomes similar to those in varus correction alone? J Knee Surg. 2018;31(4):359–69.
4. Ring D, Jupiter JB, Labropoulos PK, Guggenheim JJ, Stanitsky DF, Spencer DM. Treatment of deformity of the lower limb in adults who have osteogenesis imperfecta. J Bone Joint Surg Am. 1996;78(2):220–5.
5. Krappinger D, Zegg M, Smekal V, Huber B. [Correction of posttraumatic lower leg deformities using the Taylor Spatial Frame]. Oper Orthop Traumatol. 2014;26(5):520–31. [Article in German].
6. Alexis F, Herzenberg JE, Nelson SC. Deformity correction in Haiti with the Taylor Spatial Frame. Orthop Clin North Am. 2015;46(1):9–19.
7. Küçükkaya M, Karakoyun O, Armağan R, Kuzgun U. [Correction of complex lower extremity deformities with the use of the Ilizarov-Taylor spatial frame]. Acta Orthop Traumatol Turc. 2009;43(1):1–6. [Article in Turkish].
8. Seybold D, Gessmann J, Ozokyay L, Bäcker H, Muhr G, Graf M. Deformity correction of post-traumatic tibial non-unions using the Taylor Spatial Frame. Z Orthop Unfall. 2009;147(1):26–31.
9. Paley D, Catagni MA, Argnani F, Villa A, Benedetti GB, Cattaneo R. Ilizarov treatment of tibial nonunions with bone loss. Clin Orthop Relat Res. 1989;241:146–65.
10. Tetsworth KD, Paley D. Accuracy of correction of complex lower-extremity deformities by the Ilizarov method. Clin Orthop Relat Res. 1994;301:102–10.
11. Shtarker H, Volpin G, Stolero J, Kaushansky A, Samchukov M. Correction of combined angular and rotational deformities by the Ilizarov method. Clin Orthop Relat Res. 2002;402:184–95.
12. Sen C, Kocaoglu M, Eralp L, Cinar M. Correction of ankle and hindfoot deformities by supramalleolar osteotomy. Foot Ankle Int. 2003;24(1):22–8.
13. Chaudhary M. Taylor spatial frame-software-controlled fixator for deformity correction-the early Indian experience. Indian J Orthop. 2007;41(2):169–74.

14. Marangoz S, Feldman DS, Sala DA, Hyman JE, Vitale MG. Femoral deformity correction in children and young adults using Taylor Spatial Frame. Clin Orthop Relat Res. 2008;466(12):3018–24.
15. Rozbruch SR, Pugsley JS, Fragomen AT, Ilizarov S. Repair of tibial nonunions and bone defects with the Taylor Spatial Frame. J Orthop Trauma. 2008;22(2):88–95.
16. Rozbruch SR, Segal K, Ilizarov S, Fragomen AT, Ilizarov G. Does the Taylor Spatial Frame accurately correct tibial deformities? Clin Orthop Relat Res. 2010;468(5):1352–61.
17. Horn DM, Fragomen AT, Rozbruch SR. Supramalleolar osteotomy using external fixation with six-axis deformity correction of the distal tibia. Foot Ankle Int. 2011;32:986–93.
18. Sökücü S, Karakoyun Ö, Arıkan Y, Küçükkaya M, Kabukcuoğlu Y. Efficacy of the Taylor spatial frame in the treatment of deformities around the knee. Acta Orthop Traumatol Turc. 2013;47(2):86–90.
19. Arvesen JE, Tracy Watson J, Israel H. Effectiveness of treatment for distal tibial nonunions with associated complex deformities using a hexapod external fixator. J Orthop Trauma. 2017;31(2):e43–8.
20. Manner HM, Huebl M, Radler C, Ganger R, Petje G, Grill F. Accuracy of complex lower-limb deformity correction with external fixation: a comparison of the Taylor Spatial Frame with the Ilizarov ring fixator. J Child Orthop. 2007;1(1):55–61.
21. Dammerer D, Kirschbichler K, Donnan L, Kaufmann G, Krismer M, Biedermann R. Clinical value of the Taylor Spatial Frame: a comparison with the Ilizarov and Orthofix fixators. J Child Orthop. 2011;5(5):343–9.
22. Eren I, Eralp L, Kocaoglu M. Comparative clinical study on deformity correction accuracy of different external fixators. Int Orthop. 2013;37(11):2247–52.
23. Lark RK, Lewis JS Jr, Watters TS, Fitch RD. Radiographic outcomes of ring external fixation for malunion and nonunion. J Surg Orthop Adv. 2013;22(4):316–20.
24. Solomin LN, Paley D, Shchepkina EA, Vilensky VA, Skomoroshko PV. A comparative study of the correction of femoral deformity between the Ilizarov apparatus and Ortho-SUV Frame. Int Orthop. 2014;38(4):865–72.
25. Reitenbach E, Rödl R, Gosheger G, Vogt B, Schiedel F. Deformity correction and extremity lengthening in the lower leg: comparison of clinical outcomes with two external surgical procedures. Springerplus. 2016;5(1):2003. eCollection 2016.
26. Paley D, Tetsworth K. Mechanical axis deviation of the lower limbs. Preoperative planning of uniapical angular deformities of the tibia or femur. Clin Orthop Relat Res. 1992;280:48–64.
27. Paley D, Tetsworth K. Mechanical axis deviation of the lower limbs. Preoperative planning of multiapical frontal plane angular and bowing deformities of the femur and tibia. Clin Orthop Relat Res. 1992;280:65–71.
28. Calhoun JH, Li F, Bauford WL, Lehman T, Ledbetter BR, Lowery R. Rigidity of half-pins for the Ilizarov external fixator. Bull Hosp Jt Dis. 1992;52(1):21–6.
29. Donaldson FE, Pankaj P, Simpson AH. Investigation of factors affecting loosening of Ilizarov ring-wire external fixator systems at the bone-wire interface. J Orthop Res. 2012;30(5):726–32.
30. Lowenberg DW, Feibel RJ, Louie KW, Eshima I. Combined muscle flap and Ilizarov reconstruction for bone and soft tissue defects. Clin Orthop Relat Res. 1996;332:37–51.
31. McNally M, Ferguson J, Kugan R, Stubbs D. Ilizarov treatment protocols in the management of infected nonunion of the tibia. J Orthop Trauma. 2017;31(Suppl 5):S47–54.
32. Rozbruch SR, Kleinman D, Fragomen AT, Ilizarov S. Limb lengthening and then insertion of an intramedullary nail: a case-matched comparison. Clin Orthop Relat Res. 2008;466(12):2923–32.
33. Juan JA, Prat J, Vera P, et al. Biomechanical consequences of callus development in Hoffmann, Wagner, Orthofix and Ilizarov external fixators. J Biomech. 1992;25(9):995–1006.
34. Hegde V, Jo JE, Andreopoulou P, Lane JM. Effect of osteoporosis medications on fracture healing. Osteoporos Int. 2016;27(3):861–71.
35. Fourman MS, Borst EW, Bogner E, Rozbruch SR, Fragomen AT. Recombinant human BMP-2 increases the incidence and rate of healing in complex ankle arthrodesis. Clin Orthop Relat Res. 2014;472(2):732–9.
36. Fragomen AT, Borst E, Schachter L, Lyman S, Rozbruch SR. Complex ankle arthrodesis using the Ilizarov method yields high rate of fusion. Clin Orthop Relat Res. 2012;470(10):2864–73.
37. Lamm BM, Paley D, Testani M, Herzenberg JE. Tarsal tunnel decompression in leg lengthening and deformity correction of the foot and ankle. J Foot Ankle Surg. 2007;46(3):201–6.
38. Nogueira MP, Paley D, Bhave A, Herbert A, Nocente C, Herzenberg JE. Nerve lesions associated with limb-lengthening. J Bone Joint Surg Am. 2003;85-A(8):1502–10.
39. Seah KT, Shafi R, Fragomen AT, Rozbruch SR. Distal femoral osteotomy: is internal fixation better than external? Clin Orthop Relat Res. 2011;469(7):2003–11.
40. Paley D, Herzenberg JE, Paremain G, Bhave A. Femoral lengthening over an intramedullary nail. A matched-case comparison with Ilizarov femoral lengthening. J Bone Joint Surg Am. 1997;79(10):1464–80.
41. Catagni MA. Biomechanics of the Ilizarov method. In: Maiocchi B, editor. Treatment of fractures, nonunions, and bone loss of the tibia with the Ilizarov method. Milan: Il Quadratino; 1998. p. 21–4.
42. Bronson DG, Samchukov ML, Birch JG, Browne RH, Ashman RB. Stability of external circular fixation: a multi-variable biomechanical analysis. Clin Biomech (Bristol, Avon). 1998;13(6):441–8.

43. Gasser B, Boman B, Wyder D, Schneider E. Stiffness characteristics of the circular Ilizarov device as opposed to conventional external fixators. J Biomech Eng. 1990;112(1):15–21.
44. Hudson CC, Lewis DD, Cross AR, Horodyski M, Banks SA, Pozzi A. Axial stiffness and ring deformation of complete and incomplete single ring circular external skeletal fixator constructs. Am J Vet Res. 2012;73(12):2021–8.
45. Hudson CC, Lewis DD, Cross AR, Dunbar NJ, Horodyski M, Banks SA, Pozzi A. A biomechanical comparison of three hybrid linear-circular external fixator constructs. Vet Surg. 2012;41(8):954–6.
46. Grivas TB, Magnissalis EA. The use of twin-ring Ilizarov external fixator constructs: application and biomechanical proof-of principle with possible clinical indications. J Orthop Surg Res. 2011;6:41.
47. Antoci V, Voor MJ, Seligson D, Roberts CS. Biomechanics of external fixation of distal tibial extra-articular fractures: is spanning the ankle with a foot plate desirable? J Orthop Trauma. 2004;18(10):665–73.
48. Henderson DJ, Rushbrook JL, Harwood PJ, Stewart TD. What are the biomechanical properties of the Taylor Spatial Frame™? Clin Orthop Relat Res. 2017;475(5):1472–82.
49. Tan B, Shanmugam R, Gunalan R, Chua Y, Hossain G, Saw A. A biomechanical comparison between Taylor's Spatial Frame and Ilizarov external fixator. Malays Orthop J. 2014;8(2):35–9.
50. Antoci V, Roberts CS, Antoci V Jr, Voor MJ. The effect of transfixion wire number and spacing between two levels of fixation on the stiffness of proximal tibial external fixation. J Orthop Trauma. 2005;19(3):180–6.
51. Glatt V, Tepic S, Evans C. Reverse dynamization: a novel approach to bone healing. J Am Acad Orthop Surg. 2016;24(7):e60–1.
52. Rödl R, Leidinger B, Böhm A, Winkelmann W. [Correction of deformities with conventional and hexapod frames--comparison of methods]. Z Orthop Ihre Grenzgeb. 2003;141(1):92–8. [Article in German].
53. Henderson ER, Feldman DS, Lusk C, van Bosse HJ, Sala D, Kummer FJ. Conformational instability of the Taylor spatial frame: a case report and biomechanical study. J Pediatr Orthop. 2008;28(4):471–7.
54. Lenarz C, Bledsoe G, Watson JT. Circular external fixation frames with divergent half pins: a pilot biomechanical study. Clin Orthop Relat Res. 2008;466(12):2933–9.
55. Zenios M, Oyadiji SO. Effect of asymmetrical configuration of pins in the TSF external fixator used for tibial lengthening in a pediatric population. J Pediatr Orthop. 2014;34(6):618–24.
56. Skomoroshko PV, Vilensky VA, Hammouda AI, Fletcher MD, Solomin LN. Mechanical rigidity of the Ortho-SUV frame compared to the Ilizarov frame in the correction of femoral deformity. Strategies Trauma Limb Reconstr. 2015;10(1):5–11.
57. Faschingbauer M, Heuer HJ, Seide K, Wendlandt R, Münch M, Jürgens C, Kirchner R. Accuracy of a hexapod parallel robot kinematics based external fixator. Int J Med Robot. 2015;11(4):424–35.
58. Birkholtz F, Ferreira N. Catastrophic strut collapse with the Taylor Spatial Frame: preventing a disaster. J Med Eng Technol. 2016;40(2):52–4.
59. Podolsky A, Chao EY. Mechanical performance of Ilizarov circular external fixators in comparison with other external fixators. Clin Orthop Relat Res. 1993;293:61–70.
60. Zamani AR, Oyadiji SO. Analytical modelling of Kirschner wires in Ilizarov circular external fixators using a tensile model. Proc Inst Mech Eng H. 2008;222(6):967–76.
61. Henderson DJ, Rushbrook JL, Stewart TD, Harwood PJ. What are the biomechanical effects of half-pin and fine-wire configurations on fracture site movement in circular frames? Clin Orthop Relat Res. 2016;474(4):1041–9.
62. Dearden P, Lowery K, Sherman K, Mahadevan V, Sharma H. Fibular head transfixion wire and its relationship to common peroneal nerve: cadaveric analysis. Strategies Trauma Limb Reconstr. 2015;10(2):73–8.
63. Caja V, Kim W, Larsson S, E YC. Comparison of the mechanical performance of three types of external fixators: linear, circular and hybrid. Clin Biomech (Bristol, Avon). 1995;10(8):401–6.
64. Orbay GL, Frankel VH, Kummer FJ. The effect of wire configuration on the stability of the Ilizarov external fixator. Clin Orthop Relat Res. 1992;279:299–302.
65. Lewis RA, Lewis DD, Anderson CL, Hudson CC, Coggeshall JD, Iorgulescu AD, Banks SA. Mechanics of supplemental drop wire and half-pin fixation elements in single ring circular external fixator constructs. Vet Surg. 2016;45(4):471–9.
66. Arango J, Lewis DD, Hudson CC, Horodyski M. A biomechanical evaluation of three drop wire configurations. Vet Surg. 2013;42(6):669–77.
67. Antoci V, Ono CM, Antoci V Jr, Raney EM. Pin-tract infection during limb lengthening using external fixation. Am J Orthop (Belle Mead NJ). 2008;37(9):E150–4.
68. Moroni A, Cadossi M, Romagnoli M, Faldini C, Giannini S. A biomechanical and histological analysis of standard versus hydroxyapatite-coated pins for external fixation. J Biomed Mater Res B Appl Biomater. 2008;86(2):417–21.
69. Caja VL, Pizǎ G, Navarro A. Hydroxyapatite coating of external fixation pins to decrease axial deformity during tibial lengthening for short stature. J Bone Joint Surg Am. 2003;85-A:1527–31.
70. Pizà G, Caja VL, González-Viejo MA, Navarro A. Hydroxyapatite-coated external-fixation pins. The effect on pin loosening and pin-track infection in leg lengthening for short stature. J Bone Joint Surg Br. 2004;86:892–7.

71. Jabbar Y, Khaleel A. Experimental model for controlling shear using the Ilizarov frame. Clin Biomech (Bristol, Avon). 2015;30(9):995–1001.
72. Cavusoglu AT, Ozsoy MH, Dincel VE, Sakaogullari A, Basarir K, Ugurlu M. The use of a low-profile Ilizarov external fixator in the treatment of complex fractures and non-unions of the distal femur. Acta Orthop Belg. 2009;75(2):209–18.
73. Parameswaran AD, Roberts CS, Seligson D, Voor M. Pin tract infection with contemporary external fixation: how much of a problem? J Orthop Trauma. 2003;17(7):503–7.
74. Pommer A, Muhr G, David A. Hydroxyapatite-coated Schanz pins in external fixators used for distraction osteogenesis : a randomized, controlled trial. J Bone Joint Surg Am. 2002;84-A(7):1162–6.
75. Solomin LN. The basic principles of external skeletal fixation using the Ilizarov device. Milan: Springer; 2008. p. 1–128.
76. Elliott DS, Newman KJ, Forward DP, Hahn DM, Ollivere B, Kojima K, et al. A unified theory of bone healing and nonunion: BHN theory. Bone Joint J. 2016;98-B(7):884–91.
77. Bernstein M, Fragomen AT, Sabharwal S, Barclay J, Rozbruch SR. Does integrated fixation provide benefit in the reconstruction of posttraumatic tibial bone defects? Clin Orthop Relat Res. 2015;473(10):3143–53.
78. Napora JK, Weinberg DS, Eagle BA, Kaufman BR, Sontich JK. Hexapod stacked transport for tibial infected nonunions with bone loss: long-term functional outcomes. J Orthop Trauma. 2018;32(1):e12–8.
79. Rozbruch SR, Fragomen A, Ilizarov S. Correction of tibial deformity with use of the Ilizarov-Taylor Spatial Frame. J Bone Joint Surg Am. 2006;88(Suppl 4):156–74.
80. Ferreira N, Marais LC, Aldous C. Hexapod external fixator closed distraction in the management of stiff hypertrophic tibial nonunions. Bone Joint J. 2015;97-B(10):1417–22.
81. Shchudlo N, Varsegova T, Stupina T, Shchudlo M, Saifutdinov M, Yemanov A. Benefits of Ilizarov automated bone distraction for nerves and articular cartilage in experimental leg lengthening. World J Orthop. 2017;8(9):688–96.
82. Rozbruch SR, Zonshayn S, Muthusamy S, Borst EW, Fragomen AT, Nguyen JT. What risk factors predict usage of gastrosoleus recession during tibial lengthening? Clin Orthop Relat Res. 2014;472(12):3842–51.
83. Babatunde OM, Fragomen AT, Rozbruch SR. Noninvasive quantitative assessment of bone healing after distraction osteogenesis. HSS J. 2010;6(1):71–8.
84. Iobst CA, Mohammed W, Colley R. Determining when it is safe to remove the external fixator: results from a survey of the limb lengthening and reconstruction society. Orthopedics. 2017;40(5):e876–9.
85. Schiedel F, Vogt B, Wacker S, Pöpping J, Bosch K, Rödl R, Rosenbaum D. Walking ability of children with a hexapod external ring fixator (TSF®) and foot plate mounting at the lower leg. Gait Posture. 2012;36(3):500–5.
86. Harbacheuski R, Fragomen AT, Rozbruch SR. Does lengthening and then plating (LAP) shorten duration of external fixation? Clin Orthop Relat Res. 2012;470(6):1771–81.
87. Velazquez RJ, Bell DF, Armstrong PF, Babyn P, Tibirani R. Complications of use of the Ilizarov technique in the correction of limb deformities in children. J Bone Joint Surg Am. 1993;75(8):1148–56.
88. Paley D. Problems, obstacles, and complications of limb lengthening by the Ilizarov technique. Clin Orthop Relat Res. 1990;250:81–104.
89. Fragomen AT, Miller AO, Brause BD, Goldman V, Rozbruch SR. Prophylactic post operative antibiotics may not reduce pin site infections after external fixation. HSS J. 2017;13(2):165–70.
90. Kazmers NH, Fragomen AT, Rozbruch SR. Prevention of pin site infection in external fixation: a review of the literature. Strat Trauma Limb Recon. 2016;2: 75–85.
91. Aronson J, Harp JH Jr. Mechanical considerations in using tensioned wires in a transosseous external fixation system. Clin Orthop Relat Res. 1992;280: 23–9.
92. Gessmann J, Jettkant B, Königshausen M, Schildhauer TA, Seybold D. Improved wire stiffness with modified connection bolts in Ilizarov external frames: a biomechanical study. Acta Bioeng Biomech. 2012;14(4):15–21.
93. La Russa V, Skallerud B, Klaksvik J, Foss OA. Reduction in wire tension caused by dynamic loading. An experimental Ilizarov frame study. J Biomech. 2011;44(8):1454–8.

Part III

Tension Band Wire Principles with Case Examples

9 Biomechanics of Tension Band Constructs for Fracture Fixation

Austin Edward MacDonald, Chetan Gohal, and Herman Johal

Introduction

Tension band constructs for fracture fixation have been described for well over 50 years [1, 2]. The principles behind tension bands were derived from biomechanical studies performed by Frederich Pauwels, who made substantial contributions to our understanding of the relationship between stress, load, and bone [1]. Pauwels used these forces as the basis of fixation constructs for specific fracture types, many of which were formally described and disseminated by the AO (*Arbeitsgemeinschaft für Osteosynthesefragen*) [1–4]. In the case of tension bands, the fundamental concept involves the conversion of distractive tensile forces into compressive forces that are distributed across a fracture site, creating a favourable environment for fracture healing. While there are specific requirements for successful application of tension band principles, these constructs can be used to treat a variety of long bone and peri-articular fractures using a range of implants. This chapter will review the essential biomechanical tension band principles and review examples of their effective application.

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Tension bands are most commonly used to treat olecranon or patella fractures; however tension band principles can also successfully be applied to treat long bone fractures (i.e. femur fractures) and other peri-articular or avulsion-type fractures (such as greater trochanter fractures, greater tuberosity fractures, malleolar fractures, or styloid fractures). All of these situations require an understanding that many bones throughout the body are eccentrically loaded, resulting in tension and compression surfaces. The tensile surface must be amenable to the application of fixation, while the compression surface must be intact and able to resist load [1–4]. Not only will this create a setting that promotes fracture healing, but it will also impart stability that will facilitate early mobility and functional recovery [5–9].

Key Concepts for Tension Band Constructs

Determine the Tension and Compression Surfaces of the Fracture

Pauwels originally described that under axial loads, many curved, tubular bones have tension and compression surfaces opposite to one another. When an eccentric load is applied, the curved or convex side of the bone is subject to

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tensile forces while the concave side experiences compression (Fig. 9.1) [1, 3]. This may be best conceptualized when thinking of a long bone such as the femur, where the bone is eccentrically loaded secondary to the body weight being applied through the femoral head and down the eccentric mechanical axis, instead of neutrally through the central anatomic axis (Fig. 9.2). This results in tension or torque being applied to the lateral side of the femur, with the medial side being compressed.

While this concept applies to curved, tubular bones with axial loads, it also applies to bones that move around an eccentric centre of rotation. These are often under torque secondary to the pull of muscle tendons or ligaments, with the tension surface being further away from the centre of rotation and the compressive surface being closer. Patella fractures provide the clearest example of this, where the patellar and quadriceps tendons apply tension to the non-articular surface of the patella, while the articular surface experiences compression, as it moves along the centre of rotation within the knee (Fig. 9.3a). Similar forces around other joints act on tendon or ligament avulsion fractures including fractures of the olecranon (Fig. 9.3b), shoulder tuberosities, greater trochanter, or ankle malleoli [3–9].

The resultant tensile and compressive forces from either an eccentric axial load or centre of rotation commonly results in a transverse fracture pattern, where the convex (outer) cortex is subject to tension and the concave (inner) cortex will be subject to compression. Without fixation, the fracture would be distracted, causing an unstable healing environment that promotes gapping and non-union. The mechanical function of a tension band construct is to work against these forces and convert this distractive torque into stable compression across the fracture site.

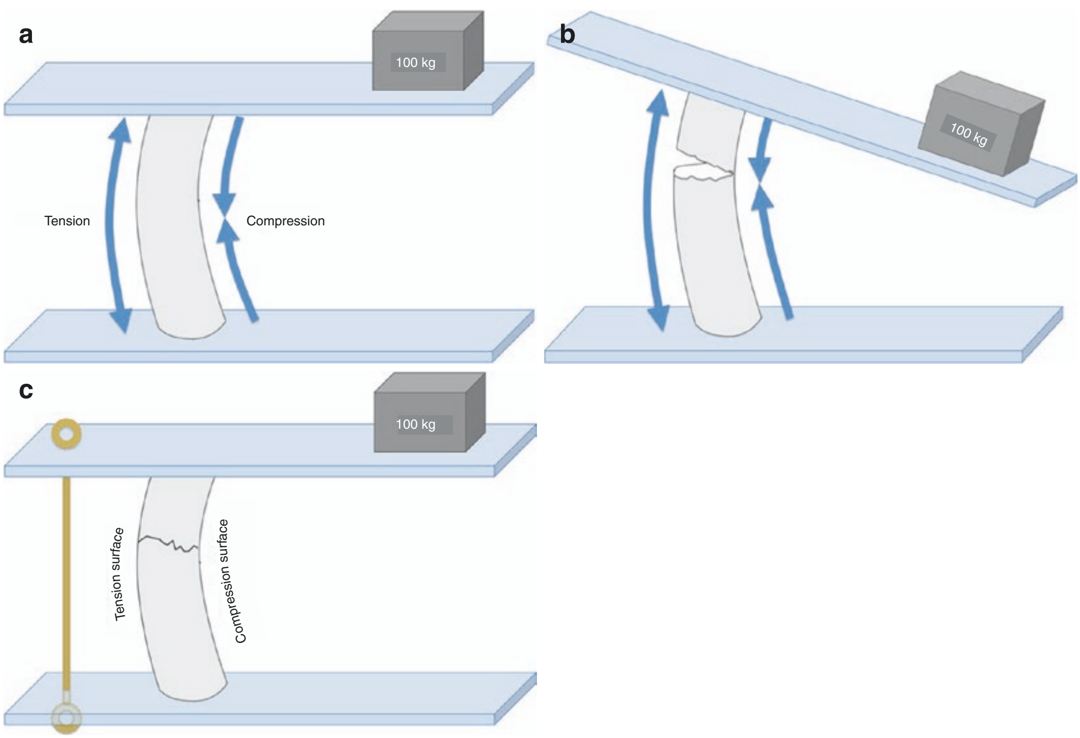

Fig. 9.1 The application of an eccentric axial load on a tubular bone will result in a creation of a tension surface along the curved (outer) cortex and a compression surface along the concave (inner) cortex (**a**). These forces result in distraction across a fracture site (**b**), which can be neutralized with a tension band placed appropriately along the outer cortex, converting tensile forces into compression ones (**c**)

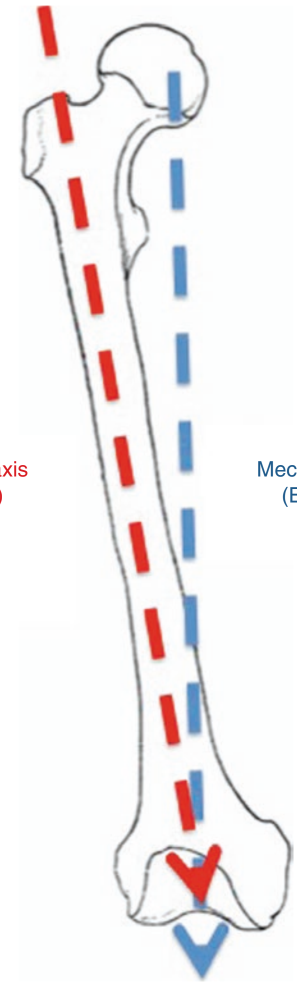

Fig. 9.2 Depiction of the eccentric axial load placed along the mechanical axis (*blue dotted line*) of the femur, which sits outside of the central anatomic axis (*red dotted line*)

Ensure the Fracture Can Withstand Stable Compression, with an Intact Opposite Cortex

Prior to applying tension band techniques, the surgeon must ensure that the fracture pattern is amenable to this type of fixation. Transverse fractures due to bone failing under tensile stress are best suited for tension band constructs. If placed appropriately, tension band fixation should be able to neutralize the forces distracting the fracture along the tensile surface, ideally converting them into a steady, compressive force at the fracture (Fig. 9.4a). If the fracture cannot withstand compression, a tension band construct will not work.

The compression side of the bone has a particularly important role as a buttress. For tension band fixation to be successful, the compression surface must be intact or reconstructed with stable bony apposition. If unable to do so, as in comminuted fractures, there will be less resistance to compression forces across the fracture site (Fig. 9.4b) [3, 10]. This will either result in immediate loss of reduction and fixation or increased motion at the fracture leading to delayed or non-union and failure of hardware [3, 10]. Therefore, tension band constructs are not typically indicated in the context of comminuted fractures, which are better served with more rigid, bridging fixation.

Apply Fixation to Withstand Tension

Placing fixation along the convex, or tension, side of the bone will allow the implant to resist the stress of tensile forces and convert them into compression (see Fig. 9.4a) [1, 3, 10]. Conversely, placing fixation on the concave, or compression, side of the fracture permits unresisted tensile forces to act along the convex cortex, leading to ongoing distraction and gapping at the fracture site (Fig. 9.4c).

Tension band constructs can be applied along the tensile surface as either static or dynamic fixation, dependent on the forces being resisted and applied by the implants [1–3]. Both provide compression forces at the fracture site, with static constructs maintaining a relatively constant force during loading and dynamic constructs providing increasing force and the tensile load escalates [3, 4]. At the time of application, static tension band constructs are maximally loaded in compression, and there is little fluctuation with eccentric forces [3–7]. Conversely, dynamic tension bands can impart further compression as it resists increasing tensile loads during joint movement or weight-bearing [8, 9, 11].

Numerous implants may be applied on the tension surface, including cerclage wires, cables, sutures, and plates, as well as appropriately

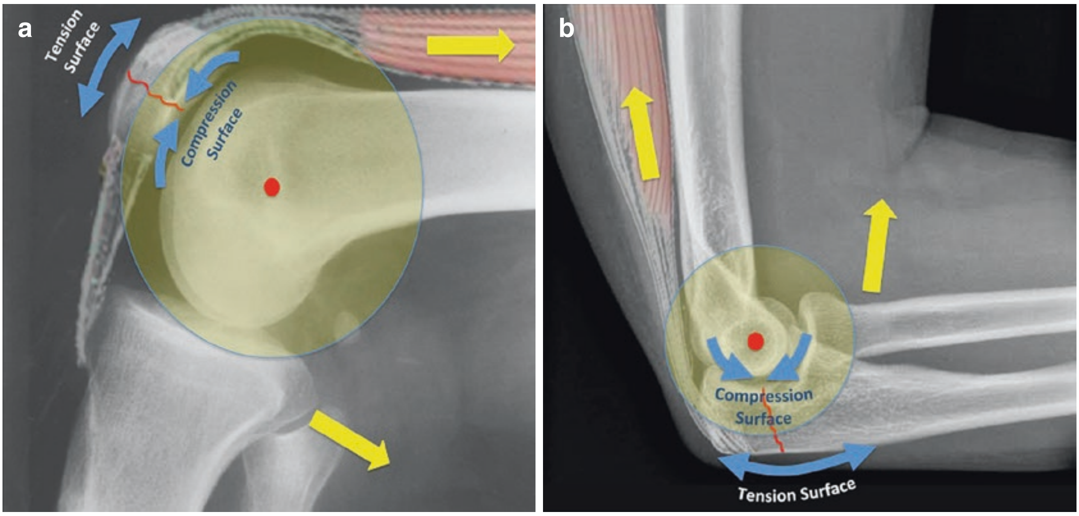

Fig. 9.3 Radiographs with superimposition of the main forces (*yellow arrows*) and muscles acting to create tensile and compressive forces as the patella (**a**) and the olecranon (**b**) move along their respective centres of rotation (*red dot*). The patella experiences tension from pull of both the patellar and quadriceps tendons as the knee flexes secondary to contraction of the hamstrings. The olecranon experiences tension from the pull of the triceps as the arm flexes secondary to contraction of the biceps

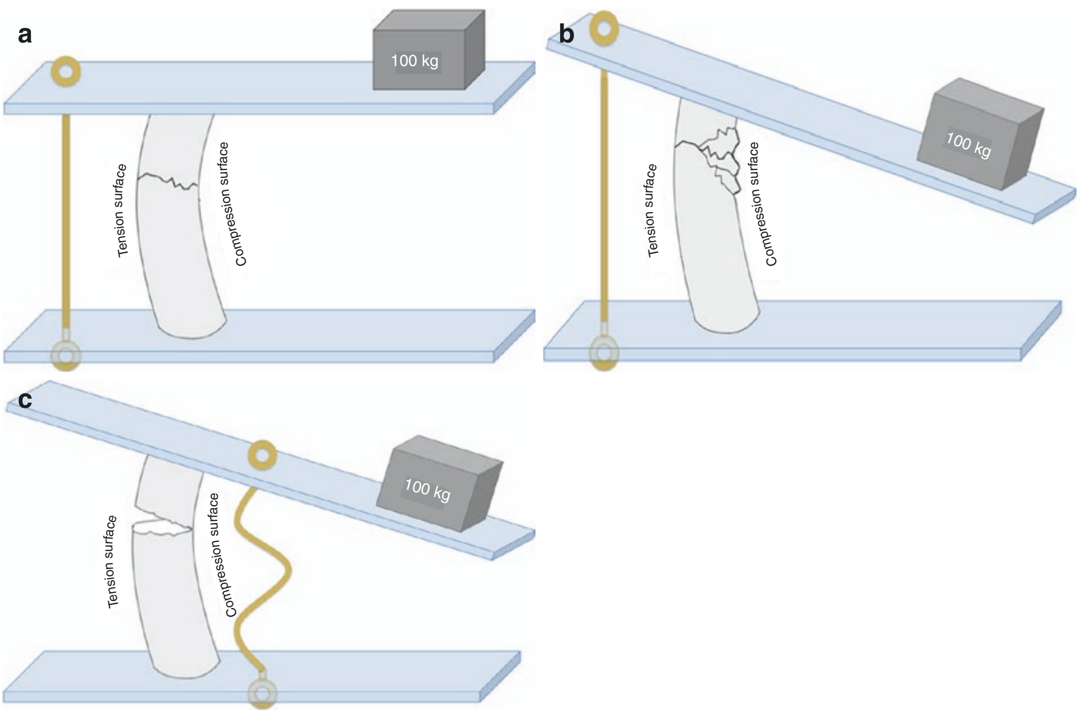

Fig. 9.4 When appropriately placed on the tension side of the bone, a tension band is able to neutralize axial loads if there is a stable, intact medial cortex able to withstand compression (**a**); however if there is substantial comminution unable to resist compression (**b**), or if the hardware is placed on the compression side allowing unresisted distraction, the construct will fail (**c**)

applied intramedullary nails and external fixators [3–11]. The most commonly applied constructs involve cerclage wires (placed through tendons, ligaments, or transosseous drill holes) that are looped around Kirschner wires (K-wires) or passed through cannulated screws placed within the bone, perpendicular to the plane of the fracture. The cerclage wires are then twisted and loaded in compression on the convex surface to impart stability to the fragments as they squeeze together, in line with the plane of the K-wires or cannulated screws. Tension band plating is also commonly applied in curved diaphyseal bones, such as femoral shaft fractures, that undergo varus bending with axial loads. Laterally placed plate fixation provides torque conversion to neutralize the tensile forces acting at the convex cortex. These plates work best when applied in compression, to further load the fracture site and biomechanically optimize the healing environment.

Case 1

A 29-year-old female sustained a mechanical fall on ice. She fell backwards and landed with a direct blow to her elbow. She sustained an isolated, closed transverse fracture to her olecranon with a fracture fragment involving approximately half of the articular surface (Fig. 9.5a, b). A tension band construct was used as for surgical stabilization of her fracture to facilitate early range of motion (Fig. 9.5c, d). A standard posterior approach to the olecranon was used, and anatomic reduction was obtained and maintained using a standard reduction clamp. Two parallel 1.8 mm Kirschner wires were placed in a posterior to anterior direction, perpendicular to the plane of the fracture. A cerclage wire was then passed through a transosseous hole distal to the fracture and looped in a figure-of-eight fashion over the tension surface of the proximal ulna. The wire was then passed deep to the insertional fibres of the triceps tendon on the proximal fracture fragment and anchored by the bent ends of the K-wires proximally. The construct was tensioned through twisting of the wire ends, with the knot bent and impacted against the bone to avoid prominence.

Postoperatively, the patient was allowed to perform a range of motion as tolerated immediately and permitted perform resistance exercises at 6 weeks. At her 3-month follow-up appointment, she had complete radiographic union of her fracture, with range of motion at the elbow from 15 to 150 degrees of flexion with full supination and pronation (Fig. 9.5e, f). She was discharged by her 6-month visit, with full clinical recovery.

Why This Works

In transverse fractures at the olecranon, the proximal fragment distracts secondary to the pull of the triceps tendon during muscle contraction. At the fracture site, the tensile forces are most prominent at the outer, curved cortex of the proximal ulna, while the compressive forces are concentrated at the articular surface. The goal of the tension band wire construct is to convert the dynamic distraction force during elbow motion into a compression forces across the articular surface during motion of the elbow. Compression at the fracture site was obtained intraoperatively using a clamp, with the implant providing further compression through loading as the wire ends were twisted. This provided immediate stable fixation at the fracture site, which only increased with further tension from the pull of the triceps.

Case 2

A 54-year-old male tripped forward while going up a flight of stairs, landing with a direct impact of his flexed knee against the riser. He sustained an isolated, closed primarily transverse fracture to his patella with only minimal comminution at the site of impact on the non-articular surface (Fig. 9.6a, b) A tension band construct was used as for surgical stabilization of her fracture to facilitate early range of motion (Fig. 9.6c, d). A standard anterior approach to the knee was used, and reduction was obtained and maintained using a standard reduction clamp, focusing primarily on anatomic restoration of the simple transverse articular fracture line. Two parallel 4.0 mm

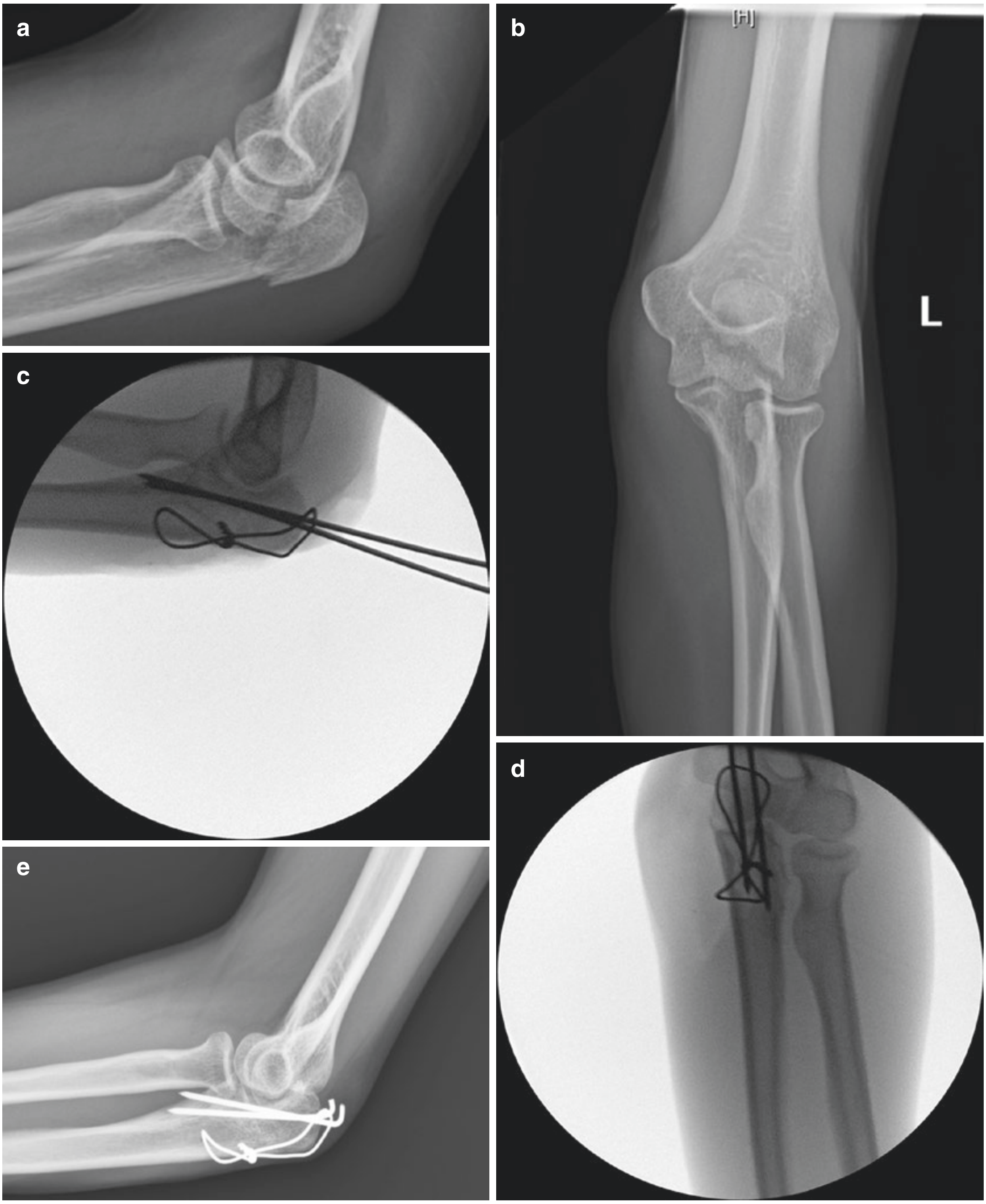

Fig. 9.5 Injury (**a**, **b**) intraoperative (**c**, **d**) and 2-month (**e**, **f**) radiographs of an elbow with a displaced, transverse olecranon fracture treated with a tension band wire construct using K-wires and a figure-of-eight cerclage wire

partially threaded cannulated screws were then placed longitudinally within the patella, perpendicular to the plane of the fracture. Careful attention was paid to ensure that the screws were adequately buried and not prominent at either side of the patella. The centrally cannulated portion of each screw can facilitate passage of a 1.4 mm cerclage wire, which was fed through

Fig. 9.5 (continued)

each screw and tensioned over the curved, outer tensile surface of the patella.

Postoperatively, the patient was allowed to immediately weight-bear with the leg in a knee immobilizer and to begin gentle range-of-motion exercises by 4 weeks. At his 2-month follow-up appointment, radiographic union was achieved (Fig. 9.6e, f), with range of motion being from 0 to 110 degrees of flexion. The patient reached functional recovery of range of motion and strength by the 4-month follow-up visit.

Why This Works

In transverse patella fractures, the proximal and distal fragments are distracted from each other secondary to the pull of the quadriceps and patellar tendons, respectively. The tensile forces are most prominent at the outer, curved cortex of the patella, while the compressive forces are concentrated at the articular surface. The goal of the tension band wire construct is to convert the dynamic distraction force into a compression force across the articular surface during motion of the knee. Compression at the fracture site was initially obtained intraoperatively using a clamp, as well as the partially threaded cannulated screws which provided interfragmentary compression.

By ensuring that the screw heads were countersunk, and that the ends of the screw were not prominent, the cerclage wire was able to impart further compression through loading of the construct as the wire ends were twisted. This held the fragments together, as they moved along the plane of the screws to compress across the fracture site.

Case 3

A 76-year-old male fell down several rungs of a ladder sustaining an injury to his right thigh. He had a previous stemmed right total knee

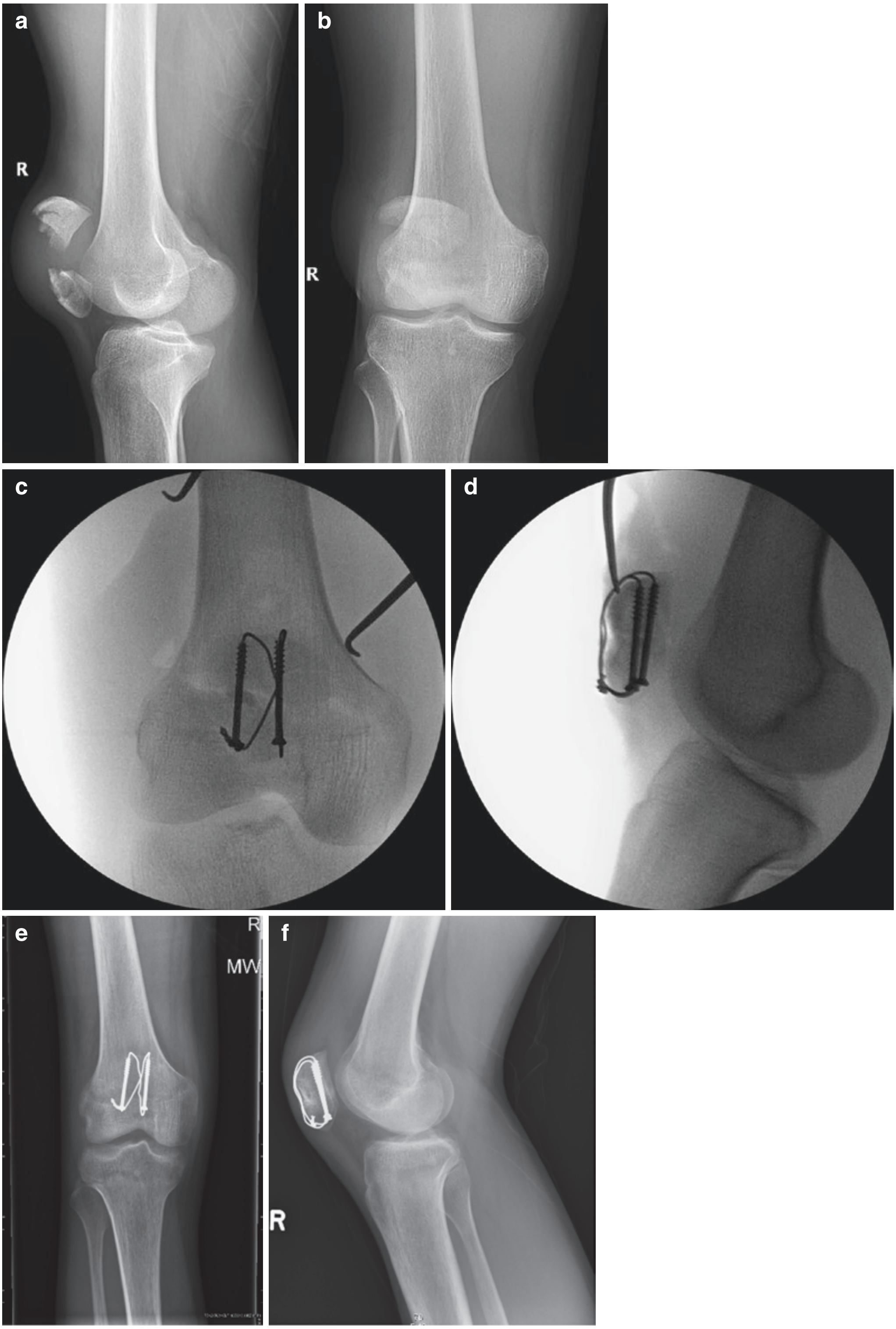

Fig. 9.6 Injury radiographs of a displaced, transverse patella fracture with minimal comminution along the non-articular surface (**a**, **b**). Intraoperative fluoroscopic images showing placement of a cannulated screw and cerclage wire tension band construct (**c**, **d**). Four-month radiographs showing fracture union (**e**, **f**)

arthroplasty (TKA) placed 10 years earlier and did not have any antecedent thigh pain. Radiographs confirmed an oblique femoral shaft fracture ending just proximal to the stem of the femoral component (Fig. 9.7a–c). Open reduction and internal fixation was chosen as the TKA appeared to be stable. Intramedullary nailing was not an option secondary to the stemmed femoral component, and plate fixation was selected. A lateral approach to the left femur was used, and the fracture was anatomically reduced using traction and clamps. Cortical keys, including a transverse medial component, helped to maintain the reduction, and an interfragmentary screw was also placed to compress along the main oblique fracture line. A long distal femoral locking plate was positioned and secured. The plate was applied initially using non-locking fixation, followed by locking fixation distally to ensure stable fixation around the femoral TKA component. Postoperatively, the patient was allowed increased range of motion and weight-bearing over the first 8 weeks. At his 2-month follow-up appointment, radiographic union was progressing (see Fig. 9.6e, f), with range of motion being from 0 to 100 degrees of flexion and the patient able to mobilize with a walker.

Why This Works

Among the several functions served by this plate, it works as a tension band as it neutralizes the tensile forces acting at the lateral femoral cortex described earlier in this chapter. Despite being primarily oblique, the simple fracture line had a transverse medial component that allowed restoration of a stable buttress along the medial cortex, able to withstand compressive forces. To add further stability, an independent interfragmentary screw was placed along the long, oblique component of the fracture more proximally. While with this facilitated anatomic reduction, this construct would still be far from adequate to resist the tensile forces that would work to distract the fracture under physiologic loads with weight-bearing. Therefore, a rigid lateral plate was placed to neutralize the tensile forces and convert them into

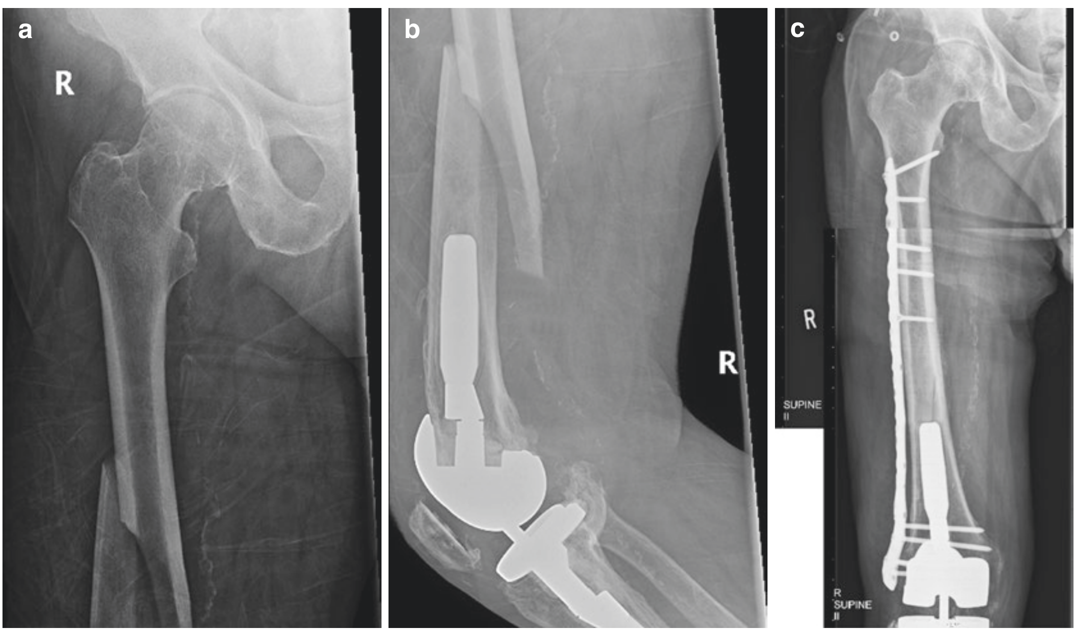

Fig. 9.7 Injury radiographs of a displaced, periprosthetic femoral shaft fracture with a simple fracture line proximal to a stable, femoral stemmed knee arthroplasty component (**a**). Postoperative radiograph showing a long lateral femoral plate acting as a tension band, with stable medial cortical apposition (**b**, **c**)

compressive forces across the anatomically reduced fractures thereby encouraging primary boney healing.

Conclusion

Many fractures throughout the body undergo eccentric axial loads or torque, making tension band constructs a viable option for achieving boney union. This is dependent on both the characteristics of the fracture and effective application of the biomechanical principles involved. The goal of tension band constructs is to convert a distractive, tensile force into a steady compressive force across the fracture site. This requires an understanding of how various bones are eccentrically loaded and where tensile and compression forces act to distract fracture fragments. These forces can be resisted and neutralized via fixation placed along the tension surface using a variety of implants, which work to either statically or dynamically convert tensile forces into compressive ones to promote fracture healing. Stable boney apposition at the opposing cortex provides an intact buttress able to resist the interfragmentary compression, imparting strength and stability to the overall fracture construct. This facilitates earlier weight-bearing of eccentrically loaded long bone fractures, as well as earlier range of motion of peri-articular fractures rotating around an eccentric centre of rotation, subject to tension from tendons or ligaments. In this manner, an appropriately used tension band construct can both accelerate fracture healing and promote early improvement of clinical outcomes, valued by both patients and surgeons alike.

References

1. Pauwels F. Biomechanics of the locomotor apparatus: contributions on the functional anatomy of the locomotor apparatus. Berlin/Heidelberg: Springer; 1980.
2. Weber BG, Vasey H. [Osteosynthesis in olecranon fractures]. Z Unfallmed Berufskr. 1963;56:90–96. [Article in German].
3. Ruedi TP, Buckley RE. Moran CG. AO principles of fracture management. New York: Thieme; 2007.
4. Müller ME, Allgöwer M, Schneider R, Willenegger H. Manual of internal fixation: techniques recommended by the AO-Group. 2nd ed. Berlin/Heidelberg: Springer; 1979.
5. Kosanovic M, Brilej D, Komadina R, Buhanec B, Pilih IA, Vlaovic M. Operative treatment of avulsion fractures of the anterior superior iliac spine according to the tension band principle. Arch Orthop Trauma Surg. 2002;122:421–3.
6. John R, Dhillon MS, Khurana A, Aggarwal S, Kumar P. Tension band wiring is as effective as a compression screw in a neglected, medial malleolus nonunion: a case-based discussion & literature review. J Orthop Case Rep. 2017;7(4):72–5.
7. Sun Y, Li L, Dai J, Wang T. Treatment of complex proximal humeral fracture: plate and tension band fixation versus conservative therapy. Int J Clin Exp Med. 2015;8(5):7143–51.
8. Duckworth AD, Clement ND, White TO, Court-Brown CM, McQueen MM. Plate versus tension-band wire fixation for olecranon fractures: a prospective randomized trial. J Bone Joint Surg Am. 2017;99(15):1261–73.
9. Lotke PA, Ecker ML. Transverse fractures of the patella. Clin Orthop Relat Res. 1981;158:180–4.
10. Reynders Frederix P, Reynders Frederix C, Illes T. Pressure plate fixation according to Brunner & Weber for the treatment of recalcitrant non-union of the femur. A theoretical-biomechanical analysis. Novel Tech Arthritis Bone Res. 2017;1(2):NTAB.MS.ID.555559.
11. Muller FJ, Galler M, Fuchtmeier B. Clinical and radiological results of patients treated with orthogonal double plating for periprosthetic femoral fractures. Int Orthop. 2014;38:2469–72.

10 Olecranon Fractures

Dominique M. Rouleau

Introduction

Tension band fixation is a classic fracture repair method which is most frequently used in olecranon fractures [1]. According to the theory developed by Pauwels in the 1980s [2], a curved tubular structure submitted to a compressive force presents with a tension side and a compressive side [3]; therefore, when a tension band is applied on the tension side, it will concomitantly increase compression on the opposite side. Following this accepted 40-year-old principle, it is mandatory for the bone on the opposite side of the tension band to present a frank fracture line, with no bone loss and no comminution.

However, the olecranon cannot exactly be described as a curved tubular bone submitted to a compression force, and studies have revealed the weak compression created by olecranon tension bands [4]. Brink et al. actually report greater compression in active extension [5]. Indeed, during active motion, multiple force vectors are applied to the first 8 cm of the olecranon in a complex axis. Anatomic causes include the proximal ulna dorsal angulation (PUDA), which varies from 0 to 14 degrees [6], and a 14-degree varus angulation, and they need to be considered [7] to obtain anatomic reduction. This portion of the ulna includes two joints, but since it is not a diaphysis, it does require precise anatomic reduction. We could also say that it is part of another "joint," the forearm pro-supination axis – now considered to be an important virtual articulation. These three joints can be negatively impacted during the treatment of proximal ulna fractures. One such example is PUDA malalignment, shown in the lab to cause radial head subluxation [8]. A case-controlled study on olecranon fracture fixation revealed that patients with a non-anatomic reduction of their PUDA of 5 degrees or more presented with worse outcomes in terms of range of motion [9]. Non-anatomic reduction of the sigmoid notch joint surface is also associated to worse outcomes. Finally, the proximal ulna is the insertion site of the elbow's collateral ligaments, which need to be preserved or repaired in complex olecranon fracture cases.

That being said, all studies comparing plate and tension band (TB) fixation in olecranon fractures have failed to show any significant clinical difference (Table 10.1) [10–15]. Some authors found more complications with the tension band method, mainly associated with a higher hardware removal rate [11, 12]. A systematic review published in 2016 reported equivalent results for both methods, but more reoperations in the TB groups [16]. One of the elements favoring tension band fixation is its much lower cost compared to locking plate fixation. Indeed, with the tension band method, total treatment costs are

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Table 10.1 Comparative table of tension band and plate for olecranon fractures in clinical studies [10–15]

Author	Year	N	Tension band ROM arc	Tension band Score	Tension band Hardware removal	Plate ROM arc	Plate Score	Plate Hardware removal
Amini [10]	2015	10/10	132°	MEPS 97 QDASH 10	4	132°	MEPS 95 QDASH 11	1
Snoddy [11]	2014	43/134	na	na	20/43	na	na	25/134
Tarallo [12]	2014	33/45	na	MEPS 88 QDASH 12	10/33	na	MEPS 89 QDASH 11	4/45
DelSole [13]	2016	23/25	132°	na	2/23	126°	na	0/25
Liñán-Padilla [14]	2017	26/23	140°	VAS 2	8/26	142°	VAS 2	10/23
Schliemann [15]	2014	13/13	na	MEPS 97 QDASH 13	12/13	na	MEPS 97 QDASH 14	7/13

ROM range of motion, *MEPS* Mayo Elbow Performance, *QDASH* QuickDASH (http://www.dash.iwh.on.ca/), *VAS* visual analogue scale

Table 10.2 Schneider criteria

Oversized Kirschner wires in terms of length
Loose figure-of-eight configuration (i.e., the wire cerclage not "flush" to the bone)
Incorrect reduction (i.e., congruent joint articular surface)
Perforation of the joint surface
Nonparallel Kirschner wires (with reference to the other Kirschner wire) on anterior–posterior view
Kirschner wires extending radially outward
Proximal ends of the Kirschner wires not bent 180 degrees back into the cortical bone of the olecranon
Two intramedullary Kirschner wires
Single wire knot
Prominent wire knot(s) (i.e., twisted ends not sufficiently bent back into direct contact with the bone)

From Claessen et al. [18], with permission

lower by more than 50%, including reoperation, and the implant itself is six times cheaper [10–15, 17]. It therefore should still be used in the case of a simple fracture, in accordance with biomechanical principles. However, two recent papers have shown that it is not as easy as it might appear at first glance to perform a "perfect" tension band of the olecranon, with a vast majority of cases not following guidelines [18, 19]. Criteria used by these authors can be found in Table 10.2 [10–15]. Contrary to clinical reports, biomechanical studies show better performance with locking plate systems [4, 20], although it is difficult to accurately compare the varying biomechanical setups from the different studies. When compression is measured, the locking plate creates 343 N of compression vs 77 N for the tension band [4], and on cyclic loading, there is less fragment displacement with plate fixation (0.25 mm vs 1.12 mm) [20].

This chapter will present, first, a descriptive classification of olecranon fractures, to help surgeons understand the injury; second, it will clarify the indication for tension band in olecranon fractures with illustrative cases; and third, it will review surgical tips to increase the solidity of olecranon tension band, based on biomechanical and clinical studies.

Descriptive Olecranon Fracture Classification

Several olecranon fracture classifications can be used, and in my daily practice, I prefer to make a list of all fractured fragments when planning for surgery. The principal fragments are presented in Fig. 10.1 and listed below:

- Tricipital fragment (Fig. 10.2)
- Intermediate fragment [21] (Fig. 10.3)
- Coronoid fragment
  - Tip of the coronoid
  - Anteromedial facet
- Posterior fragment (Fig. 10.4)
- Supinator crest (lateral collateral ligament)
- Sublime tubercle (medial collateral ligament)

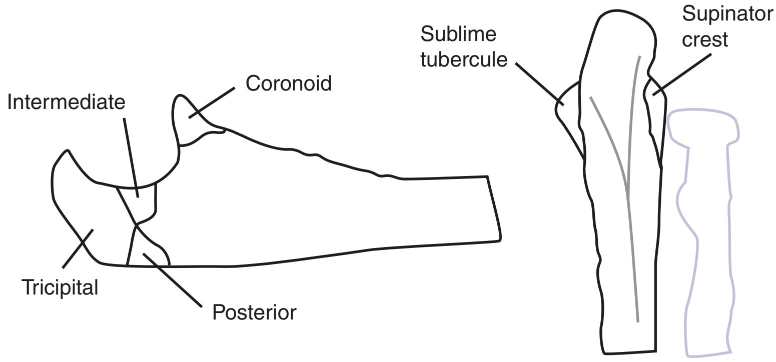

Fig. 10.1 Fragment specific classification of olecranon fractures

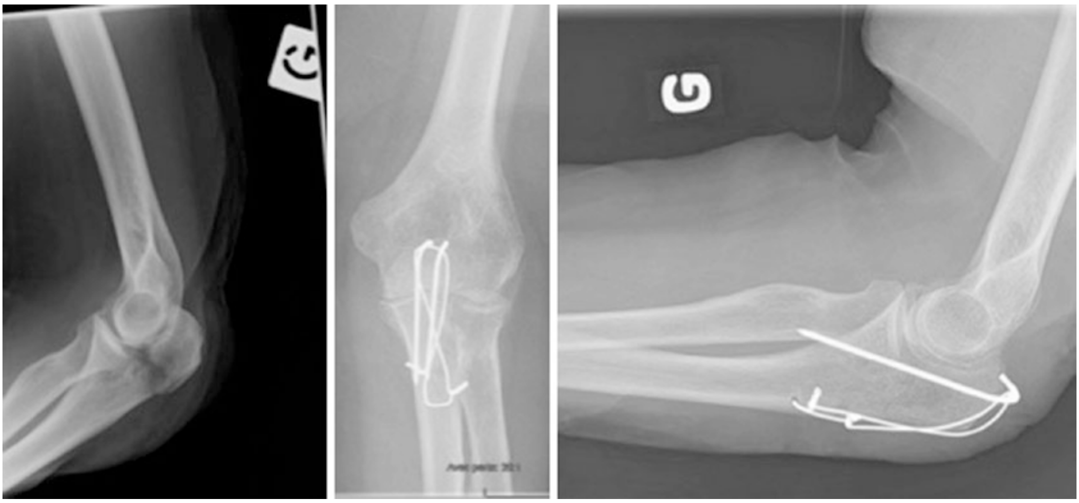

Fig. 10.2 Image of the tricipital fragments present in an olecranon fracture, including the triceps insertion

Fig. 10.3 Image of the intermediate fragment. This fragment is usually covered by cartilage and should be reduced to match the trochlea curvature. It can be supported by a threaded K-wire prior to closure of the acromion

Each fragment needs to be fixed in the case of an olecranon fracture to recreate a stable and mobile elbow. Whenever more than the tricipital fragment is involved, a CT scan should be performed, with 3D reconstruction if possible. Failure to identify and treat all fragments can lead to disastrous results (Fig. 10.5).

Indications for Olecranon Fracture Tension Band

Tension band surgical fixation, using K-wires and metallic wires, is a good surgical fixation option in cases of a simple fracture, without elbow instability or dislocation. For example, a fracture with a tricipital fragment and an intermediate fragment could be fixed by the tension band method in non-osteopenic bone. The intermedi-

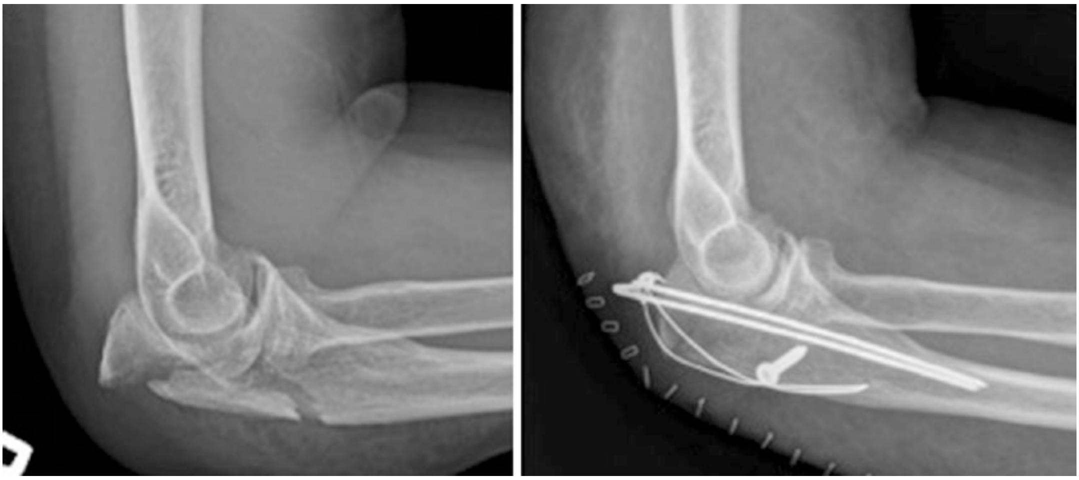

Fig. 10.4 Image of the posterior fragment. This fragment is not articular. Its anatomic reduction is very important to recreate patient specific proximal ulna dorsal angulation. It is frequently associated with anterior subluxation of the distal ulna and radial head

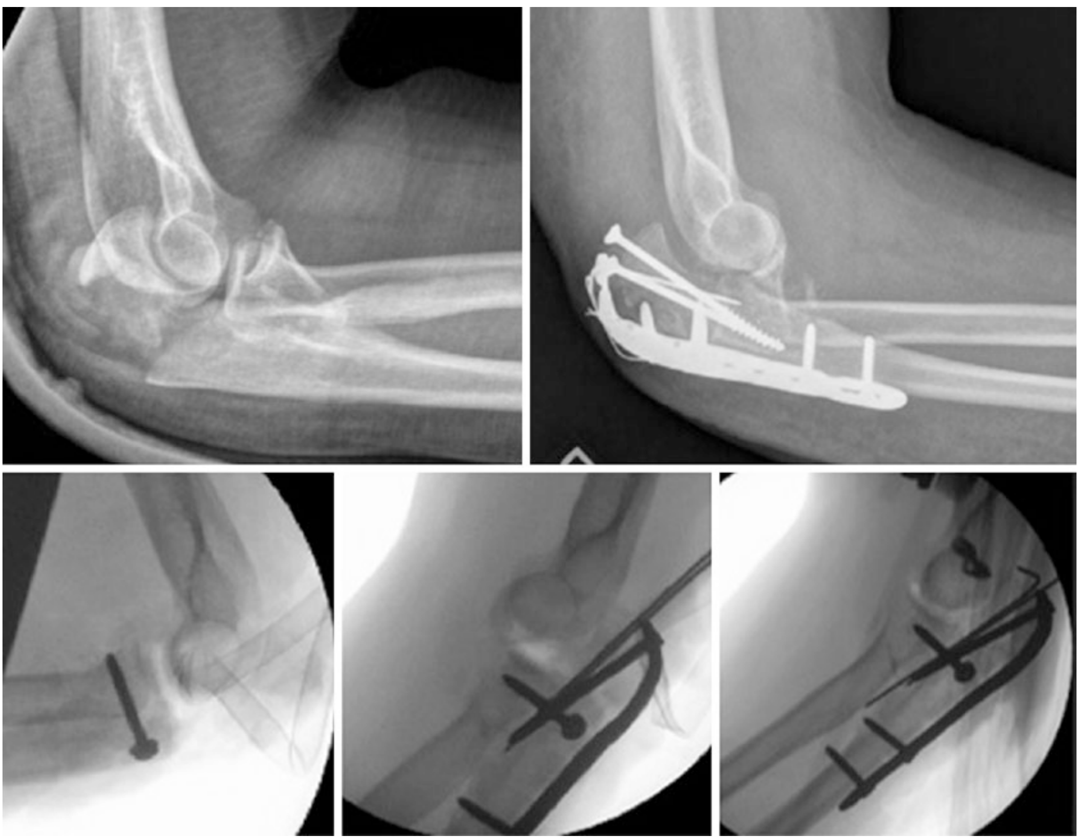

Fig. 10.5 Complex proximal ulna fracture with initial treatment neglecting the coronoid fragment leading to elbow subluxation. In the presence of concomitant olecranon fracture and coronoid fracture, the coronoid should be fixed first, in flexion. The olecranon is then fixed in extension. Tension band is not recommended in that situation. Fracture fixation revision is showed with coronoid fixation first

Fig. 10.6 Example of a complex case of proximal radius and ulna fracture dislocation. Plate fixation is preferred in the presence of a radial head fracture and/or elbow dislocation

ate fragment is reduced and fixed, first with a threaded K-wire, followed by a classic tension band. When the coronoid is involved or there is a combined fragmentation of the intermediate fragment and the posterior fragment, plate fixation will create a more stable construct. When there are associated injuries to the radial head and/or ligaments, plate fixation is also more stable (Fig. 10.6).

Surgical Tips and Tricks Based on Biomechanical Studies

A preoperative x-ray of the normal side is useful in complex fractures to achieve patient specific PUDA. It is easier to repair olecranon fractures in lateral decubitus, with the fractured elbow on top. An elbow support is used under the arm and positioned as close as possible to the shoulder, allowing for fluoroscopy visualization. Skin incision is done as a lazy C shape starting on the ridge of the ulna, 7 cm distal to the tip of the olecranon. The incision is directed proximally, 1 cm lateral to the tip of the olecranon. The incision ends in the center of the posterior elbow, 2 cm proximal to the olecranon to expose and protect the triceps. Full-thickness skin flaps are created with a number 15 blade, just enough to see the fracture fragments. Soft tissues are reflected from the fracture edge and the medial and lateral side of the ulnar ridge. Ulnar nerve and collateral ligament insertions should be preserved but not necessarily identified. A sterile Mayo table is also used to support the forearm, with the elbow in extension for the reduction of the posterior and tricipital fragments. Coronoid fragments are reduced with the elbow in flexion [22]. The coronoid is the keystone for elbow stability and usually requires plate fixation [23]. A sterile tourniquet is used and inflated as little as possible, to decrease postoperative edema and pain. Anatomic reduction of all six potential fragments, of both joint surfaces, the PUDA, and the varus angle, need to be as similar as possible to the contralateral side prior to definitive fixation. Small threaded K-wires could be used for interfragmentary fixation of intermediate fragments before "closing" the tricipital fragment [24].

K-Wires

After anatomic reduction with a reduction clamp and temporary K-wires, two 1.6-mm K-wires are drilled in the ulna; this is easier with the elbow in 30 degrees of flexion. The entry points need to be 5 mm anterior to the tip of the olecranon and should be parallel. Views differ on whether or not to enter the anterior cortex:

- *The anterior cortex fixation philosophy* is supported by biomechanical studies revealing a stronger pull-out strength [25]. When chosen, it needs to be angled at 25 degrees on the lateral view. This angle represents a compromise, decreasing the probability of intra-articular penetration [26] and neurovascular injuries [27]. Structures at risk are the ulnar artery and anterior interosseous nerve [27]. To prevent synostosis, K-wires should be directly aligned with the ulna and not directed toward the radius [28]. If a surgeon chooses an anterior cortex fixation, it is important to retract the wire by 5 mm, prior to bending and cutting it, making it possible to bend the pin 5 mm away from the cortex. A 5 mm of bent stump is left and the remaining wire is cut. The K-wire is then twisted 180 degrees to grab the metallic wire. The K-wire is finally impacted in the bone for 5 mm. Doing this achieves maximal fixation and minimizes the risk to the anterior structures.

- *The intramedullary fixation philosophy* is supported by clinical studies, which report neurovascular complications with the anterior cortex fixation as well as a higher risk of synostosis [29–31]. This method is weaker and K-wires are more likely to back out, especially if they are not impacted into the olecranon after having been bent and cut (Fig. 10.7). To create a stronger intramedullary fixation, surgeons can choose longer K-wires or a 6.5-mm cancellous screw [32, 33]. We do not recommend using large screws because of the risk of triceps fragment fragmentation [33].

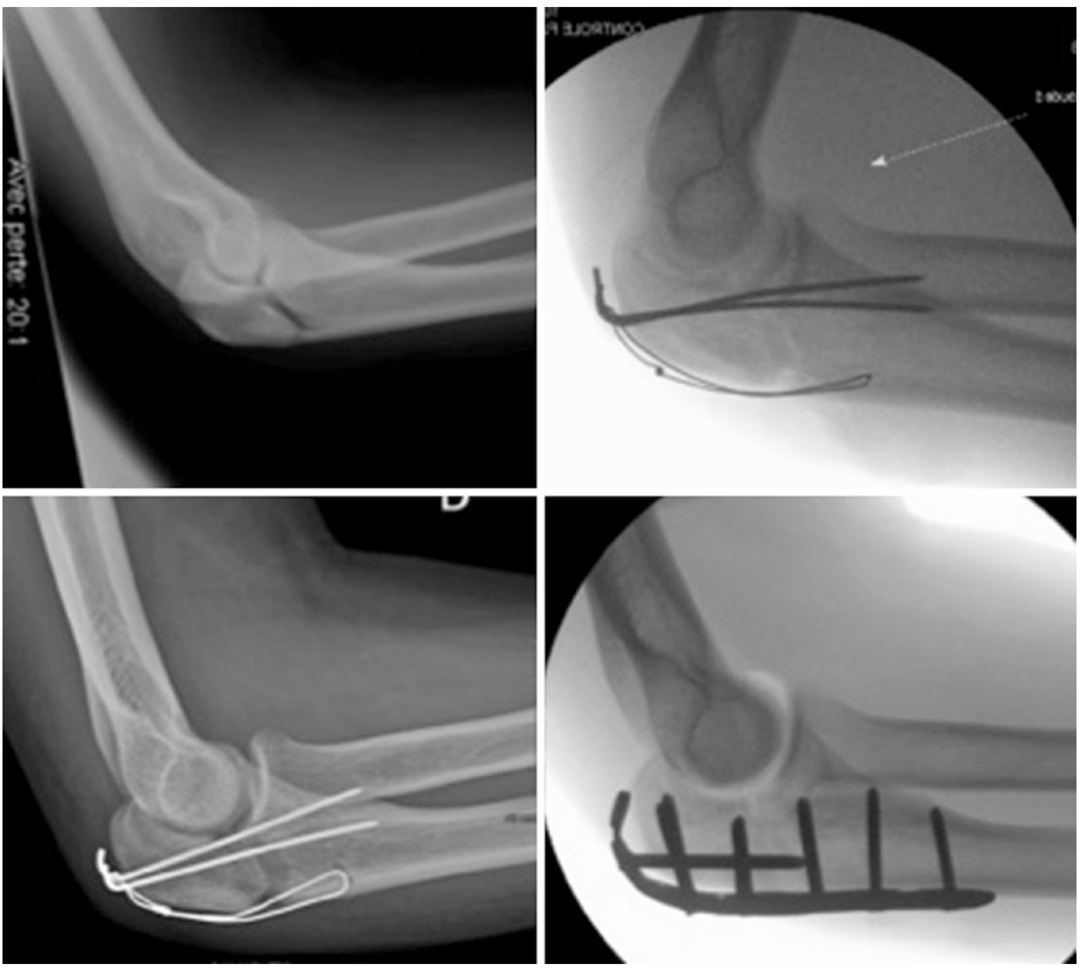

Fig. 10.7 Example of a failed tension band fixation secondary to insufficient anterior cortex fixation, failure to impact the k-wire in the olecranon and choice of too small implants

Wire

A 1-mm wire is inserted through a 2-mm cortical tunnel, distal to the fracture. The two limbs are then crossed. A second wire then goes under the K-wires in the triceps tendon. Precautions should be taken to protect the ulnar nerve on the medial side. K-wires are then retracted 5 mm, one at a time, to prevent fixation failure, before being bent, cut, and re-impacted in the ulnar cortex. Limbs of both wires can now be connected. Compression with the wires is created by a symmetrical rotation and gentle traction of the wires on each side of the fracture. Limbs are then cut and knots are buried in the soft tissue [34]. Alternatively, in an olecranon osteotomy model, Lalliss et al. showed similar strength using a heavy suture (FiberWire, Arthrex, Naples, FL, USA) [35].

Conclusion

Olecranon fractures are very common, requiring surgical interventions in the vast majority of cases. Tension band fixation is a cost-effective procedure favored in simple fractures; however, proper fixation methods need to be followed to prevent failure. These include lateral decubitus positioning, safe surgical approach, anatomic reduction of each fragment with independent small threaded wires, and tension band fixation of the tricipital fragment. Complex fractures and fracture dislocations are preferably treated with designated periarticular locking plates.

References

1. Wood T, Thomas K, Farrokhyar F, Ristevski B, Bhandari M, Petrisor B. A survey of current practices and preferences for internal fixation of displaced olecranon fractures. Can J Surg. 2015;58(4):250–6.
2. Pauwels F. Biomechanics of the locomotor apparatus. 1st ed. Berlin/Heidelberg/New York: Springer; 1980.
3. Hak DJ, Stewart RL. 3.2.3 Tension band principle. In: Rüedi TP, Buckley RE, Moran CG, editors. AO principles of fracture management. 2nd ed. Stuttgart: Thieme; 2007. https://www2.aofoundation.org/wps/portal/!ut/p/a0/04_Sj9CPykssy0xPLMnMz0vMAfGjzOKN_A0M3D2DDbz9_UMMDRyDXQ3dw9wMDAzMjfULsh0VAbWjLW0!/?bone=Radius&segment=Distal&showPage=F&contentUrl=srg/popup/further_reading/PFxM2/323_Tension_bnd.jsp. Accessed 25 Oct 2018.
4. Wilson J, Bajwa A, Kamath V, Rangan A. Biomechanical comparison of interfragmentary compression in transverse fractures of the olecranon. J Bone Joint Surg Br. 2011;93(2):245–50.
5. Brink PR, Windolf M, de Boer P, Brianza S, Braunstein V, Schwieger K. Tension band wiring of the olecranon: is it really a dynamic principle of osteosynthesis? Injury. 2013;44(4):518–22.
6. Rouleau DM, Faber KJ, Athwal GS. The proximal ulna dorsal angulation: a radiographic study. J Shoulder Elb Surg. 2010;19(1):26–30.
7. Puchwein P, Schildhauer TA, Schöffmann S, Heidari N, Windisch G, Pichler W. Three-dimensional morphometry of the proximal ulna: a comparison to currently used anatomically preshaped ulna plates. J Shoulder Elb Surg. 2012;21(8):1018–23.
8. Sandman E, Canet F, Petit Y, Laflamme GY, Athwal GS, Rouleau DM. Radial head subluxation after malalignment of the proximal ulna: a biomechanical study. J Orthop Trauma. 2014;28(8):464–9.
9. Chapleau J, Balg F, Harvey EJ, Ménard J, Vauclair F, Laflamme GY, et al. Impact of olecranon fracture malunion: study on the importance of PUDA (proximal ulna dorsal angulation). Injury. 2016;47(11):2520–4.
10. Amini MH, Azar FM, Wilson BR, Smith RA, Mauck BM, Throckmorton TW. Comparison of outcomes and costs of tension-band and locking-plate osteosynthesis in transverse olecranon fractures: a matched-cohort study. Am J Orthop (Belle Mead NJ). 2015;44(7):E211–5.
11. Snoddy MC, Lang MF, An TJ, Mitchell PM, Grantham WJ, Hooe BS, et al. Olecranon fractures: factors influencing re-operation. Int Orthop. 2014;38(8):1711–6.
12. Tarallo L, Mugnai R, Adani R, Capra F, Zambianchi F, Catani F. Simple and comminuted displaced olecranon fractures: a clinical comparison between tension band wiring and plate fixation techniques. Arch Orthop Trauma Surg. 2014;134(8):1107–14.
13. DelSole EM, Pean CA, Tejwani NC, Egol KA. Outcome after olecranon fracture repair: does construct type matter? Eur J Orthop Surg Traumatol. 2016;26(2):153–9.
14. Liñán-Padilla A, Cáceres-Sánchez L. Type II olecranon fractures in patients over 65. Tension band or preformed plate? Analysis and results. Rev Esp Cir Ortop Traumatol. 2017;61(5):339–42.
15. Schliemann B, Raschke MJ, Groene P, Weimann A, Wähnert D, Lenschow S, Kösters C. Comparison of tension band wiring and precontoured locking compression plate fixation in Mayo type IIA olecranon fractures. Acta Orthop Belg. 2014;80(1):106–11.
16. Ren YM, Qiao HY, Wei ZJ, Lin W, Fan BY, Liu J, et al. Efficacy and safety of tension band wiring versus plate fixation in olecranon fractures: a systematic

review and meta-analysis. J Orthop Surg Res. 2016;11(1):137.
17. Francis T, Washington T, Srivastava K, Moutzouros V, Makhni EC, Hakeos W. Societal costs in displaced transverse olecranon fractures: using decision analysis tools to find the most cost-effective strategy between tension band wiring and locked plating. J Shoulder Elb Surg. 2017;26(11):1995–2003.
18. Claessen FMAP, van den Bekerom MPJ, van Dijk CN, Goslings JC, Kerkhoffs GMMJ, Doornberg JN. Shoulder elbow platform. Tension band wiring for simple olecranon fractures: evaluation of surgical technique. J Orthop Traumatol. 2017;18(3):275–81. Erratum: J Orthop Traumatol. 2018;19(1):16.
19. Schneider MM, Nowak TE, Bastian L, Katthagen JC, Isenberg J, Rommens PM, et al. Tension band wiring in olecranon fractures: the myth of technical simplicity and osteosynthetical perfection. Int Orthop. 2014;38(4):847–55.
20. Gruszka D, Arand C, Nowak T, Dietz SO, Wagner D, Rommens P. Olecranon tension plating or olecranon tension band wiring? A comparative biomechanical study. Int Orthop. 2015;39(5):955–60.
21. von Rüden C, Woltmann A, Hierholzer C, Trentz O, Bühren V. The pivotal role of the intermediate fragment in initial operative treatment of olecranon fractures. J Orthop Surg Res. 2011;6:9.
22. Rouleau DM, Sandman E, van Riet R, Galatz LM. Management of fractures of the proximal ulna. J Am Acad Orthop Surg. 2013;21(3):149–60. Review.
23. Niéto H, Billaud A, Rochet S, Lavoinne N, Loubignac F, Pietu G, et al. Proximal ulnar fractures in adults: a review of 163 cases. Injury. 2015;46(Suppl 1):S18–23.
24. Suresh SS. Management of comminuted olecranon fractures with precut K-wires and tension band wiring. Tech Hand Up Extrem Surg. 2009;13(2):82–4. Review.
25. Saeed ZM, Trickett RW, Yewlett AD, Matthews TJ. Factors influencing K-wire migration in tension-band wiring of olecranon fractures. J Shoulder Elb Surg. 2014;23(8):1181–6.
26. Özsoy MH, Kızılay O, Günenç C, Özsoy A, Demiryürek D, Hayran M, et al. Modified tension band wiring technique for olecranon fractures: where and how should the K-wires be inserted to avoid articular penetration? Acta Orthop Traumatol Turc. 2015;49(2):190–6.
27. Catalano LW 3rd, Crivello K, Lafer MP, Chia B, Barron OA, Glickel SZ. Potential dangers of tension band wiring of olecranon fractures: an anatomic study. J Hand Surg Am. 2011;36(10):1659–62.
28. Willinger L, Lucke M, Crönlein M, Sandmann GH, Biberthaler P, Siebenlist S. Malpositioned olecranon fracture tension-band wiring results in proximal radioulnar synostosis. Eur J Med Res. 2015;20:87.
29. De Carli P, Gallucci GL, Donndorff AG, Boretto JG, Alfie VA. Proximal radio-ulnar synostosis and nonunion after olecranon fracture tension-band wiring: a case report. J Shoulder Elb Surg. 2009;18(3):e40–4.
30. Lee HJ, Jung JW, Cho DW, Jeon IH. Morphometric analysis of the proximal ulna using three-dimensional computed tomography and computer-aided design: varus, dorsal, and torsion angulation. Surg Radiol Anat. 2014;36(8):763–8.
31. Rompen JC, Vos GA, Verheyen CC. Acute ischemia of the hand seven months after tension-band wiring of the olecranon. J Shoulder Elb Surg. 2010;19(3):e9–11.
32. Huang TW, Wu CC, Fan KF, Tseng IC, Lee PC, Chou YC. Tension band wiring for olecranon fractures: relative stability of Kirschner wires in various configurations. J Trauma. 2010;68(1):173–6.
33. Mauffrey CP, Krikler S. Surgical techniques: how I do it? Open reduction and tension band wiring of olecranon fractures. Injury. 2009;40(4):461–5.
34. Powell AJ, Farhan-Alanie OM, Bryceland JK, Nunn T. The treatment of olecranon fractures in adults. Musculoskelet Surg. 2017;101(1):1–9.
35. Lalliss SJ, Branstetter JG. The use of three types of suture and stainless steel wire tension banding for the fixation of simulated olecranon fractures: a comparison study in cadaver elbows. J Bone Joint Surg Br. 2010;92(2):315–9.

Patella Fractures

11

Jessica L. Page, Ross K. Leighton, and Chad P. Coles

Introduction

Patella fractures account for approximately 1% of fractures of the appendicular skeleton, with 77% of patella fractures occurring as a result of simple falls [1]. The soft tissue envelope overlying the patella is extremely thin which increases the risk of open fractures in this area, even at relatively low-energy mechanisms. Rates of open fracture have been reported from 2.7% to 29% depending on the mechanism of injury [1, 2]. Stellate fracture patterns result from a direct blow to the anterior knee. Transverse fracture patterns can result from forced knee flexion during contraction of the quadriceps muscles, resulting in a tension failure of the patella.

Anatomy

The patella is the largest sesamoid bone in the body, and its articular cartilage is the thickest in the body, measuring up to 7 mm in depth [3]. Its articular surface consists of two main facets: lateral and medial. The lateral facet is broader with a deeper concavity than the medial facet [4]. These two facets are separated by a distinct interfacet ridge. A more medial facet, known as the odd facet, is present in some individuals [5]. The articular cartilage of the patella first engages the trochlear groove at around 15–30 degrees of knee flexion. The congruity of the trochlea and the patella contributes to patellar stability [5]. A bipartite patella, believed to arise from a failure of fusion of a secondary ossific nucleus, is found in 2–3% of individuals and may be bilateral [6]. The non-fused portion is most commonly found at the superolateral aspect of the main patella, and care is required to ensure it is not mistaken for a fracture.

As a sesamoid bone, the patella is encased in a sheath of soft tissue. The tendinous insertions of the rectus femoris, vastus lateralis, and vastus medialis converge to form the quadriceps tendon which inserts on the superior aspect of the patella (Fig. 11.1) [7]. The vastus intermedius lies beneath the rectus femoris and inserts on the superior patella just deep to the main quadriceps tendon, separated by a thin bursa [5]. The vastus lateralis muscle is the largest of the quadriceps muscles, but all four muscles provide roughly equal contributions to leg extension due to the trajectory of their vector of pull, as demonstrated in biomechanical studies [8]. The aponeurosis of the quadriceps muscles encases the patella and blends into the anterior knee joint capsule,

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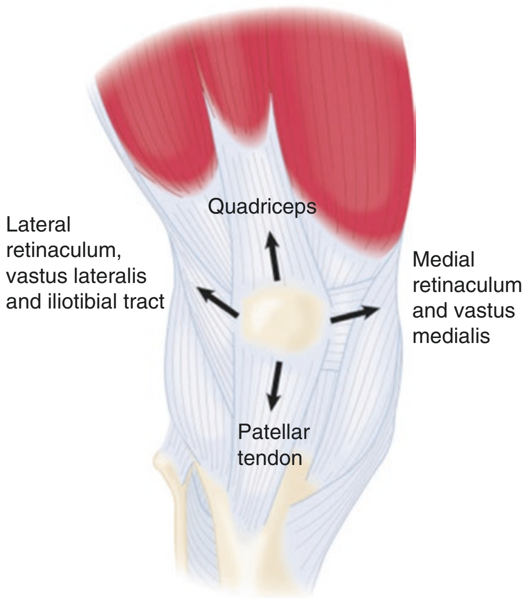

Fig. 11.1 The aponeurosis of the quadriceps tendon encapsulates the patella and forms the retinaculum. The quadriceps tendon inserts on the superior aspect of the patella, and the patellar tendon arises from the inferior aspect of the patella and then inserts on the tibial tubercle distally. (From Dejour and Saggin [7] ©2018, with permission from Elsevier)

creating the retinaculum. This retinaculum extends distally over the superficial layer of the patellar tendon until it blends into the periosteum of the tibia [5]. The patellar tendon originates from the inferior aspect of the patella and inserts on the tibial tubercle. It is surrounded by a paratenon layer.

The vascular supply of the patella is created by a ring anastomosis contributed to by five genicular arteries plus the recurrent anterior tibial artery [9]. Although the blood supply to the patella is robust, the trauma of the injury can disrupt a significant portion. Therefore, it is incumbent on the surgeon to understand the vascular networks and minimize further disruption through surgical dissection. When raising medial and lateral flaps, care should be taken to remain superficial to the periosteum to avoid devitalizing the patella of its dorsal network of vessels [9]. The correct layer can be difficult to identify in traumatized tissue; therefore, we recommend beginning proximally at the level of healthier quadriceps tendon to identify the correct depth and then working distally. The medial and lateral blood supply travels through the patellar retinaculum [9]. The surgeon should work through any traumatic rents already created in the tissue and avoid creating an additional parapatellar arthrotomy so as to preserve what retinacular blood supply remains.

Biomechanics of the Extensor Mechanism and Its Relation to Injury

The extensor mechanism is responsible for transmitting the powerful contractile force of the quadriceps muscles to the tibia to cause extension at the knee joint. The patella serves as a lever to increase the efficiency of the pull of the quadriceps muscles and redirect the vector of their contractile force (Fig. 11.2) [3, 7]. This results in an increase in moment arm [3]. The femoral condyles serve as the fulcrum of the lever [10]. During a closed kinetic chain bend of the knee (i.e., a squat), the patella experiences tension forces resultant from the quadriceps tendon and patellar tendon pulling in opposing directions, as well as a compressive force at the cartilage as the patella rides against the femoral condyles [11]. Accordingly, the anterior patella experiences tensile stresses during the periods of articular compressive force [12].

A fall forward with the knee bent can result in a transverse patella fracture. The eccentric contraction of the quadriceps muscle coupled with the sudden flexion of the knee from the weight of the body can pull the patella apart, resulting in a transverse fracture. Stellate and comminuted fracture patterns commonly result from a direct blow to the anterior patella, such as from a fall from height onto the knee or a dashboard impaction injury. Often, the mechanism of injury is mixed, and the patient can present with a primary transverse fracture line from a distraction injury in addition to comminution from the subsequent impact during the fall.

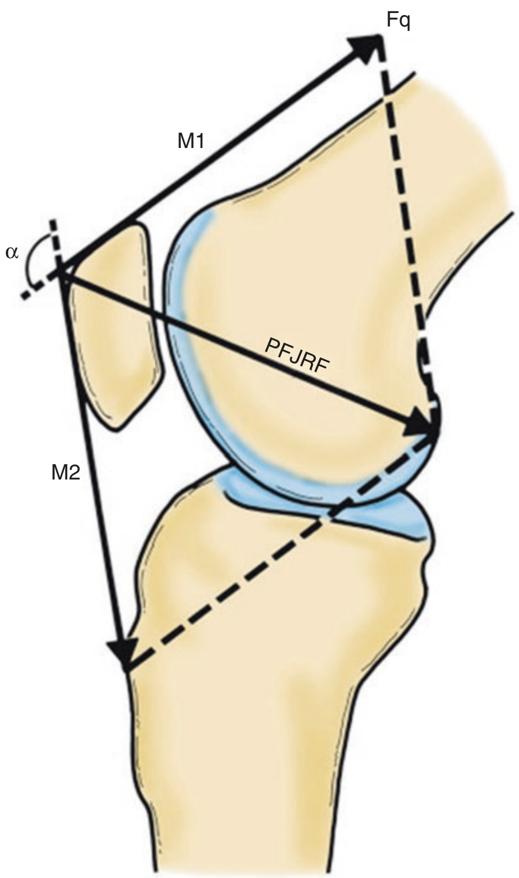

Fig. 11.2 The patella acts as a lever for the opposing forces of the quadriceps tendon (M1) and the patellar tendon (M2), while the femoral condyles act as a fulcrum. The patella increases the moment arm of the extensor mechanism and increases the efficiency. A patellofemoral joint reaction force (PFJRF) results. (From Dejour and Saggin [7] ©2018, with permission from Elsevier)

An intact extensor mechanism is essential for normal gait and for climbing activities, as it is the only means to initiate and maintain extension force at the knee. A complete disruption of the extensor mechanism is an indication for surgical management. However, due to the extremely robust properties of the retinaculum and its interconnections with the quadriceps and patellar tendons, it is possible for the patella to be fractured within the encasing soft tissues while the overall integrity of extensor mechanism remains preserved. In these cases, with an intact extensor mechanism, nonoperative management may be appropriate.

General Management Principles

The ultimate goals of patella fracture management are to restore an intact extensor mechanism and to reconstruct the body of the patella to allow a biomechanical advantage for the quadriceps to extend the knee joint. Additional goals are to restore a smooth, anatomic articular surface and to provide stable fixation to allow early knee range of motion. The knee, as with any synovial joint, is predisposed to stiffness when immobilized for an extended period of time. Early range of motion of the joint is one of the desirable aspects of operative reduction and fixation of the patella. This prevents flexion and extension contractures of the knee joint and enhances muscular strength around the knee.

When the extensor mechanism and the patella are damaged sufficiently to prevent the previous goals from being met, then surgical management is indicated. A tension band construct is designed to transform the distracting forces of the quadriceps and patellar tendon on the patella into compressive forces across the fracture site, as explained in Chap. 9. Most patella fractures are amenable to reconstruction using tension band principles, although some will require additional fixation augmentation.

Case 1

A 59-year-old woman sustained a fall onto her flexed right knee from standing height when she tripped over a curb. She had pain in her knee with range of motion and weight-bearing attempts and was brought to hospital for assessment. This was an isolated, closed injury.

X-rays are shown (Fig. 11.3). These demonstrate a minimally displaced transverse patella fracture with minimal comminution visible on the anteroposterior view.

The patient's extensor mechanism was examined and found to be intact. Therefore, nonoperative management was pursued after a discussion with the patient. She was initially splinted in extension and kept non-weight-bearing with crutches until clinic follow-up at 2 weeks' time

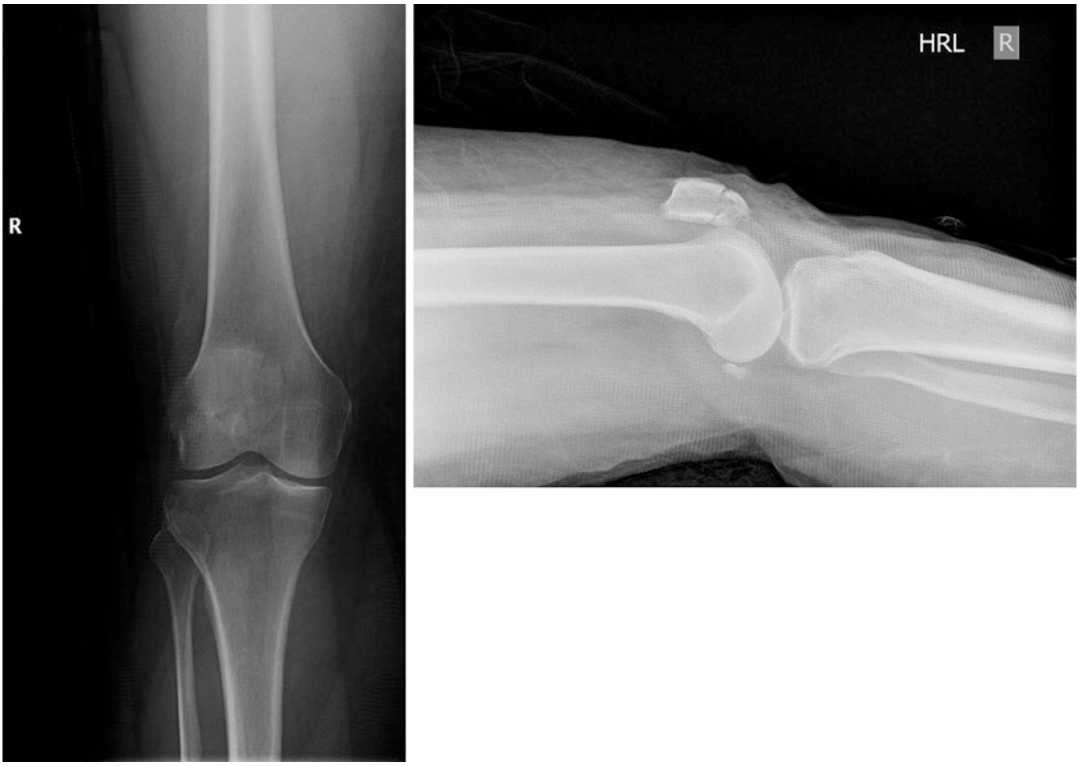

Fig. 11.3 Anteroposterior and lateral radiographs demonstrating a minimally displaced transverse patella fracture

(Fig. 11.4). At this point, she was transitioned to a removable knee immobilizer for mobilization, and her range of motion was gently advanced. Weight bearing as tolerated in and extension brace is permitted from the first clinic visit onwards (see section "Nonoperative Management").

After 4 months, the fracture was radiographically and clinically healed (Fig. 11.5). The patient had returned to full daily activities but was still working on targeted quadriceps and hamstring strengthening exercises at the gym to regain optimal strength.

Physical Examination

Patella fractures may present as open injuries due to the thin overlying soft tissue envelope at the level of the knee. A careful examination of skin integrity is important to ensure an open fracture is not missed. Open injuries should receive standard open fracture management including prompt administration of antibiotics and tetanus prophylaxis, provisional debridement of gross debris, splinting, and urgent surgical irrigation and debridement.

If a patella fracture is grossly distracted (as evidenced by a palpable gap on physical examination or a displaced fracture on x-ray), then the fracture meets operative indications due to an obvious disruption of the extensor mechanism. Physical examination of the integrity of the extensor mechanism is not required in these situations. If, however, complete disruption of the extensor mechanism is in doubt, then physical examination is essential. In an acute setting, pain will often prevent the patient from performing a straight leg raise off of the bed against gravity even if the extensor mechanism is intact. In these situations, have the patient lie supine on the examination bed. Extend the patient's leg by placing your hand under the heel and lifting the foot. Support the flexed thigh either by placing a bump under the thigh or by placing the examiner's contralateral arm under the thigh. Ask the

Fig. 11.4 Anteroposterior and lateral radiographs at a two-week follow-up, with the patient splinted in extension. No further displacement has occurred

Fig. 11.5 Anteroposterior and lateral radiographs demonstrating a healed fracture 4 months from injury

patient to maintain the straight leg raise and gently lower your hand from the heel. Most patients will be able to maintain the extension, at least for a few moments, if their extensor mechanism is intact. A disrupted extensor mechanism will prevent the patient from being able to keep the leg straight at all, and the foot will follow your hand as you attempt to lower it away. The exam can also be performed with the patient in the lateral decubitus position to eliminate gravity, although this positioning can make it harder to elucidate frank disruption.

Patella fractures are associated with concomitant injuries such as multi-ligamentous knee injuries, femoral condyle fractures, tibial plateau fractures, and femoral neck fractures. Dashboard impaction mechanisms are particularly high risk for associated injuries. Perform a full physical exam of the affected extremity to rule out additional injuries.

Nonoperative Management

The requirements for nonoperative management of patella fractures are a minimally displaced fracture, articular joint incongruity of <1–2 mm, and an intact extensor mechanism as demonstrated on physical examination [13, 14]. Note that it is difficult to assess articular joint incongruity on x-ray due to the radiolucency of the thick cartilage; however, displacement of the osseous patella can be used as a proxy for intra-articular displacement. Vertical fracture patterns almost always retain a competent extensor mechanism.

There is no universally accepted protocol for nonoperative management of patellar fractures. To our knowledge, the 1993 series by Braun remains the largest published group of nonoperatively managed patients in modern orthopedics [14]. They reviewed 40 conservatively managed patella fractures and found 80% to be "free of pain" with the remaining 20% experiencing "occasional pain." They reported 90% to have regained full range of motion. Their protocol involved immobilization in full extension for 3–4 days and then initiation of passive range of motion within the limits of pain. They did not immobilize the limbs after the initial immobilization period and allowed partial weight-bearing with crutches until the fractures were healed.

Our practice is to immobilize the limb in a bulky dressing with an anterior slab in extension for 1 week. We do not routinely aspirate the hemarthrosis. After 1 week, repeat radiographs are obtained to rule out interval displacement. The patient is transitioned into a removable knee immobilizer which is worn at all times except for when working on range-of-motion exercises. The patient is allowed to weight bear as tolerated with a knee extension splint, with crutches. Both passive and active range of motion from 0 to 90 degrees is allowed during this time. After 6 weeks, the patient is allowed to advance to weight-bearing as tolerated and the knee extension splint is weaned off over 10 to 14 days. Knee range of motion is allowed without restriction. Once the patient is ambulating without crutches (usually 6 to 8 weeks), targeted quadriceps and hamstring strengthening is initiated.

Case 2

A 54-year-old male sustained a fall forward onto his knees while at work. He had immediate pain to his right knee but was able to ambulate home with assistance. He presented to hospital the next day. Physical exam demonstrated a large knee effusion with ecchymosis over the right knee. Knee range of motion was 10–90 degrees, limited by pain. His extensor mechanism was demonstrated to be intact on physical examination. Radiographs demonstrated a vertical patella fracture with less than 2 mm of gapping (Fig. 11.6).

The patient was treated nonoperatively. At three-month follow-up, he was ambulating independently and working on strengthening. Radiographs demonstrated a healed fracture (Fig. 11.7).

Vertical Fractures

Vertical fractures are most commonly caused by a direct blow to the knee, either from a fall or an impaction (i.e., hockey boards or knee-on-knee contact). These are stable fracture patterns which do not disrupt the integrity of the extensor mechanism. Therefore, they rarely require surgical intervention unless there is significant displacement. Since the force exerted by the pull of the patella tendon and the quadriceps tendon is parallel to these fractures, there is no risk of further displacement from muscle contraction. It is safe to advance knee range of motion more quickly with vertical fractures than with transverse or stellate fracture patterns. Our practice is to splint the patient in extension for 1 week for comfort and then advance knee range of motion as tolerated. Weight-bearing as tolerated with crutches is allowed from the start. An extension splint is used for pain relief for 2 to 3 weeks and crutch walking and full range of motion is permitted. The patient is weaned off crutches once they are free of pain (usually 6–8 weeks).

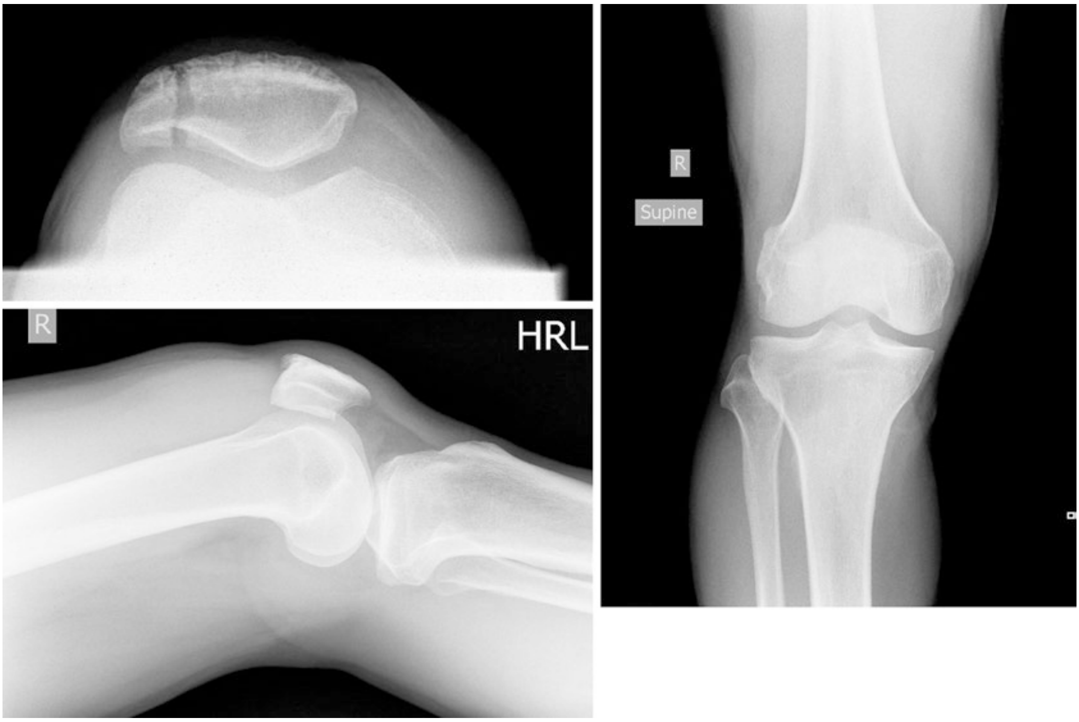

Fig. 11.6 These skyline, lateral, and anteroposterior radiographs demonstrate a minimally displaced vertical patella fracture

Case 3

An 18-year-old male sustained an injury to his right knee when he impacted the boards while playing elite hockey. He was unable to bear weight without assistance. Physical exam demonstrated a significant knee effusion which limited range of motion. The extensor mechanism was intact on physical exam. This was a closed injury. Radiographs demonstrated a minimally displaced vertical patella fracture (Fig. 11.8).

Nonoperative management was offered, but after discussion with the patient and his parents, the decision was made to undergo surgical treatment to facilitate faster rehabilitation and return to sport. The fracture was clamped and two 3.5-mm partially threaded screws were placed percutaneously from a medial to lateral direction to maintain compression (Fig. 11.9). After 6 weeks, the fracture had healed clinically and radiographically (Fig. 11.10). He was able to return to his elite sport career.

Case 4

A 30-year-old male fell off his dirt bike at speeds of around 30 km/h, landing on his left knee. He had significant swelling and inability to bear weight. There were superficial abrasions over his knee, but the injury was closed. A palpable defect was present at the knee which was tender to palpation. The extensor mechanism was clearly disrupted clinically. X-rays revealed a transverse patellar fracture with an additional lateral fracture fragment (Fig. 11.11).

Due to the wide displacement and disruption of the extensor mechanism, surgical management

Fig. 11.7 Skyline and anteroposterior radiographs of a healed fracture 3 months from injury

of the patient was recommended. He underwent modified anterior tension band wiring with Kirschner wires and two figure-of-eight stainless-steel wires (Fig. 11.12).

He was mobilized weight bearing as tolerated with a knee immobilizer. He could remove the knee immobilizer to shower as required. At the six-week follow-up appointment, the fracture was radiographically healed, and he was allowed to begin resisted strengthening exercises (Fig. 11.13).

Modified Anterior Tension Band Wiring with Kirschner Wires

With a transverse fracture pattern and disruption of the associated retinaculum, the opposing forces of the quadriceps tendon and patellar tendons create a distracting force across the fracture site with each quadriceps contraction or flexion of the knee joint. A tension band construct transforms these distracting forces to

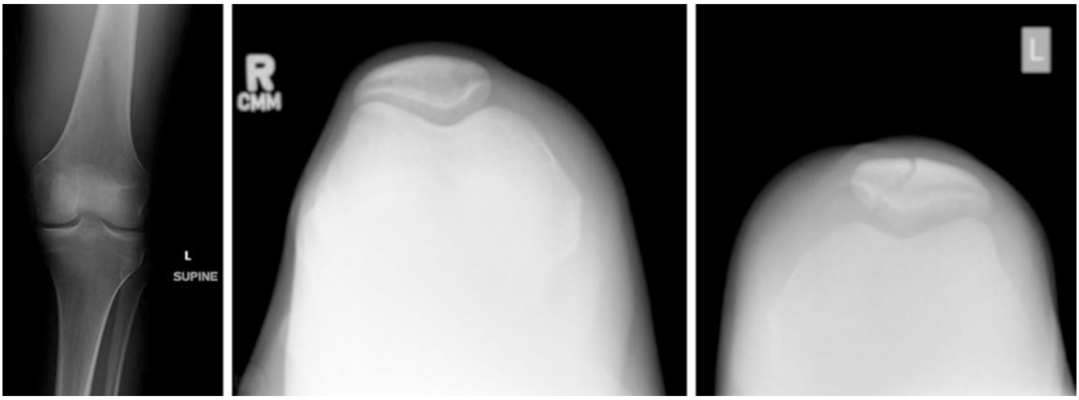

Fig. 11.8 Anteroposterior and skyline views of a minimally displaced vertical patella fracture

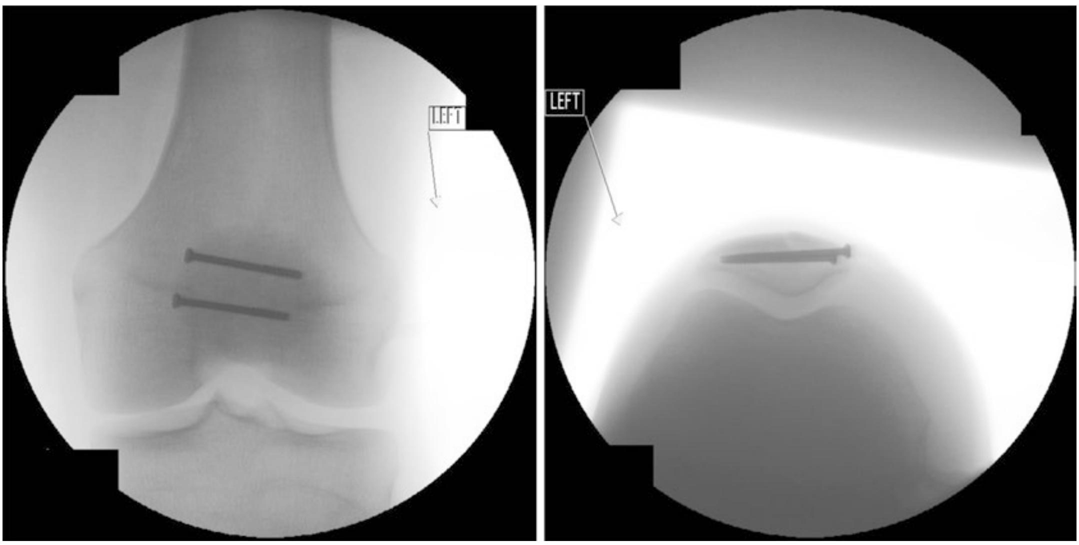

Fig. 11.9 Intraoperative fluoroscopy shots demonstrating final construct with two percutaneously placed transverse screws

compressive forces at the level of the articular surface (see Chap. 10). Many types of tension band constructs for the patella are described, and these can use a variety of materials, including Kirschner wires (K-wires), heavy gauge steel wire, cannulated screws, metal mesh, low-profile anterior plates, nonabsorbable suture, and nonabsorbable tape.

One of the most common and least expensive constructs is the modified anterior tension band using a figure-of-eight heavy gauge wire around two longitudinal Kirschner wires.

Surgical Technique: Modified Anterior Tension Band Wiring with Kirschner Wires

The patient is positioned supine on a radiolucent table with a bump under the ipsilateral hip and

Fig. 11.10 Skyline, anteroposterior, and lateral radiographs demonstrating a healed patella fracture with excellent joint congruity 6 weeks after injury

the leg on a radiolucent foam ramp. A midline approach to the knee is used with the incision extending approximately 3 cm superior and distal to the poles of the patella. Care must be taken to remain superficial to the retinaculum and periosteum of the patella in order to preserve as much blood supply as possible. This layer is most easily identified proximally at the level of the quadriceps tendon. The knee joint is irrigated through the fracture site, and any adherent organized hematoma at the fracture ends is removed to facilitate better interdigitation of the fracture ends. A sharp reduction clamp is used to reduce and compress the fracture. The clamp tines should be placed at the midline aspect of the superior and inferior patella poles so that they

Fig. 11.11 Anteroposterior and lateral radiographs of a transverse patella fracture with mild comminution

will not interfere with K-wire insertion. Alternatively, two peripherally placed clamps can be used. With the knee extended to relax the quadriceps, articular reduction is confirmed by digital palpation, exploiting any retinacular rents which are already present. Do not make an additional incision in the retinaculum as this will further disrupt the already compromised blood supply to the patella. Intraoperative fluoroscopy can be used to confirm reduction, but beware of the complex nature of the lateral and medial facets of the patella [4]. Oblique views of 20 degrees external rotation and 30 degrees of internal rotation should be obtained in addition to a true lateral to properly visualize the articular surface of the patella and to confirm reduction [4]. Once reduction is confirmed, two 1.6-mm K-wires are passed in parallel across the fracture. The wires should enter and exit the patella as close to the articular surface as possible, just above subchondral bone. Techniques using an anterior cruciate ligament (ACL) tunnel drill guide to position and pass the wires have been described but are usually not necessary. Next, an 18-gauge stainless-steel wire is passed in a figure-of-eight fashion around the K-wires. Care is taken to pass

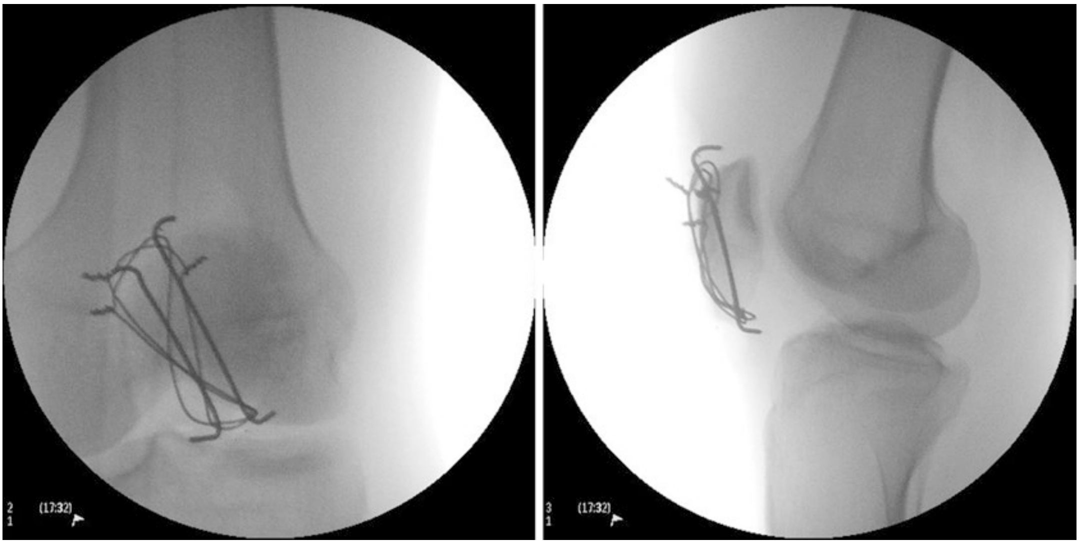

Fig. 11.12 Intraoperative fluoroscopic images of a modified anterior tension band construct with Kirschner wires and figure-of-eight stainless-steel wires

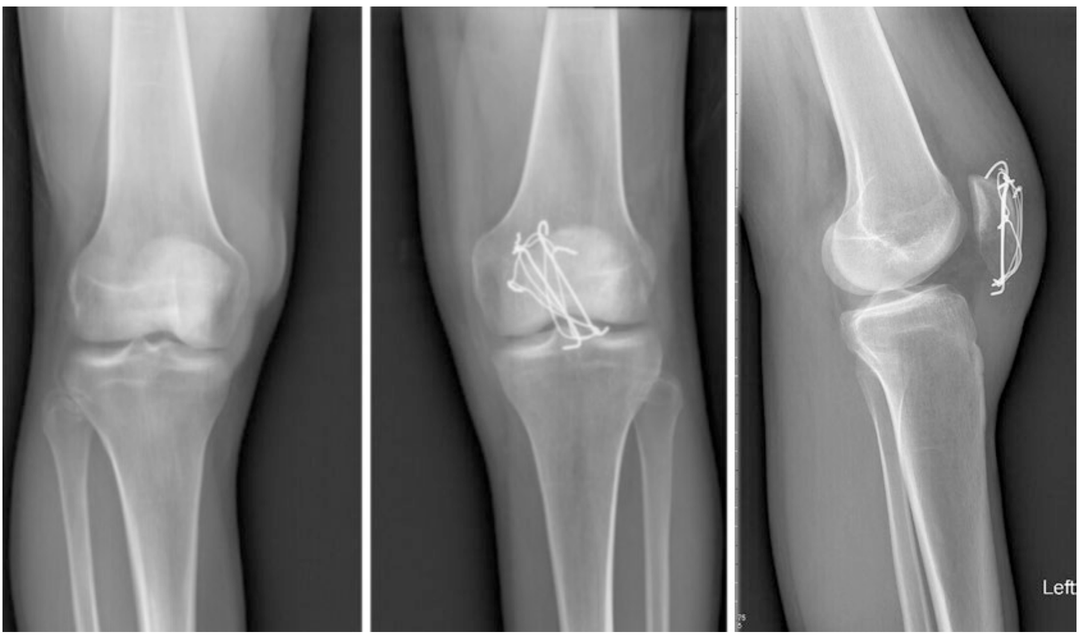

Fig. 11.13 Anteroposterior and lateral radiographs demonstrating a healed patella fracture 6 weeks from injury with no hardware failure

the wire deep to both K-wires and through soft tissue, as close to bone as possible. A 16-gauge hypodermic needle can be used to assist in passing the wire through the soft tissue. The wire is then twisted to tighten and generate compression across the fracture. No clinical difference has been demonstrated at 1 year for using a single point for wire tension twists versus using two corners for wire tensioning twists; however, limited biomechanical evidence suggests two sites of twist may be preferable for maintaining sustained compression [15, 16]. Our practice is to have two sites of wire twisting with each tightened in an alternating fashion. Once the wire is tight, the knee is cycled through a few bends to 90 degrees of flexion. The wire is then re-tensioned as needed and the ends are trimmed. Care is taken to bury the twists in soft tissue to minimize prominence. The inferior K-wires are then bent, cut, rotated, and impacted superiorly to capture the inferior wire loop. The superior ends of the K-wire are then similarly bent, cut, and rotated to capture the superior wire. We prefer to loop both ends of the K-wires to limit migration, as demonstrated in Case 5. The retinaculum is repaired with heavy absorbable suture, and the incision is closed in layers. It is essential to repair the retinaculum as this is an important contributor to the integrity of the extensor mechanism.

Postoperatively, the patient is placed in an anterior slab splint in full extension, non-weight-bearing with crutches, for 2 weeks until clinic follow-up. A wound check is performed and sutures are removed. The patient is then placed in a removable knee immobilizer. Mobilization is again allowed with the patient restricted weight bearing as tolerated with crutches and a knee immobilizer in place to non-weight-bearing with crutches and the knee immobilizer in place. Knee range of motion out of the immobilizer is permitted as tolerated, but no resisted extension is allowed. After 6 weeks, weight-bearing as tolerated is allowed, and the knee immobilizer is weaned off over 7 to 10 days. Resisted extension exercises and aggressive range-of-motion therapy are allowed at this point. After 3 months, all restrictions are removed.

Case 5

A 65-year-old female sustained a ground level fall resulting in a simple transverse patella fracture. This was treated with modified anterior tension band wiring using Kirschner wires (Fig. 11.14). Unhappily, by the patient's six-week follow-up appointment, the wires had already begun to back out (Fig. 11.15a, b). By 3 months, there was significant back-out of the K-wires; however, the fracture had fortunately united (Fig. 11.15c, d). This happened as the wires were not bent inferior to the patella. The wires should be bent at 90 degrees or more at either end of the patella. The patient was booked for hardware removal electively. This demonstrates the potential benefit of bending both ends of the Kirschner wires.

Case 6

A 61-year-old woman sustained a fall. She had pain in her left knee and inability to ambulate. Examination in the emergency department revealed an incompetent extensor mechanism. This was a closed injury. Radiographs demonstrated a transverse patella fracture with displacement of greater than 3 mm at the articular surface (Fig. 11.16).

She underwent surgical fixation with partially threaded cannulated screws and a heavy gauge stainless-steel tension band wire. This is a good technique but must use partially threaded screws which provide compression at the fracture site and the bone has to be reasonable and not comminuted. Radiographs at 1 year demonstrate a healed fracture with good maintenance of alignment (Fig. 11.17).

Tension Band Wiring Using Cannulated Screws

Issues with failure of fixation and painful hardware have motivated surgeons to search for a better construct for patellar fractures. Tension band wiring through cannulated screws has been suggested as an alternative. Partially threaded cannulated screws provide improved interfragmentary compression at the articular surface, while the figure-of-eight wire supports the anterior patella and neutralizes tension forces during quadriceps contraction. Biomechanical studies using cadaver models have shown cannulated screw tension band wire constructs to maintain compression during cyclical knee range of

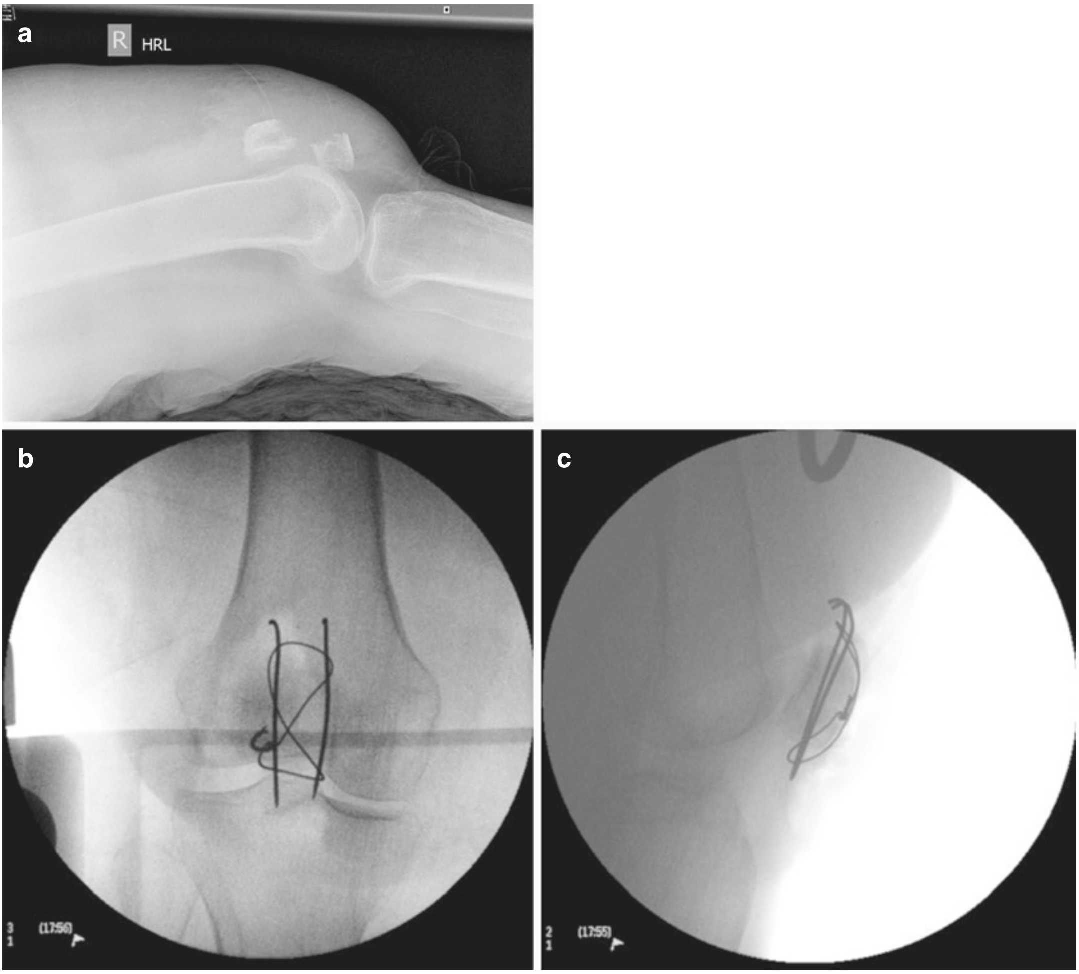

Fig. 11.14 (**a**) Lateral radiograph demonstrating a displaced, distracted transverse patella fracture. Anteroposterior (**b**) and lateral (**c**) intraoperative fluoroscopic images demonstrating a modified anterior tension band construct. Note the single hook on the ends of the Kirschner wires

motion better than Kirschner wire constructs; however, we are aware of no studies demonstrating a clinical difference in healing [17]. Multiple studies have shown lower rates of hardware removal with cannulated screw constructs than with K-wire constructs, although these are retrospective studies [18–20].

Surgical Technique

The patient is positioned, the patella is exposed, and the fracture is reduced as described for tension band wiring technique. Instead of 1.6-mm Kirschner wires being passed parallel in a longitudinal fashion, the 1.25-mm guide wires from the cannulated screw set are used. Use of the guide sleeve is suggested as the wires have a tendency to bend in hard bone. Position both wires so that the tips of the wires are just at the far edge of the patella, as confirmed with fluoroscopy, and then measure over the wires. Select a screw length about 5 mm shorter than measured so that the tips of screws do not protrude beyond the bone. Protruding screw ends are hypothesized to increase stress on the wire, leading to early failure. Once measurements have been obtained, the wires are advanced through the patella so that

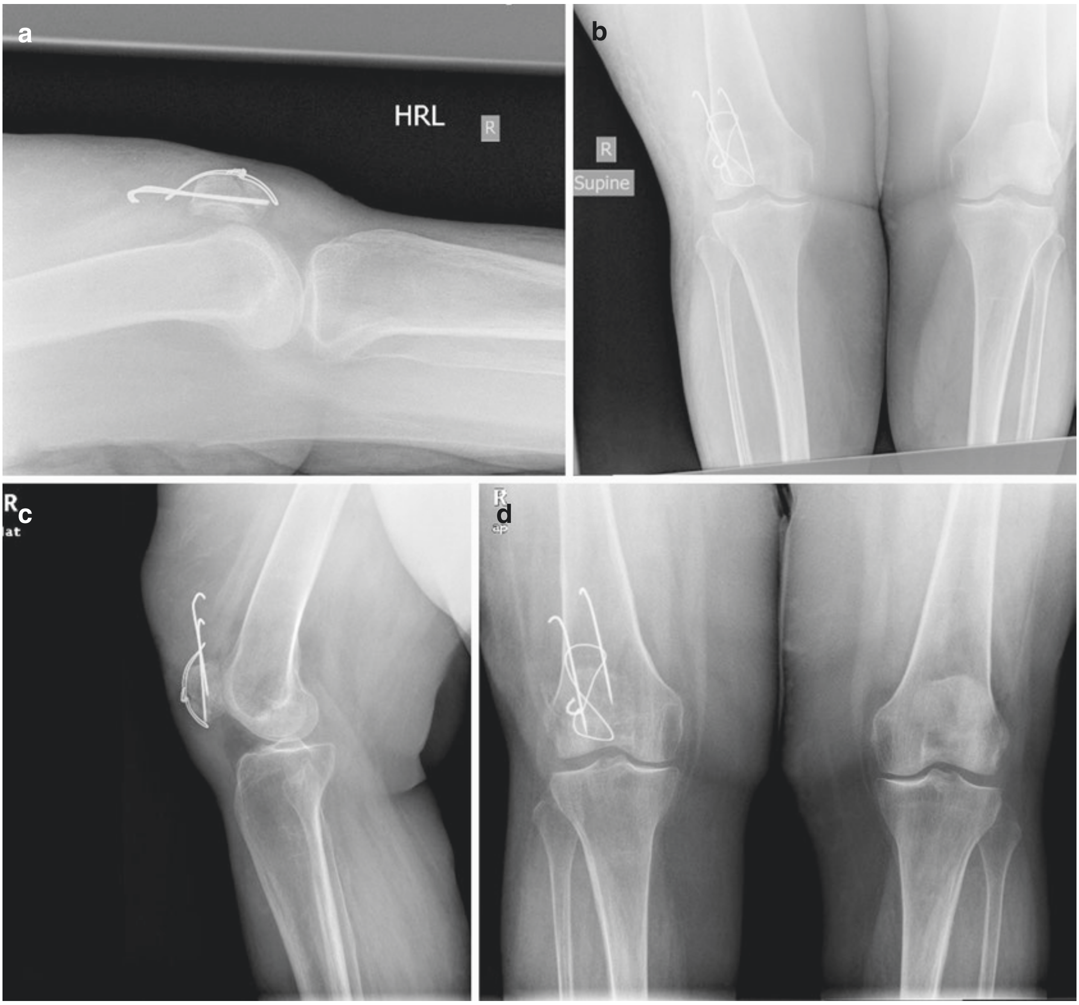

Fig. 11.15 At the six-week follow-up appointment, the Kirschner wires are seen backing out of the patella on the lateral (**a**) and anteroposterior (AP) (**b**) views. At 3 months from surgery, the Kirschner wires are almost completely backed out of the patella as seen on the lateral (**c**) and AP (**d**) views. Bending hooks on both ends of the Kirschner wires may prevent wire back-out

both are firmly seated. The ends of the wires are grasped with an instrument so that the wires are not removed inadvertently while drilling. Overdrill with the 2.7-mm cannulated drill, and then advance a 3.5-mm or 4.0-mm partially threaded cannulated screw until appropriate compression is achieved. The screws should be directed from the side with the smaller osseous fragment (i.e., if the inferior pole is the smallest fragment, the screws should be inserted from inferior to superior). Leave the screwdriver engaged in the screw head with the guide wire through both. Next, an 18-gauge straight stainless-steel wire is advanced through the cannulated screwdriver and through the screw, while the 1.25-mm guide wire is simultaneously removed from the opposite side. The process is repeated with the next screw so that two 18-gauge wires are through the patella. They are then bent and connected to each other in a figure-of-eight fashion with the two tensioning loops at the superior medial and superior lateral aspect of the construct. The figure of eight is

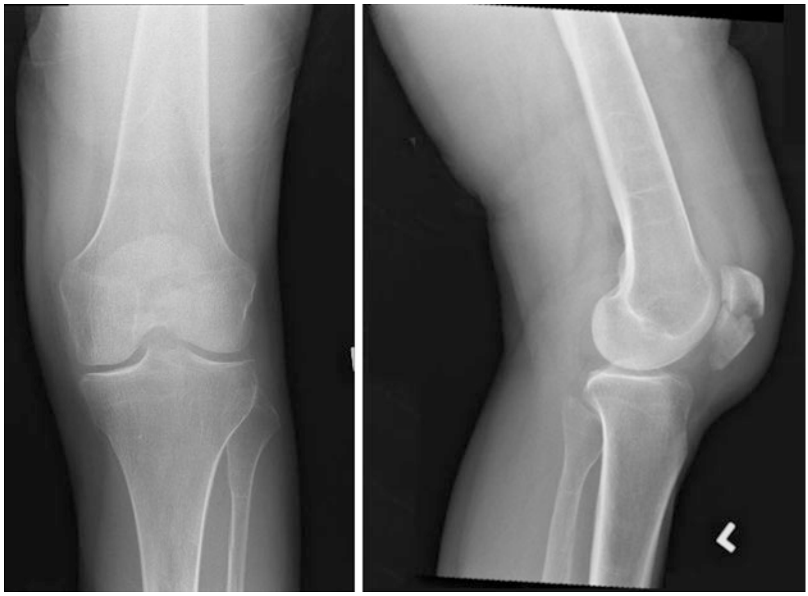

Fig. 11.16 Anteroposterior and lateral radiographs demonstrating a displaced, transverse patella fracture with minimal comminution

Fig. 11.17 Anteroposterior and lateral radiographs showing a healed patella fracture with partially threaded cannulated screws and a stainless-steel tension band wire construct. These radiographs were obtained 1 year from injury

tightened with alternating tensioning twists. The wires are then trimmed, and the wire twists buried into the quadriceps tendon. The retinaculum is repaired, and the knee is closed as previously described. The leg is splinted in full extension.

Adjunctive Techniques

For highly comminuted or stellate fracture patterns, additional fixation techniques may be required. Whenever possible, we create a tension band construct with the main central fragments

Fig. 11.18 A stainless-steel cerclage wire can be a helpful adjunct to contain peripheral comminution

and then augment our fixation for the peripheral fragments.

A cerclage wire or nonabsorbable suture/tape threaded through the retinaculum at its interface with osseous fragments can be an excellent mechanism to contain comminuted fragments. Since the force experienced by the patella is mainly directed longitudinally, these peripheral fragments only require a minimal construct to hold them in place until the overlying retinaculum scars and bony healing can occur (Fig. 11.18).

Interfragmentary compression with small screws away from the main tension band hardware can also assist to capture larger fragments (Fig. 11.19) [21]. Note that if the retinaculum is fully disrupted, interfragmentary compression alone (in the absence of a tension band of some kind) is rarely strong enough to maintain fixation once range of motion is initiated, and it should only be used as supplemental fixation.

In cases with particularly tenuous fixation of the patella, either due to excessive comminution or poor bone quality, a defunctioning wire may be used. The goal of a defunctioning wire is to transmit force from a contracted quadriceps directly to the tibial tubercle as a load-bearing device, thereby bypassing the patella entirely. Theoretically this will prevent displacement of the osseous fragments while the patella is allowed time to heal. Defunctioning wires usually require a secondary operation for their removal and we use them sparingly. It is not uncommon for the wire to break prior to elective removal. We place a bicortical 3.5-mm screw through the tibial crest just distal to the tibial tubercle. An 18-gauge stainless-steel wire is then wrapped below the head of the screw and brought up through the quadriceps tendon. The wire is tensioned and secured with the leg in 30 degrees of flexion. Alternatively, the wire may be passed directly through a transverse tunnel drilled in the tibia.

Figure 11.20 demonstrates a defunctioning wire used to protect a patellar tendon avulsion repair in a 30-year-old man.

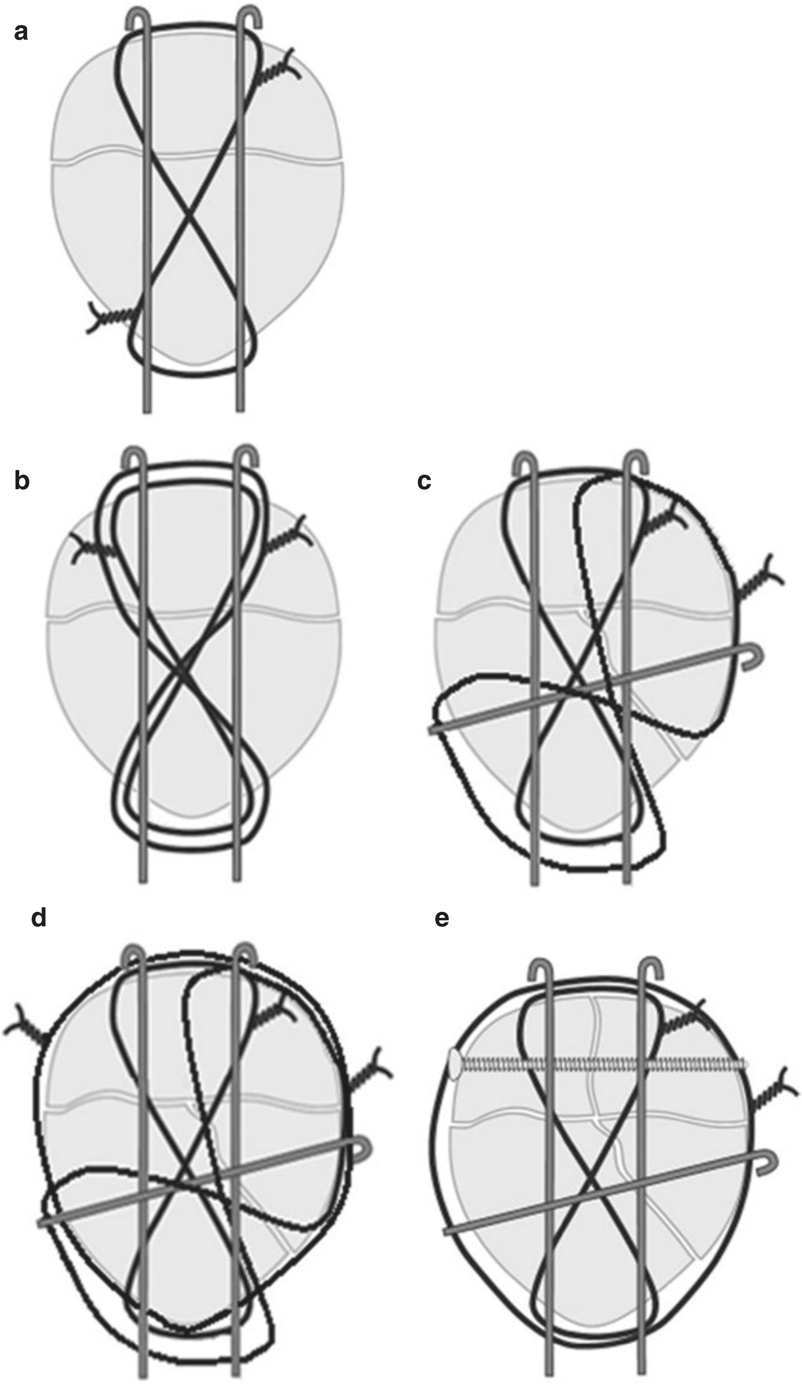

Fig. 11.19 Multiple constructs can be created using additional wires or interfragmentary screws as adjuncts to a primary tension band construct. (From Hambricht et al. [21], with permission. https://doi.org/10.1097/BOT.0000000000000686). (**a**) two wires with a figure of eight tension band. (**b**) two wires with two tension band wires tensioned on either side. (**c**) mutiplanbe wires with mutiplane tension band wires. (**d**) mutiplane wires with a citclage wire and tension band wire. (**e**) mutiplane wires plus a compression screw plus tension band wires

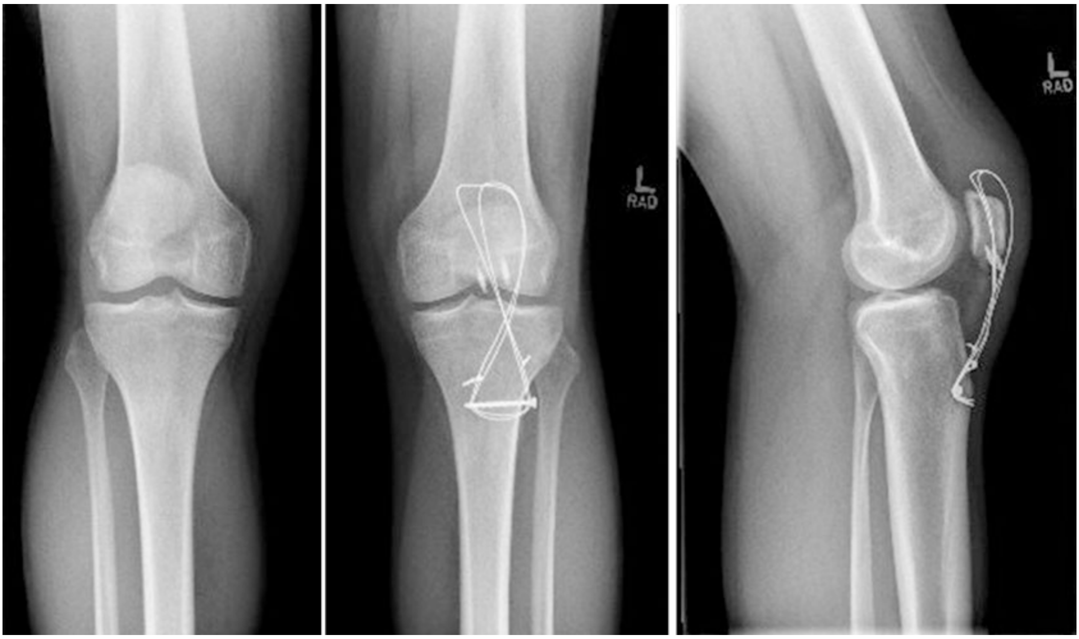

Fig. 11.20 Anteroposterior and lateral radiographs illustrating a defunctioning wire utilized to protect a patellar tendon avulsion repair. The wire here is placed in a figure-of-eight fashion and loops through the quadriceps tendon proximally and then wraps around a trans-tibial screw placed at the level of the tibial tubercle distally. Defunctioning wires generally require surgical removal once the injury has healed. Weight bearing as tolerated with and extension splint and crutches is utilized as above but ROM of the knee is not permitted for at least 6 weeks

Outcomes

Operatively managed patella fractures generally have excellent union rates. The rate of catastrophic loss of reduction requiring revision fixation is reported from 0% to 8%, with rates as high as 11% reported for extremely comminuted fractures [18, 19, 21–23]. Authors report the most common reasons for catastrophic failure are falls and technical failure (i.e., tension band loops not being posterior to K-wire struts or tension band wire not being directly adjacent to the patella) [19, 21, 22]. Failure can also occur if the wires are not bent at each end of the patella. Infection and soft tissue complications are rare, with most series reporting 0–4% incidence [19, 21, 22, 24]. Diabetes and immunosuppressive states are known risk factors for soft tissue complications [22, 25].

Anterior knee pain is often cited as a primary complaint after patella fixation; however, distinction is rarely made between pain from prominent hardware and intra-articular knee symptoms. It may be almost impossible to make this distinction. Anterior knee pain is a common complaint with nonoperatively managed patellar fractures as well [26]. It is hypothesized that even if articular reduction is restored to be near anatomic, chondral damage from the initial injury itself may be irreparable [24]. Additionally, quadriceps inhibition during the acute injury phase may lead to worsening knee kinematics and patellar maltracking, further exacerbating the complaints of anterior knee pain [24]. Despite excellent union rates for patellar fractures, long-term functional impairment is common. One study with average follow-up of 6.5 years noted that Knee Injury and Osteoarthritis Outcome Scores (KOOS) as well as physical component scores of the SF-36 were significantly worse than matched population norms, although it should be noted that only 36% of the eligible patients agreed to repeat assessment which could present significant bias [18]. Clinically significant strength and power deficits have also been measured to persist even at 12 months out from surgery [18, 24].

Hardware removal is commonly performed after patella fixation. Reported removal rates vary widely from 13% to 70%, with most authors reporting rates between 15% and 30% [18, 19, 21, 22, 24, 25, 27]. Although no randomized trials have been performed to directly compare the two fixation methods, steel tension band wiring using cannulated screws instead of Kirschner wires appears to require fewer operations for hardware removal, but may have higher failure rates if the bone is poor or comminuted [18–20]. Hoshino et al. performed a retrospective review of 448 operatively managed patella fractures and found that the odds ratio for hardware removal with K-wires compared to cannulated screws was 2.17 ($P = 0.002$) [19]. Hardware removal has been shown to correlate with improved visual analogue pain scores as well as improved quality of life scores; however, they have failed to demonstrate a change in functional outcome score after hardware removal [25]. The authors noted that diabetic patients had less consistent pain improvement after hardware removal.

Biomechanical studies on synthetic models and cadavers suggest that constructs using cannulated screws with tension band wiring provide more consistent compressive forces with less gapping after cyclic loading, as compared to tension band constructs using Kirschner wires [17, 28]. However, no clinical studies have demonstrated a statistically significant difference in union or failure rates, and no randomized controlled trials have been performed to our knowledge. There is very little high-quality evidence available on the use of alternative materials, such as nonabsorbable suture and bioabsorbable screws, for patella fractures. A recent systematic review suggested that use of these materials was safe with 90% of included patients being free of complications; however, the analysis was limited by the heterogeneity of the studies which were available [29].

Open reduction and internal fixation of the patella with tension band constructs is a reliable procedure with high union rates, although secondary hardware removal is commonly required.

References

1. Court-Brown CM, Clement ND, Duckworth AD, Biant LC, McQueen MM. The changing epidemiology of fall-related fractures in adults. Injury. 2017;48(4):819–24.
2. Eugene E. Extensile exposure of comminuted patella fractures using a tibial tubercle osteotomy: results of a new technique. J Orthop Trauma. 1998;12(5):351–5.
3. Sherman SL, Plackis AC, Nuelle CW. Patellofemoral anatomy and biomechanics. Clin Sports Med. 2014;33(3):389–401.
4. Berkes MB, Little MT, Pardee NC, Lazaro LE, Helfet DL, Lorich DG. Defining the lateral and accessory views of the patella: an anatomic and radiographic study with implications for fracture treatment. J Orthop Trauma. 2013;27(12):663–71.
5. Flandry F, Hommel G. Normal anatomy and biomechanics of the knee. Sports Med Arthrosc. 2011;19(2):82–92.
6. Gwinner C, Märdian S, Schwabe P, Schaser K-D, Krapohl BD, Jung TM. Current concepts review: fractures of the patella. GMS Interdiscip Plast Reconstr Surg DGPW. 2016;5:Doc01. https://doi.org/10.3205/iprs000080.
7. DeJour D, Saggin PR. Disorders of the patellofemoral joint. In: Scott WN, editor. Insall & Scott surgery of the knee. 5th ed. Philadelphia: Elsevier; 2012. p. 592–623.
8. Upadhyay N, Vollans SR, Seedhom BB, Soames RW. Effect of patellar tendon shortening on tracking of the patella. Am J Sports Med. 2005;33(10):1565–74.
9. Lazaro LE, Cross MB, Lorich DG. Vascular anatomy of the patella: implications for total knee arthroplasty surgical approaches. Knee. 2014;21(3):655–60.
10. Wurm S, Augat P, Bühren V. Biomechanical assessment of locked plating for the fixation of patella fractures. J Orthop Trauma. 2015;29(9):e305–8.
11. Escamilla R. Knee biomechanics of the dynamic squat exercise. Med Sci Sports Exerc. 2001;33(1):127–41.
12. Crowther MA, Mandal A, Sarangi PP. Propagation of stress fracture of the patella. Br J Sports Med. 2005;39(2):e6.
13. Sayum Filho J, Lenza M, Teixeira de Carvalho R, Pires OG, Cohen M, Belloti JC. Interventions for treating fractures of the patella in adults. Cochrane Database Syst Rev. 2015;2:CD009651. https://doi.org/10.1002/14651858.CD009651.pub2.
14. Braun W, Wiedemann M, Rüter a KK, Kolbinger S. Indications and results of nonoperative treatment of patellar fractures. Clin Orthop Relat Res. 1993;(289):197–201.
15. Lee SK, Hwang YS, Choy WS. Horizontal versus vertical orientation of the loop for tension band

wiring of transverse patella fractures. Orthopedics. 2014;37(3):e265–71.
16. John J, Wagner WW, Kuiper JH. Tension-band wiring of transverse fractures of patella. The effect of site of wire twists and orientation of stainless steel wire loop: a biomechanical investigation. Int Orthop. 2007;31(5):703–7.
17. Thelen S, Schneppendahl J, Jopen E, Eichler C, Koebke J, Schönau E, et al. Biomechanical cadaver testing of a fixed-angle plate in comparison to tension wiring and screw fixation in transverse patella fractures. Injury. 2012;43(8):1290–5.
18. Lebrun CT, Langford JR, Sagi HC. Functional outcomes after operatively treated patella fractures. J Orthop Trauma. 2012;26(7):422–6.
19. Hoshino CM, Tran W, Tiberi JV, Black MH, Li BH, Gold SM, et al. Complications following tension-band fixation of patellar fractures with cannulated screws compared with Kirschner wires. J Bone Joint Surg Am. 2013;95(7):653–9.
20. Egol K, Howard D, Monroy A, Crespo A, Tejwani N, Davidovitch R. Patella fracture fixation with suture and wire: you reap what you sew. Iowa Orthop J. 2014;34:63–7.
21. Hambright DS, Walley KC, Hall A, Appleton PT, Rodriguez EK. Revisiting tension band fixation for difficult patellar fractures. J Orthop Trauma. 2017;31(2):e66–72.
22. Smith ST, Cramer KE, Karges DE, Watson JT, Moed BR. Early complications in the operative treatment of patella fractures. J Orthop Trauma. 1997;11(3):183–7.
23. Malik M, Halwai MA. Open reduction and internal fixation of patellar fractures with tension band wiring through cannulated screws. J Knee Surg. 2014;27(5):377–82.
24. Lazaro LE, Wellman DS, Sauro G, Pardee NC, Berkes MB, Little MT, et al. Outcomes after operative fixation of complete articular patellar fractures : assessment of functional impairment. J Bone Joint Surg Am. 2013;95(e96):1–8.
25. Greenberg A, Kadar A, Drexler M, Sharfman ZT, Chechik O, Steinberg EL, et al. Functional outcomes after removal of hardware in patellar fracture: are we helping our patients? Arch Orthop Trauma Surg. 2018;138(3):325–30.
26. Bostrom A. Fracture of the patella: a study of 422 patellar fractures. Acta Orthop Scand. 1972;43(Suppl143):1–80.
27. Tian Q, Hai Y, Du X, Xu Z, Lu T, Shan L. Comparison of tension-band wiring with the cable pin system in patella fractures: a randomized prospective study. J Orthop Trauma. 2015;29(12):459–63.
28. Thelen S, Schneppendahl J, Baumgärtner R, Eichler C, Koebke J, Betsch M, et al. Cyclic long-term loading of a bilateral fixed-angle plate in comparison with tension band wiring with K-wires or cannulated screws in transverse patella fractures. Knee Surg Sports Traumatol Arthrosc. 2013;21(2):311–7.
29. Camarda L, Morello S, Balistreri F, Arienzo AD, Arienzo MD. Non-metallic implant for patellar fracture fixation: a systematic review. Injury. 2016;47(8):1613–7.

Part IV

Plating Principles with Case Examples

Biomechanics of Plate and Screw Constructs for Fracture Fixation

12

Mauricio Kfuri, Fabricio Fogagnolo, and Robinson Esteves Pires

Introduction

Bone is a unique tissue due to its mechanical properties and the ability of self-repair. Fractures result from mechanical variables, including the magnitude and direction of applied loads as well as the structural properties of the bone, which are determined by its density and physical structure [1–3]. The surgical treatment of fractures gained popularity with the introduction of the principles of fracture care by the Association for the Study of Internal Fixation (ASIF) [4]. Stable fixation of fractures has been a significant advance in fracture management allowing for bone healing while maintaining the function of the joints. In the 1960s, compression of the fracture site through absolute stability was considered the recipe for successful outcomes. Anatomical reduction and absolute stability, however, required a more extensive surgical exposure to the fracture site, resulting in a second hit to an area where the index trauma had already compromised the vascular supply of bone fragments. In the 1990s, the emphasis changed to the internal "biological" fixation. The goals were to restore the length, alignment, and axis of the bone, utilizing indirect reduction and a bridging construct for non-articular fracture components [5, 6]. Plates and screws may be used to provide either absolute or relative stability to a fracture site, allowing respectively for primary bone healing or callus formation [7]. The same implant constructs may perform different biomechanical functions including neutralization, compression, buttressing, bridging, and tension band. In this chapter, the use of bone-plate constructs will be illustrated under the perspective of their biomechanical function and expected bone healing outcomes.

A Historical Perspective

The first plates designed for the fixation of fractures were introduced more than a century ago by Lane [8]. Those plates had poor metallurgical properties and were soon abandoned due to corrosion [9]. Robert Danis, in 1949, developed a new plate system which allowed for axial compression of the fracture [10]. This was a turning point for fixation of fractures with plates and screws. Anatomical reduction and absolute

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stability at the fracture site promoted union without callus formation. Compression of the fractures with plates became the primary goal of fracture treatment in the late 1950s. This was the mechanical era of internal fixation. Bagby and Janes proposed a plate with oval holes that would allow for axial compression of the fracture, depending on the way the screws would be inserted [11]. In 1965, the Arbeitsgemeinschaft für Osteosynthesefragen (AO) Group developed a tension device that could be coupled to one of the ends of the plate, allowing for axial compression [12]. The dynamic compression plate (DCP) was developed in 1969 [13]. This plate allowed for static axial compression of the fracture site once the screws on one end of the plate were applied eccentrically. Although compression plates proved to be beneficial for the treatment of fractures that had been anatomically reduced, there was some degree of cortical necrosis under the plate. This has been interpreted as a result of periosteal vascular compromise due to plate application [14]. Aiming to reduce the cortical necrosis under the DCP implants, newly designed plates with limited bone contact (LC-DCP) were developed [15]. It has not been proved, however, that this new generation of plates promoted less cortical necrosis.

Absolute stability with plate fixation for diaphyseal fractures created challenges. The most common issue was the lack of radiographic feedback about complete fracture healing, due to the absent callus formation, or even refractures associated with hardware removal [16]. The 1990s was the decade of the biological fixation. Callus formation was a desirable response in diaphyseal fractures, and the use of bridge plate constructs was associated with a smaller incidence of mechanical failures and infection [5, 6]. The history of plate development points out the evolution of concepts in fracture fixation. Anatomical reduction of the bone fragments is still pursued in the articular fractures, but not necessarily in the management of extra-articular ones.

The mechanical fixation of fractures with LC-DCP plates depended mainly on the torque of the screws and the friction generated between the hardware and the underlying bone. If the loading forces to the fracture site were higher than the combination of achieved torque and friction forces, the bone-implant construct would fail. In osteoporotic bones, the torque of the screws is compromised due to the thin cortices and limited thread purchase of the screws. The development of a new generation of implants was needed to overcome this challenge. The mechanical solution was to add threads to the screw heads and the plate holes. Therefore, screw-hole constructs became fixed angle units. Locking plates were designed to be more stable and biologically friendly [17–20]. Loading the fracture site once stabilized with a locking plate converts pull-out forces into compression forces to the screw-hole units. Periarticular locking plates are anatomically pre-contoured, allowing for the insertion of multiple angle stable screws into short epiphyseal segments [21, 22].

The metal alloys used to produce plates is another topic of relevance. Stainless steel has been used for decades due to its corrosion resistance, adequate strength, low cost, and intraoperative malleability allowing for easy contouring of the plates. More recently, implants made out of titanium alloys have gained popularity since their elastic modulus is closer to the bone compared to stainless steel, and they are considered to have better osseointegration properties and potentially lower infection risk. The newest trend is a generation of implants made out of carbon-fiber-reinforced-polyetheretherketone composite that has an elastic modulus even closer to the bone. Carbon-fiber plates are radiolucent and allow for easier intraoperative evaluation of fracture reduction and decreased artifact with computed tomography (CT), or magnetic resonance image (MRI) when evaluating bone healing or associated soft tissue damage. Future studies will determine if this new generation of implants proves to be beneficial in the clinical setting [23].

Plates are versatile implants which may be used for the treatment of the majority of the fractures of the skeleton. The complete understanding of the biomechanical properties of bone-plate constructs is critical for the internal fixation of fractures. The diversity of biomechanical bone-implant constructs is the reason why many surgeons do not consider plate fracture fixation a technique, but an art.

Biomechanical Functions of a Plate

Plates may result in absolute or relative stability of a fracture site. Absolute stability requires circumferential contact of the fracture site, which is obtained by anatomical reduction. Absolute stability is the principle of fixation pursued in the management of simple fracture patterns (Fig. 12.1), articular fractures, and hypertrophic nonunions. Relative stability is based on an indirect reduction of the fracture site aiming to restore the overall length, rotation, and alignment of the bone. Relative stability is mainly applied to the management of comminuted diaphyseal and metaphyseal fractures, where the anatomical reduction of every fragment will compromise the vascular supply of the fracture site. Biomechanically, bone-plate constructs may have a variety of functions, depending on the goals of the treatment. Compression, buttress, neutralization, and bridging and tension band are the main biomechanical functions of bone-plate

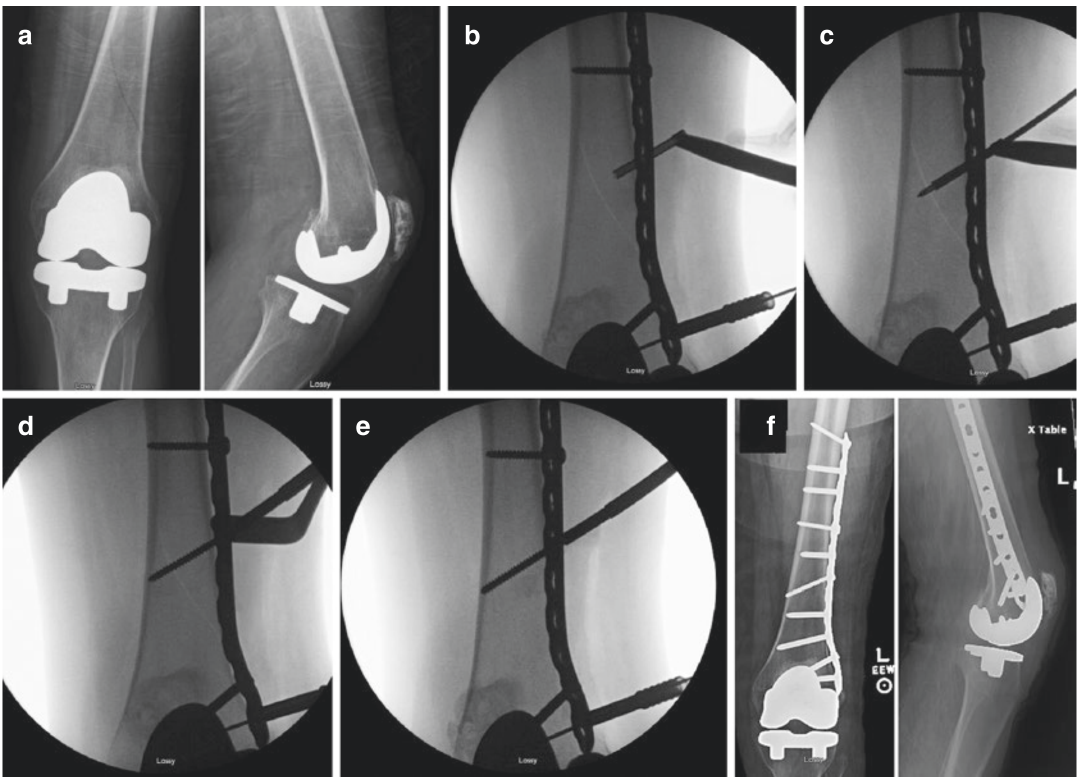

Fig. 12.1 Absolute stability using a lag screw through the plate. (**a**) Anteroposterior and lateral radiographic projections of the left knee revealing a non-displaced oblique simple periprosthetic fracture. (**b**–**e**) Intraoperative fluoroscopic images illustrating the step-by-step application of a lag screw through the plate. (**b**) After the initial fixation of the plate proximal and distal to the fracture site, the lateral cortex is drilled with a drill bit with a diameter that matches the outer diameter of the lag screw. A sleeve is inserted in this hole for the drilling of the opposite cortex. (**c**) A drill bit with a diameter that matches the core diameter of the lag screw is inserted through the sleeve reaching the opposite cortex. (**d**) The lag screw is inserted in the drill hole, and at this point, it has not engaged yet on the opposite cortex. (**e**) Once the screw engages the opposite cortex, it will compress the fracture site, and the fracture line disappears. (**f**) Immediate postoperative radiographs depicting an anatomical reduction of the fracture and the principle of absolute stability obtained by a plate applied to the tension surface of the bone and in association with a lag screw

constructs. Those functions are accomplished according to the surgical technique, not the specific plate, adopted for the application of the hardware.

Biomechanical Properties of a Plate

Bone segments are subject to bending, torsional, and axial forces. Bending is generated when an external load is applied perpendicularly to the longitudinal axis of the bone. Bending loads will result in tension and compression stresses relative to the cortices of the bone. Torsional loads determine the twisting of the bone by the exertion of forces tending to turn one of its ends about a longitudinal axis while the other end is held fast or turned in the opposite direction. Axial loads are those that are perpendicular to the cross section of the bone (Fig. 12.2).

The biomechanical properties of a bone-plate construct are dependable on the density of the bone, the fracture geometry, the thickness of the plate, and the friction between the plate and the bone. Stiffness is affected by the plate thickness—the thicker the plate, the greater the stiffness and resistance to bending forces. The bending stiffness of a plate is proportional to the third power of its thickness [24].

When a bone-plate construct is loaded, the forces are transmitted through the interface between the hardware and the underlying cortex. The stability of the construct is dependable on frictional and mechanical interlocking forces [25].

Non-locking plates rely on the friction generated between the plate and the bone by the torque of the screws. The higher the density of the bone, the higher the torque of the screws and the frictional forces. The loading forces are transmitted to the interface between the plate and the bone and also through the screw heads (Fig. 12.3).

Locking plates have a different principle. They function as internal fixators. The threaded screw heads engage the threaded holes of the plate establishing an angle stable unit. The loads are mainly transmitted through the implant, and the mechanical interlocking forces determine the stability of the bone-implant construct [26] (Fig. 12.4).

The distribution of the screws within a plate significantly impacts the biomechanics of bone-plate constructs [27]. The working length of a bone-plate construct is the distance between the first two screws on each side of the fracture (Fig. 12.5). The closer the screws are to the fracture site, the stiffer the construct. The screws that see the most load in the bone-plate construct are

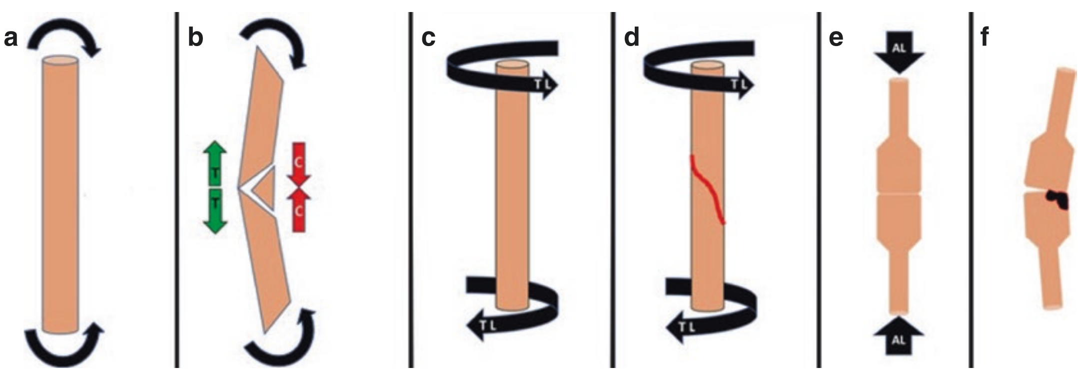

Fig. 12.2 Typical fracture patterns in association with different loading patterns. (**a**) Bending load; (**b**) split wedge fracture as result of bending forces. Observe the side of compression (*C*) and the side of tension (*T*). (**c**) Torsional load; (**d**) helical fracture pattern as a result of torsional forces; (**e**) axial load; (**f**) compression fracture of the joint as a result of an axial load

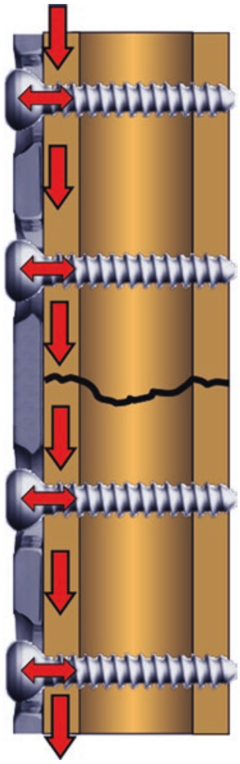

Fig. 12.3 Distribution of load through a non-locking bone-plate construct. The loads are transmitted through the fracture as well as the interface between the hardware and the underlying bone. The higher the density of the bone, the higher the torque of the screws and the friction between the hardware and the cortical bone

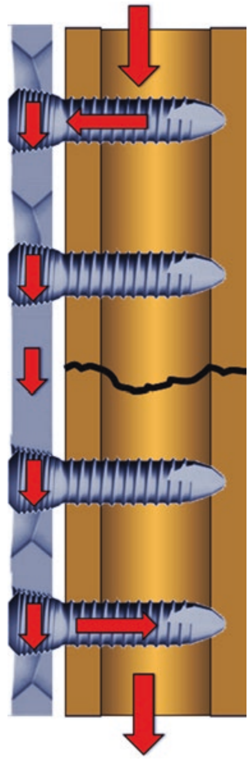

Fig. 12.4 Distribution of the load through a locking bone-plate construct. The loads are mostly transmitted through the plate and the angle stable units established between the threaded screw heads and the threaded plate holes. The density of the bone in this scenario is less relevant as high torque will be achieved at the interface between screw heads and plate holes

the screws closest and furthest from the fracture on each side. These are the screws that are subject to the higher pull-out forces.

The length of the plate and the distribution of the screws within plate holes affect the resistance of the construct to failure [28]. The longer the length of the plate on each side of the fracture and the more spread of the screws in the plate, the higher the resistance of the construct to pull-out forces (Fig. 12.6). The greater the distance between the inner and the outer screw on each side of the plate, the higher the control that the implant has over that bone segment, and the higher the resistance against pull-out forces. Torsional rigidity is increased by adding a third screw on each side.

Bones may be subjected to eccentric loading. This happens to the femur due to the eccentric position of the femoral head in relationship to the femoral shaft. In cases of bone malunions and nonunions, the convex side of the bone is the one subjected to tension, while the concave side is exposed to compression forces. Plates applied on the tension side of the bone may function as tension band devices, converting tension forces into compression ones (Figs. 12.7, and 12.8).

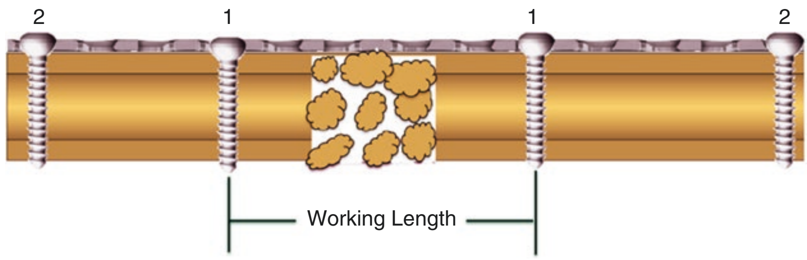

Fig. 12.5 The concept of fracture working length. A bone-plate construct is depicted. The inner screws (1) are those closest on each side of the fracture site. The outer screws (2) are the most distant ones on each side of the fracture site. The working length may be adjusted according to the fracture pattern and affects the flexibility to the fracture site

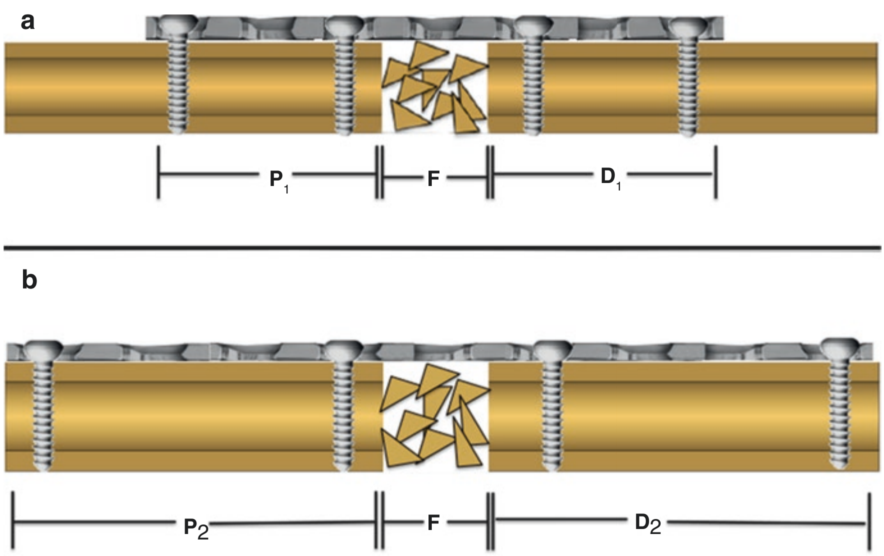

Fig. 12.6 The impact of plate length and screw distribution on bone-plate constructs. (**a**) Comminuted shaft fracture stabilized by a short plate. Observe the relationship between the length of the fracture site (F) and the length of the bone fixed by the plate on the proximal (P1) and on the distal (D1) bone segments. The smaller the ratio between the length of the plate on each side of the fracture and the length of the fracture, the higher the likelihood of a mechanical failure. (**b**) A comminuted fracture fixed by a long spanning plate. The length of each fixed bone segment (P2 and D2) is much higher than the length of the fracture. This allows for better control of each bone segment and increased stability to bending and torsional loads

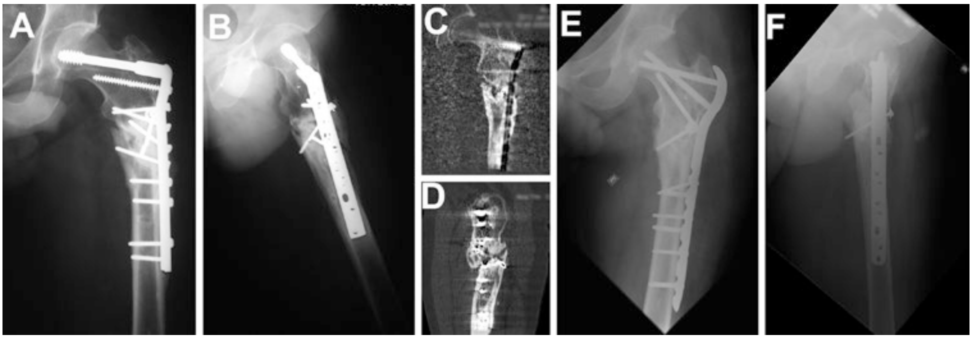

Fig. 12.7 Example of absolute stability with the use of a plate on the tension surface of the bone. (**a**, **b**) Anteroposterior and lateral radiographic projections of the proximal femur revealing a subtrochanteric nonunion, after multiple attempts of surgical treatment. Observe a broken lag screw at the fracture site and the varus angulation. (**c**, **d**) Computed tomography scan confirming the presence of a nonunion at the subtrochanteric level. (**e**, **f**) Final radiographs after surgical treatment of the nonunion and complete bone healing. The strategy was to perform a subtrochanteric closing wedge osteotomy at the level of the nonunion to resect the fibrous tissue associated with an atrophic nonunion. The osteotomy aimed to correct the varus deformity and was fixed with a plate applied to the tension surface of the femur. An articulating tension device was applied to the distal aspect of the plate to promote extra compression, before inserting the distal screws of the plate. A lag screw was applied outside of the plate to reinforce the compression. The sequence of the fixation was osteotomy, reduction, a plate fixed proximally, articulating tension device applied distally, eccentric screws applied distally, a lag screw applied from anterior to posterior, perpendicular to the fracture site. This is an example of multiple strategies to achieve absolute stability at the fracture site

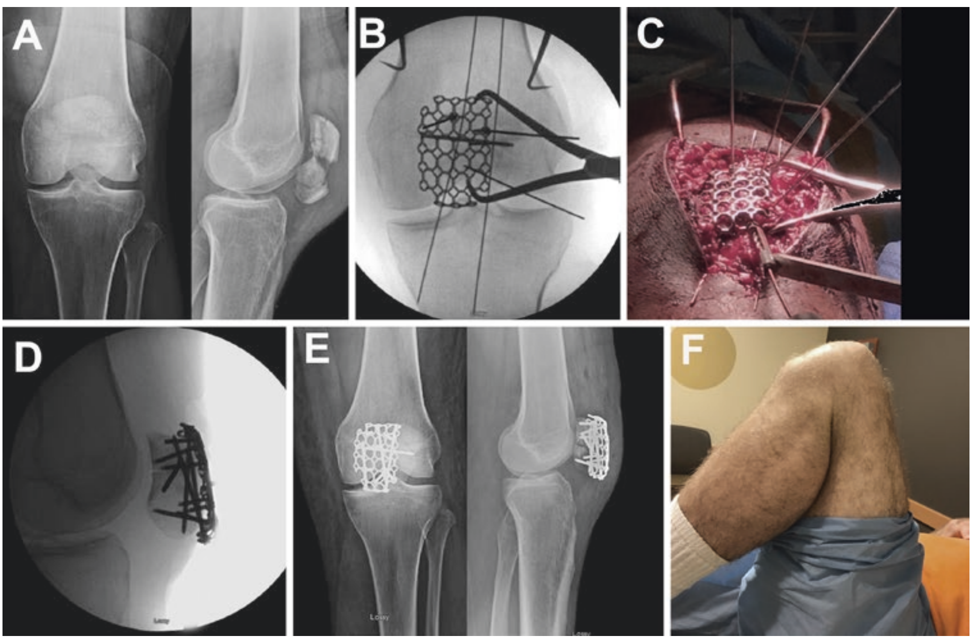

Fig. 12.8 Example of a dynamic tension band plate. (**a**) Radiographs illustrating a comminuted patellar fracture. (**b**) Intraoperative fluoroscopic control of the application of a low-profile locking plate to the anterior surface of the patella. (**c**) Intraoperative image illustrating the clinical application of a plate on the tension surface of the bone. (**d**) Fluoroscopic control after completion of the fixation revealing a satisfactory reduction. (**e**) Immediate postoperative radiographs. The plate is applied to the anterior surface of the patella, and it will convert tension forces into compression forces once the patient mobilizes the knee from extension to flexion. (**f**) Clinical outcomes after 6 months of the fracture fixation. The patient is asymptomatic and has a full range of motion

Conclusions

Plates and screws are essential tools in orthopedic surgery. They may be used with a broad spectrum of biomechanical functions allowing for either absolute stability or relative stability. Bone fixation with plates requires precise preoperative planning and meticulous execution to accomplish with the biomechanical goals of the fixation. The length and thickness of the plate, the distribution of the screws on the plate, the density of the bone, the friction generated between hardware and the underlying bone, the mechanical interlocking of the screws, and the characteristics of the screw heads and the plate holes are all determinants of the biomechanical properties of the bone-plate construct. Although many developments have been achieved in the area of hardware design and technology, the principles of fracture care remain the same, and the outcomes of treatment are directly related to the proper indication and application of the hardware. Subsequent chapters in this section will address the individual functions and biomechanical properties of both locking and non-locking plates.

References

1. Hayes WC. Biomechanical measurements of bone. In: Burstein A, editor. CRC handbook of engineering in medicine and biology: section B. Instruments and measurements. Cleveland: CRC Press; 1978. p. 333–72.
2. Hayes WC. Biomechanics of fracture healing. In: Heppenstall RB, editor. Fracture treatment and healing. Philadelphia: WB Saunders; 1980. p. 124–72.
3. Rahn BA, Gallinaro P, Baltensperger A, Perren SM. Primary bone healing: an experimental study in the rabbit. J Bone Joint Surg Am. 1971;53(4):783–6.
4. Mueller M, Allgower M, Willenegger H. Technik der operativen Frakturenbehandlung. Berlin: Springer; 1963.
5. Gerber C, Mast J, Ganz R. Biological internal fixation of fractures. Arch Orthop Trauma Surg. 1990;109(6):295–303.
6. Ganz R, Mast J, Weber B, Perren S. Clinical aspects of “bio-logical” plating. Injury. 1991;22:4–5.
7. Perren SM. Physical and biological aspects of fracture healing with special reference to internal fixation. Clin Orthop. 1979;138:175–96.
8. Lane WA. Some remarks on the treatment of fractures. Br Med J. 1895;1(1790):861–3.
9. Uhthoff HK, Poitras P, Backman DS. Internal plate fixation of fractures: short history and recent developments. J Orthop Sci. 2006;11(2):118–26.
10. Danis R. Theorie et practique de l’osteosynthèse. Paris: Masson & Cie Éditeurs; 1949.
11. Bagby GW, Janes JM. The effect of compression on the rate of fracture healing using a special plate. Am J Surg. 1958;95(5):761–71.
12. Müller ME, Allgöwer M, Willenegger H. Compression fixation with plates. In: Technique of internal fixation of fractures. Berlin: Springer; 1965. p. 47–51.
13. Perren SM, Russenberger M, Steinemann S, Müller ME, Allgöwer M. A dynamic compression plate. Acta Orthop Scand. 1969;135:31–41.
14. Perren SM, Cordey J, Rahn BA, Gautier E, Schneider E. Early temporary porosis of bone induced by internal fixation implants: a reaction to necrosis, not to stress protection? Clin Orthop. 1988;232:139–51.
15. Gautier E, Perren SM. [Limited Contact Dynamic Compression Plate (LC-DCP)–biomechanical research as basis to new plate design]. Orthopade. 1992;21(1):11–23. . Review. [Article in German]..
16. Kessler SB, Deiler S, Schiffl-Deiler M, Uhthoff HK, Schweiberer L. Refractures: a consequence of impaired local bone viability. Arch Orthop Trauma Surg. 1992;111(2):96–101.
17. Perren SM, Buchanan JS. Basic concepts relevant to the design and development of the point contact fixator (PC-Fix). Injury. 1995;26(Suppl 2):S-B1–4.
18. Tepic S, Perren SM. The biomechanics of the PC-Fix internal fixator. Injury. 1995;26(Suppl 2):S-B5–10.
19. Frigg R, Appenzeller A, Christensen R, Frenk A, Gilbert S, Schavan R. The development of the distal femur Less Invasive Stabilization System (LISS). Injury. 2001;32(Suppl 3):SC24–31.
20. Frigg R. Locking Compression Plate (LCP). An osteosynthesis plate based on the dynamic compression plate and the Point Contact Fixator (PC-Fix). Injury. 2001;32(Suppl 2):63–6.
21. Wagner M. General principles for the clinical use of the LCP. Injury. 2003;34(Suppl 2):B31–42.
22. Schmal H, Strohm PC, Jaeger M, Südkamp NP. Flexible fixation and fracture healing: do locked plating ‘internal fixators’ resemble external fixators? J Orthop Trauma. 2011;25(Suppl 1):S15–20.
23. Hak DJ, Mauffrey C, Seligson D, Lindeque B. Use of carbon-fiber-reinforced composite implants in orthopedic surgery. Orthopedics. 2014;37(12):825–30.
24. Hayes WC, Perren SM. Flexural rigidity of compression plate fixation (Nordic Meeting on Medical and Biological Engineering, 2d, Oslo, 1971. Proceedings). Med Biol Eng. 1971;2:242–4.
25. Hayes WC, Perren SM. Plate-bone friction in the compression fixation of fractures. Clin Orthop. 1972;89:236–40.
26. Wagner M, Frigg R, editors. AO manual of fracture management. Internal fixators: concept and cases using LCP and LISS. Stuttgart/New York: Thieme; 2006.
27. Tornkvist H, Hearn TC, Schatzker J. The strength of plate fixation in relation to the number and spacing of bone screws. J Orthop Trauma. 1996;10(3):204–8.
28. Gautier E, Sommer C. Guidelines for the application of the LCP. Injury. 2003;34(Suppl 2):63–76.

13 Nonlocking Plate Functions

Jonathan G. Eastman

Open reduction and internal fixation (ORIF) with plate and screw constructs has been successfully performed for over 100 years [1, 2]. While many surgeons opt for intramedullary nails or precontoured locking plates to stabilize some osseous injuries, many fractures are still amenable to reduction and fixation utilizing simple nonlocking screw constructs. The basis of stability of nonlocking plate and screw constructs relies heavily on the screw to generate enough insertional torque to generate contact between the plate and the bone and the ability of the screw to maintain that compression during load until fracture healing [2]. While many shapes, sizes, and thicknesses of nonlocking plates exist, the function of the plate is determined by the surgeon and how the plate is applied to a specific fracture. The main functions a plate can serve are compression, neutralization, buttress, tension band, and bridging. The first four plate functions are typically used when absolute stability is desired and primary bone healing is expected. Bridge plating is typically used when relative stability is desired and secondary bone healing is expected [2]. The purpose of this chapter is to review the main functions of nonlocking plating through descriptive clinical examples of when it was applied both correctly as well as to illustrate potential pitfalls to avoid.

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Compression Plating

Compression plating is commonly used for simple diaphyseal or metadiaphyseal fracture patterns. While this technique can be applied to any bone with a transverse or short oblique fracture, it is most utilized in the humerus, radius, ulna, clavicle, tibia, and fibula. When biologically friendly surgical exposure and anatomical reduction are achieved, and a biomechanically sound construct is applied, the anticipated union rate is greater than 95% [3–6]. Outside of acute fractures, compression plating has been used to treat nonunions of diaphyseal fractures in many settings with success [7–9].

While the ideal nonlocking plate and screw construct may vary between surgeons, variables like patient size, medical comorbidities, bone quality, and anticipated compliance need to be considered. The size of the bone being stabilized often dictates the size of the implant. Typically, the humerus is treated with a large fragment implant, while the radius and ulna often benefit from a small fragment construct [2, 10]. While some authors question the amount of fixation needed for simple transverse

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upper extremity fractures, most surgeons opt for six cortices of appropriately sized screw fixation on both sides of the fracture [11].

Diaphyseal Compression Plating: Patient Example and Surgical Technique

A 28-year-old right-hand-dominant female was involved in a motor vehicle collision sustaining a diaphyseal injury of the radius (Fig. 13.1). A standard volar approach to the forearm was performed [12]. After surgical exposure and reduction with pointed reduction clamps, a seven-hole limited contact dynamic compression plate (LC-DCP) (DePuy Synthes, Raynham, USA) was selected and centered on the fracture. The plate was secured to the proximal segment with the first screw placed in the neutral position followed by placement of the screw adjacent to the fracture in the distal segment in the eccentric or compression mode. Due to the plate and screw design, as the screw engaged the plate in the eccentric position, it contacted the sloped contour of the plate screw hole and generated 1 mm of translation of the plate construct. Since the plate was rigidly attached to only one side of the fracture, this translation induced 1 mm of compression between the two segments [2] (Fig. 13.2). Surgeons must utilize care with this technique as placing the screw on the opposite side of the hole (toward the fracture) in this example would generate 1 mm of distraction by translating the plate screw segment away from the fracture. After the first screw was tightened, alignment and fracture reduction were verified visually and fluoroscopically. Further compres-

Fig. 13.1 Anteroposterior (**a**) and lateral (**b**) forearm radiographs demonstrating a displaced transverse fracture of the left forearm

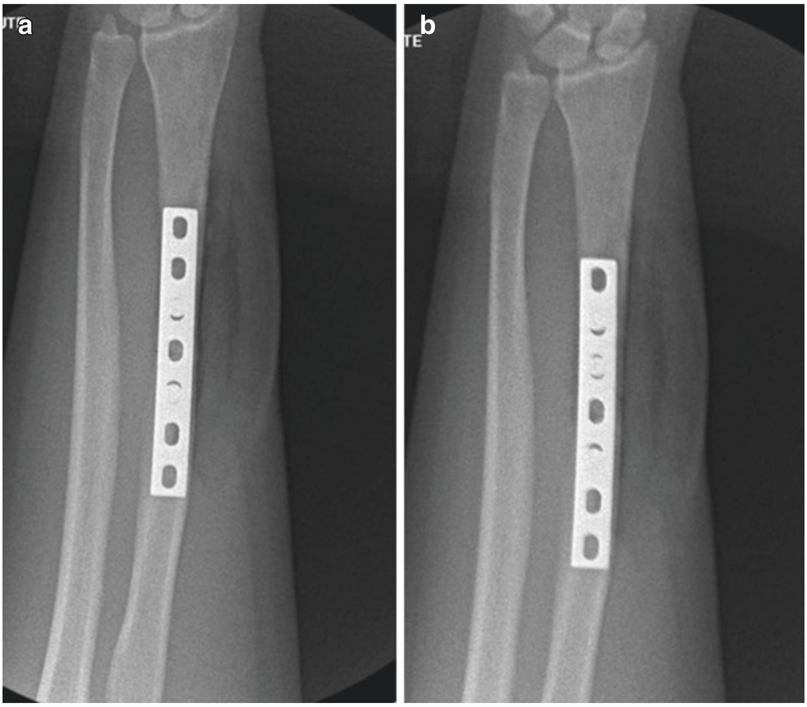

Fig. 13.2 Intraoperative anteroposterior fluoroscopic image demonstrating the placement of a seven-hole limited contact dynamic compression plate (LC-DCP) centered on the fracture. In (**a**), the plate was secured to the proximal segment first with the screw placed in a neutral position in the fifth hole from the top of the plate. Next, a screw is placed in the third hole immediately adjacent to the fracture with the screw placed eccentrically. During tightening, it induces 1 mm of compression between the two segments. In (**b**), a screw is then placed in the second plate hole eccentrically inducing an additional 1 mm of compression. The initial eccentric screw in the third hole is loosened, while the additional screw in hole 2 is being inserted to allow further compression. Once the screw in hole 2 is completely seated, the screw in hole 3 is retightened. Note the eccentric position of the screw in the third hole in (**a**) and its neutral position in (**b**), which indicates that more compression was applied with the screw in hole 2

sion was achieved by adding a second screw in the appropriate eccentric position in hole 2. As soon as this screw engaged the plate, the initial eccentric screw in hole 3 was loosened so the screw head was not engaging the hole slope, and the second screw was fully tightened generating an additional 1 mm of compression. The screw in the third hole was fully tightened to secure the reduction, and screws were placed in the remaining holes in static mode to maintain the fracture compression and alignment (Figs. 13.3 and see Fig. 13.2).

In addition to using eccentric screw placement, other methods of generating compression can be used. External clamp placement can facilitate reduction as well as provide compression. Typically for transverse fractures, this can be performed with modified pointed reduction clamps applied through drill holes on each side of the fracture. Other external methods to generate compression include screws peripheral to the plate attached to a clamp, commonly a Verbrugge clamp, or an articulating tensioning device. Recent studies have compared these different methods of compression and have showed statistically significant differences, although clinical correlation of these different amounts is not quite clear [13, 14].

While compression plating is a relatively clear and simple technique, there are several potential pitfalls. As discussed above, if the eccentrically drilled hole is performed on the

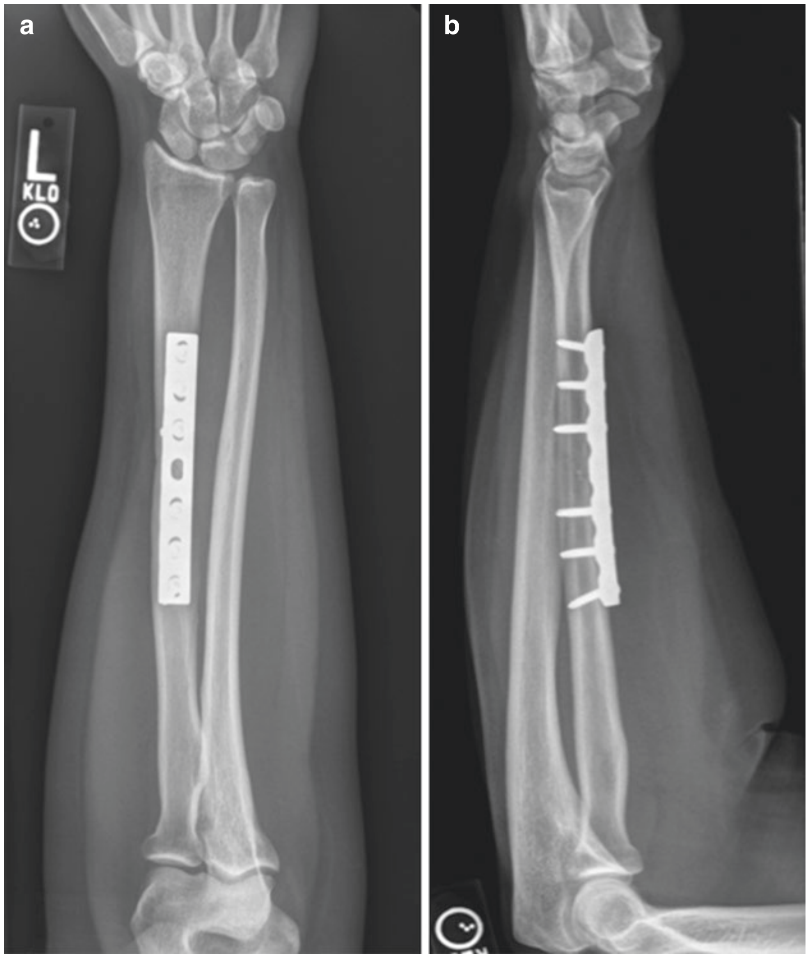

Fig. 13.3 Anteroposterior (**a**) and lateral (**b**) forearm radiographs at 3 months demonstrating maintenance of reduction and healing of the fracture without callus indication and primary bone healing

side of the hole nearest to the fracture, distraction will be generated. While most instrumentation sets have drill guides that are clearly labeled, surgeons need to understand the underlying principles to avoid this simple mistake. If it does happen, it can be recognized and corrected with screw placement into the correct location, but the risk of having two drill holes very near to each other can create a stress riser and risk losing fixation due to the cortex failing.

A second pitfall is having a plate position that is not perpendicular to the transverse fracture. If the plate is placed obliquely, the compression generated by any method will not be purely linear and can lead to unbalanced compression at the fracture site. The clinical relevance is likely related to the degree of malposition and amount of malreduction.

A third pitfall is utilizing the compression plating technique in a simple transverse fracture without having the fracture anatomically reduced. Typically, surgeons treating these fractures obtain an anatomical reduction and try to achieve absolute stability and primary bone healing with a rigid fixation construct. If a small fracture gap is present in a rigid healing environment, the possibility of nonunion increases due to the higher strain present at the fracture site [2, 15].

A fourth pitfall is not contouring the plate appropriately, which results in iatrogenic malreduction. This most commonly occurs with a

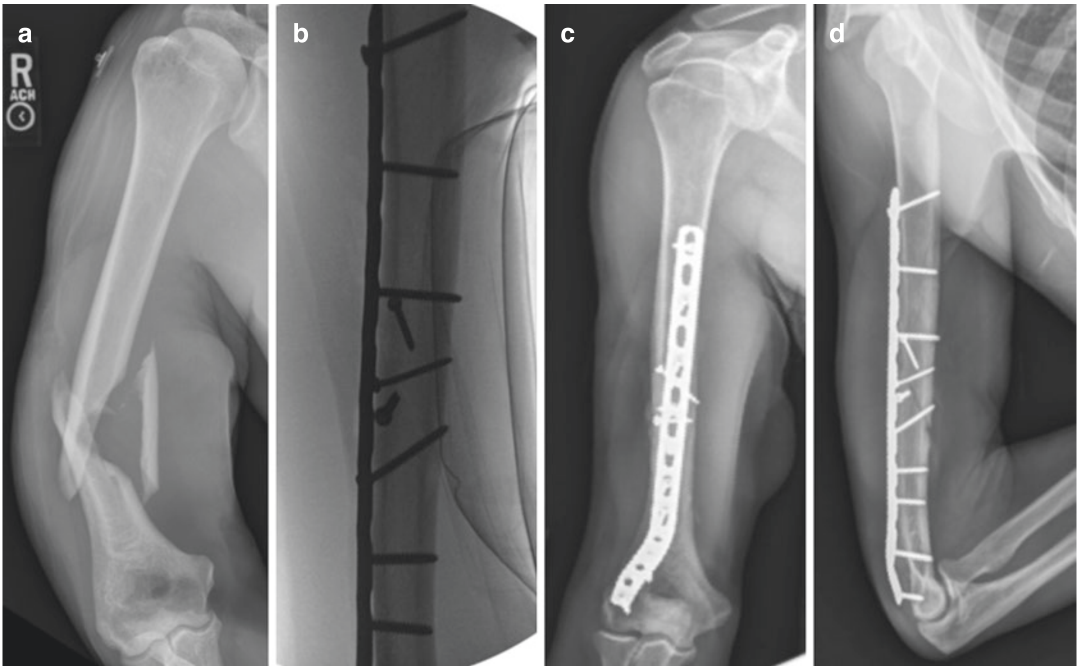

Fig. 13.4 Preoperative (**a**), intraoperative (**b**), and postoperative (**c**, **d**) anteroposterior (**a**, **c**) and lateral (**b**, **d**) images of a right distal humerus fracture with large segment of comminution. In analysis, note how the original surgeon was attempting to obtain an anatomic reduction and absolute stability with the direct exposure, clamp reduction, interfragmentary lag screw compression of the intercalary segment, and long neutralization plate. On the fluoroscopic lateral view, note how there is no sagittal plane bow of the humerus as the stiff plate was not appropriately contoured to the bone. As a result, the bone is forced to accommodate the malcontoured plate and is straight, with an anterior gap at the distal fracture site, as a result (**b**, **d**). In the 4.5-month postoperative images (**c**, **d**), note the continued presence of fracture lines as well as some callus at the fracture sites. While no implants have failed and it is not a definite nonunion, the expected healing environment is not as expected. By not having the stiff plate appropriately contoured, the fracture could not maintain anatomical reduction, and instead of healing with absolute stability and primary bone healing, there is a shift toward more of a secondary healing with callus formation

transverse fracture. Since the plate generates the compression on the same side of the bone as the plate, if the plate is not slightly over-contoured, the opposite cortex will gap as the compression is generated [2, 15] (Figs. 13.4 and 13.5). This can be problematic as noted above with a small fixed gap and rigid construct. Another way to minimize this risk is to drill both holes eccentrically on either side of the fracture with the slightly over-contoured plate and alternating which screw is tightened until both are tight.

Two potential pitfalls can occur when trying to perform compression plating on fractures with an oblique orientation. To perform compression plating in this setting correctly, the plate is stabilized to the first segment so that an acute (<90°) angle is created with the plate and fracture obliquity. This creates an axilla between the plate and the bone that prevents escape of the other segment and leads to compression as the bone is driven into this axilla. If a lag screw is to be placed during this sequence, it should be placed after the plate compression is performed. When the screw is placed first and the fracture undergoes dynamic compression with the plate, stress is placed across the screw-bone interface and can lead to failure of the lag screw compression. If the plate is first incorrectly stabilized to the segment where an obtuse angle is created between the plate and fracture obliquity, no axilla is

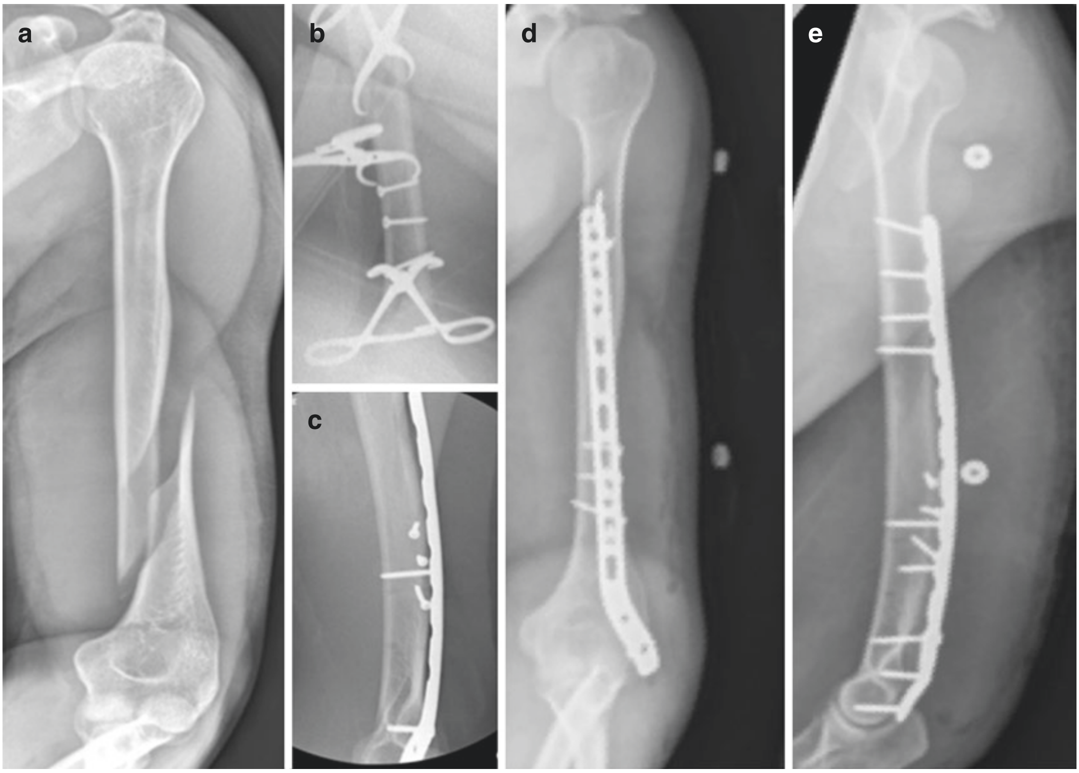

Fig. 13.5 Preoperative (**a**), intraoperative (**b**, **c**), and postoperative (**d**, **e**) anteroposterior (**a**, **b**, **d**) and lateral (**c**, **e**) images of a left spiral oblique distal humerus fracture. Image B demonstrates direct reduction and initial interfragmentary lag screw compression with countersunk 2.7-mm lag screws. Image C demonstrates appropriate sagittal plane contour of the plate reestablishing the sagittal bow of the humerus. Postoperative images (**d**, **e**) demonstrate final construct with maintained anatomical reduction and neutralization plate construct

created. When the fracture is compressed, the fracture will slide down the obliquity of the fracture inducing malalignment as there is no containment with the plate and no compression [2, 15] (Fig. 13.6).

Lastly, the correctly sized implant needs to be used. While it may be possible to use too large of an implant, typically, the error is to use an implant that is too small or not strong enough for the particular bone and anticipated healing environment. For the forearm, for example, small fragment compression plates are commonly used. However, more malleable plates, such as reconstruction plates, are not recommended due to their ability to withstand the torsional forces seen in the forearm [2]. Similarly, the humerus should be stabilized with large fragment implants unless the patient is notably small stature. Small fragment nonlocking implants do not typically have the strength to sufficiently maintain the reduction until adequate healing occurs [2, 10] (Fig. 13.7).

Neutralization/Protection Plating

The concept of neutralization or protection plating arose from treating oblique fractures that underwent ORIF with the goal of absolute stability and primary bone healing. This technique can be used for oblique diaphyseal and metadiaphyseal acute fractures and nonunions with oblique fracture morphology. A hallmark method of obtaining interfragmentary compression between two fracture fragments is a lag screw placed perpendicularly between two fracture fragments either by drilling technique or by screw design. In good bone, a properly inserted lag screw can

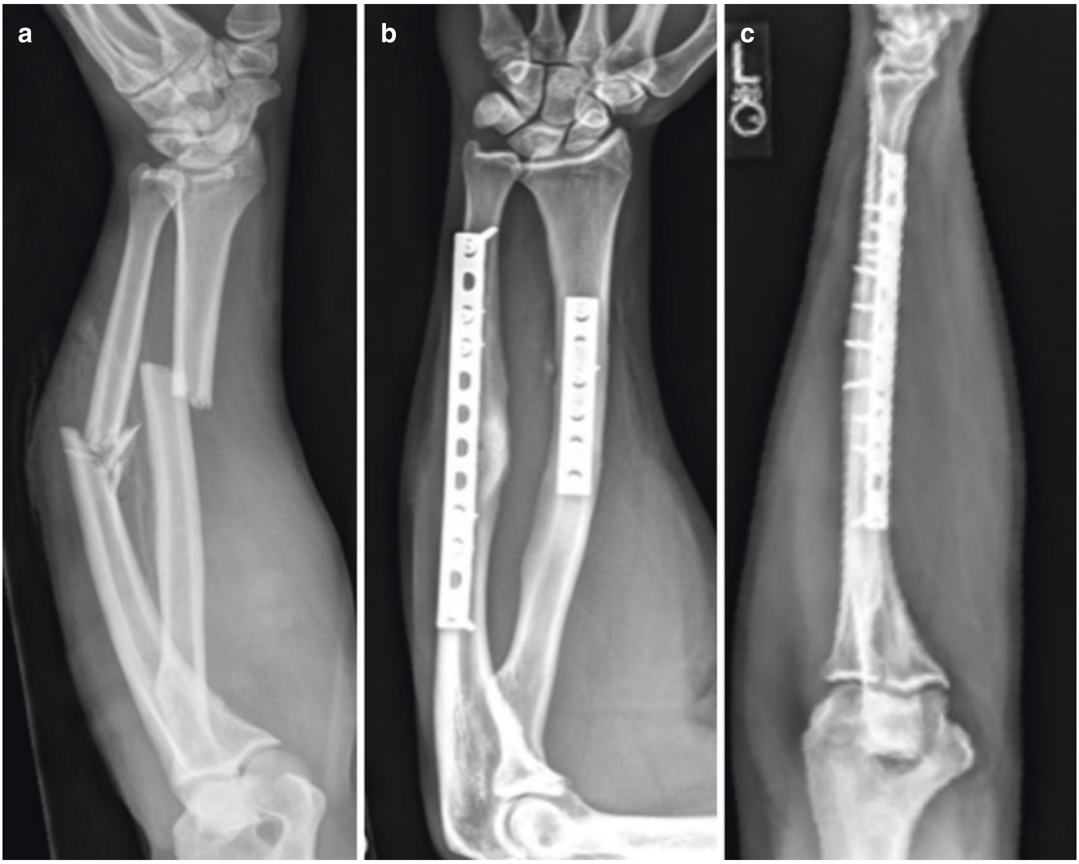

Fig. 13.6 Injury anteroposterior radiograph (**a**) demonstrating a comminuted diaphyseal ulna fracture and oblique diaphyseal radius fracture. The fracture obliquity proceeded from proximal anterior to distal posterior. The plate was first secured to the proximal segment with the hole nearest to the fracture creating an acute angle and axilla for the fracture. The next screw was the most distal screw placed in an eccentric position inducing dynamic compression into the previously created axilla securing the fracture adequately. The remaining screws were placed in neutral position. Postoperative anteroposterior (**b**) and lateral (**c**) radiographs at 4 months demonstrate primary healing of the radius with no callus and abundant callus and secondary healing of the ulna

generate compression forces up to 3000 N and should be considered when possible [2]. The lag screw provides excellent interfragmentary compression; however, it does not adequately resist the residual shear, bending, and torsional forces present. The lag screw can be placed outside of the plate, as well as through a plate hole depending on the fracture orientation, the surgical approach used, and the plate location.

Even when using a lag screw, errors in fracture reduction or malpositioned lag screws can lead to malreduction during compression leading to small residual gaps. This increased strain at the fracture site will likely be too high, increasing the risk for a nonunion and implant failure [2, 15]. Furthermore, lag screw sequence and location is an opportunity for failure. Studies have shown that screws placed through a plate are biomechanically superior, but this construct may not be possible for all fractures and surgical exposures. If the screw is placed first and outside the plate, care must be taken to ensure the screw is either adequately countersunk or sufficiently outside of the path of the neutralization plate. Having a plate sit on a proud screw head can limit the ability to maintain plate-bone contact and create suboptimal construct strength. If not avoidable, the plate must be contoured appropriately to accommodate

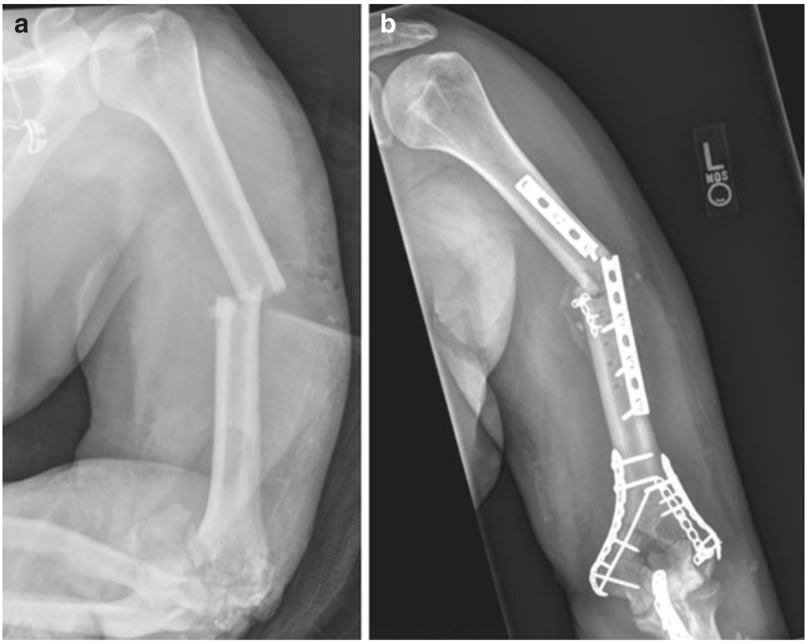

Fig. 13.7 Anteroposterior injury radiograph (**a**) of an adult left transverse diaphyseal and complex comminuted distal humerus fracture. Anteroposterior radiograph of the left humerus 6 weeks after surgical treatment (**b**) demonstrating plate failure through a screw hole and redisplacement of the diaphyseal humerus fracture component of the injury. Note the 3.5-mm limited contact dynamic compression plate (LC-DCP) utilized that is not strong enough to sustain the bending and torsional loads present in this larger patient's humerus despite adding an even smaller 2.0-mm supplemental T-plate. Furthermore, increased periosteal stripping was performed to place the supplemental plate, which potentially disrupted the fracture healing potential

the screw head. If a stout plate is not adequately contoured, whether over the screw head or to the bone, inserting and tightening a cortical screw will force the bone to contour to the plate. If the inappropriate contour is substantial enough, this can induce iatrogenic deformity with loss of reduction or fracture comminution around the lag screw. Sometimes in good bone, the bone can yield enough to accommodate the inappropriate contour leading to alignment change with a change in expected healing (see Fig. 13.4).

Neutralization Plating Patient Examples

A 32-year-old right-hand-dominant male sustained an oblique fracture to his left distal humerus during a sky diving landing (see Fig. 13.5). He was neurologically intact before and after splint reduction. After full discussion of treatment options, he opted for operative fixation. He was placed into a left lateral decubitus position, and the fracture was exposed through a posterior (triceps sparing) approach [16, 17]. The fracture was reduced with several pointed reduction clamps. Once reduction was verified, initial stabilization was provided with three countersunk 2.7-mm lag screws. These were placed at different angles along the varying obliquity of the fracture. Once all three were successfully placed, an appropriate length extra-articular distal humerus locking compression plate (LCP) was contoured appropriately in the sagittal plane and then applied with three bicortical non-locking screws proximally and three non-locking and one unicortical locking screw distally. The single locking screw was needed in the lateral distal humerus as any bicortical implant would be intra-articular in the radiocapitellar joint. The fracture went on to uneventful healing by 3 months (see Fig. 13.5).

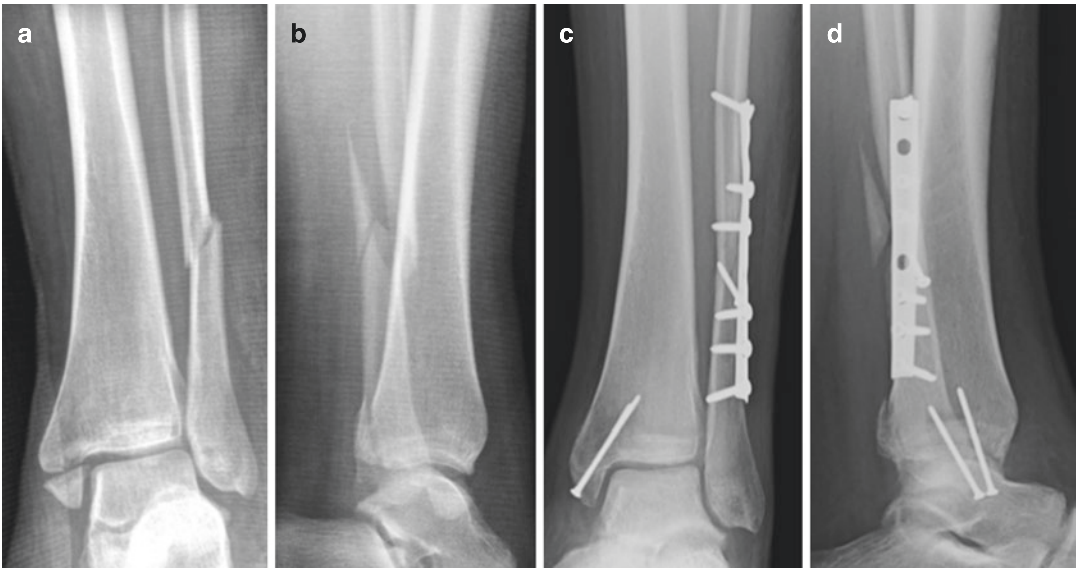

Fig. 13.8 Injury anteroposterior and lateral (**a**, **b**) and postoperative (**c**, **d**) radiographs of the right ankle demonstrating a supination external rotation injury with the cranial/posterior to caudal/anterior orientation of the fibular fracture line. There is fragmentation of the proximal aspect of the fibula fracture that was not amenable to reconstruction. This underwent an open reduction and internal fixation with an independent 2.4-mm lag screw between the main proximal and distal fracture fragments and then neutralization of the torsional forces with an appropriately contoured one-third tubular plate

Finally, a 27-year-old male sustained a twisting injury to his right ankle sustaining a trimalleolar ankle fracture (Fig. 13.8). The fibula fracture had fragmentation proximally that was not amenable to fixation. The fibula underwent a posterolateral incision with direct reduction and fixation with an independent 2.4-mm lag screw between the main proximal and distal segments and then neutralization of the remaining torsional forces with a lateral one-third tubular plate and 3.5-mm cortical screws.

Buttress/Antiglide Plating

Utilizing a plate in buttress or antiglide function is applicable to metaphyseal and partial intra-articular fractures with vertical instability and the tendency to displace in a cranial or caudal direction under physiological load. The terms are often used interchangeably, but there is a subtle distinction. The term buttress is applicable to fractures with the potential for caudal displacement of fracture fragment with physiologic load. The most common sites where buttress plating is used are tibial plateau fractures. On the contrary, antiglide plating is applicable to fractures with the potential for cranial displacement of the distal segment with physiologic load—i.e., lateral malleolus, posterior malleolus, medial malleolus, distal humerus, and distal femur fractures [2, 18–23]. With these fractures, reduction is obtained, and temporary fixation is applied. To negate the shear forces, a plate is applied at the apex of the fracture to prevent axial displacement. This biomechanically favorable construct creates an axilla where the unstable fragment is contained and diminishes the ability for any displacement. Many plates can serve this function—small fragment, anatomically precontoured, etc. Most times, standard small fragment or even more malleable plates, such as reconstruction or tubular plates, can be appropriately contoured and can function in either fashion [2]. Not infrequently, even precontoured plates will need modification to ensure the correct interface between the plate and the bone. As with any plate, ensuring the ideal contour is important to avoid inducting malreduction with plate application.

With rigid plates, the plate is positioned, and the screw closest to, but opposite, the apex of the fracture is inserted first. This initiates the plate-bone interaction and ensures that the axilla is created. Doing so diminishes the potential path of any future displacement. One subtle tip when inserting this screw is to place it eccentric in the side of the hole closest to the fracture as noted previously in the compression plating discussion. By using a universal drill/soft tissue guide, the surgeon can place the screw eccentrically in the hole abutting the side of the plate. This technique does not shift the plate during insertion, and it places the screw as close to implant as possible. If for some reason the distal fixation in the plate including the apex screw loosens and the head disengages, the shaft of the screw is closer to the plate, and the potential displacement of the plate is limited by that position in comparison to a central screw location. Manufactured soft tissue drill guides can assist with this to place this screw in "buttress mode" [2].

Two errors that occur are underappreciation of the full extent of the fracture including fracture comminution/orientation and potential for vertical instability. If the surgeon does not recognize the correct exit point of the fracture and does not place the plate at the apex of the fracture, the ability of the plate to resist the shear forces is suboptimal due to poor containment. Likewise, the surgeons may either not be able to buttress the compression side of the fracture due to anatomical or soft tissue-related reasons or choose not to do it. This is commonly seen with tibial plateau and distal femur fractures when surgeons opt for stabilization of a medial-sided injury with only a laterally based implant—commonly a lateral locking plate on the tension side of the fracture (Fig. 13.9). This

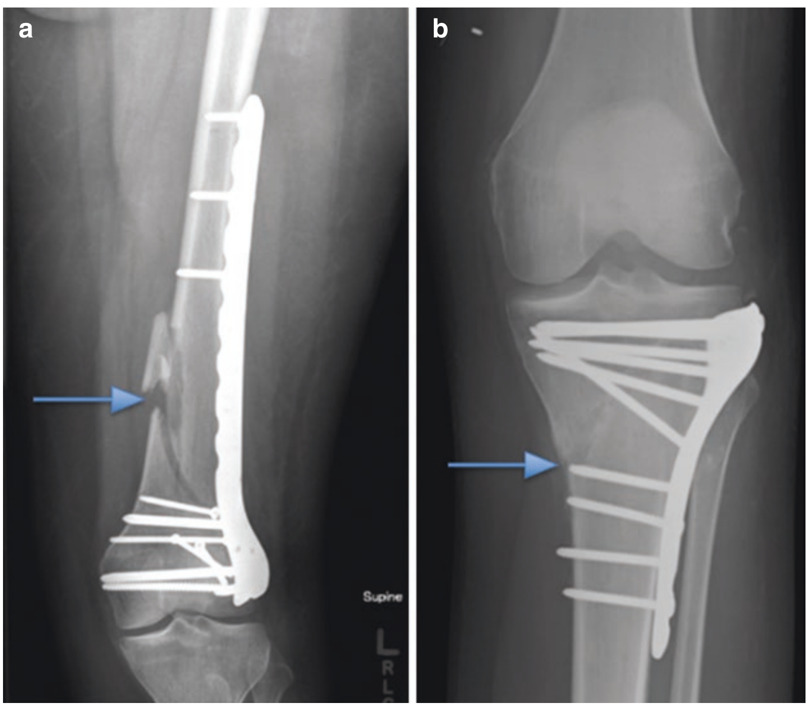

Fig. 13.9 Anteroposterior radiograph of left femur (**a**) demonstrating an intra-articular distal femur fracture with diaphyseal extension and a medial sided fracture apex that was stabilized with a lateral distal femoral locking plate. AP radiograph of the left knee (**b**) demonstrates a medial tibial plateau fracture stabilized with a lateral locking plate. For both fractures, a more optimal biomechanical construct would include reduction and fixation with a buttress plate positioned at the apex of the fracture with the first point of fixation near the apex of each fracture (*arrows*). Note that both fractures were in young patients and locking implants were not indicated

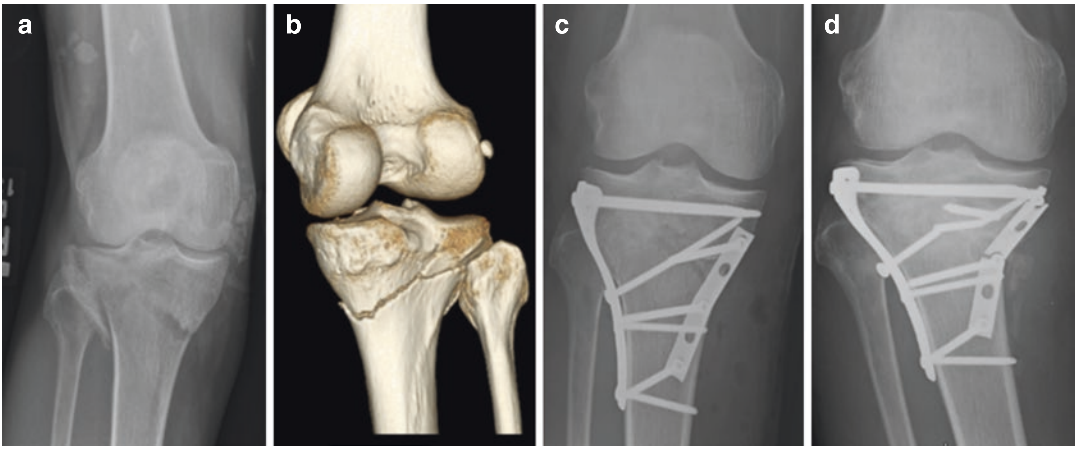

Fig. 13.10 Injury anteroposterior radiographs (**a**) and (**b**) posterior oblique CT reconstruction of the right knee demonstrating a bicondylar variant tibial plateau fracture. The fracture was stabilized with a one-third tubular plate medially at the apex of the fracture (buttress) and a precontoured lateral proximal tibia plate (**c**). Articular congruency and coronal alignment were restored. Although this patient was not obese, he admitted to walking on his leg after a few weeks from surgery. AP radiograph at 8 weeks (**d**) demonstrates fatigue failure of the medial implant with resultant varus alignment. A stronger medial construct could have potentially sustained the forces experienced during the early weight-bearing

creates a significant risk when comminution is present on the compression side, or fracture configuration leads to limited fixation by the laterally based implant. In this case example, only the distal part of the screws engaged the medial side of the fracture, leading to inadequate fixation and stability, and the potential for loss of reduction and malunion/nonunion was high. Specific to tibial plateau fractures with a posteromedial fragment, laterally based locking plates applied in standard fashion do not adequately capture the fragment because of the designed trajectory of the screws [23]. Surgeons should opt for reduction and application of an appropriate posteromedial buttress plate for these fractures instead of hoping for the success by using a biomechanically inferior construct.

Another error is using a plate that is not strong enough to withstand the forces that will occur during fracture healing. Similar to the above compression plating discussion, the size of the bone and fracture morphology often dictates implant size. For many ankle fractures, small fragment, one-third tubular, one-quarter tubular, and reconstruction plates can be utilized. Other factors such as bone quality, body habitus, and anticipated patient compliance should also be considered. For larger or elderly patients who may neither be able nor choose to be compliant with weight-bearing restrictions, or patients with prolonged healing times, a larger and stronger construct should be considered (Fig. 13.10).

Buttress/Antiglide Plate Function Patient Examples

A 37-year-old male involved in a motorcycle collision sustained a right bicondylar tibial plateau fractures with soft tissue injury indicating delayed definitive management and initial temporary knee-spanning external fixation until his soft tissue injury improved. He ultimately underwent a direct posteromedial approach for ORIF of the posteromedial fragment. A precontoured posteromedial plate was placed at the apex of the fracture functioning in buttress mode to prevent caudal and posterior displacement of the posteromedial fragment. Once secured, an anterolateral approach was performed to reduce and stabilize the lateral condylar segment with a precontoured anterolateral proximal tibia plate also functioning

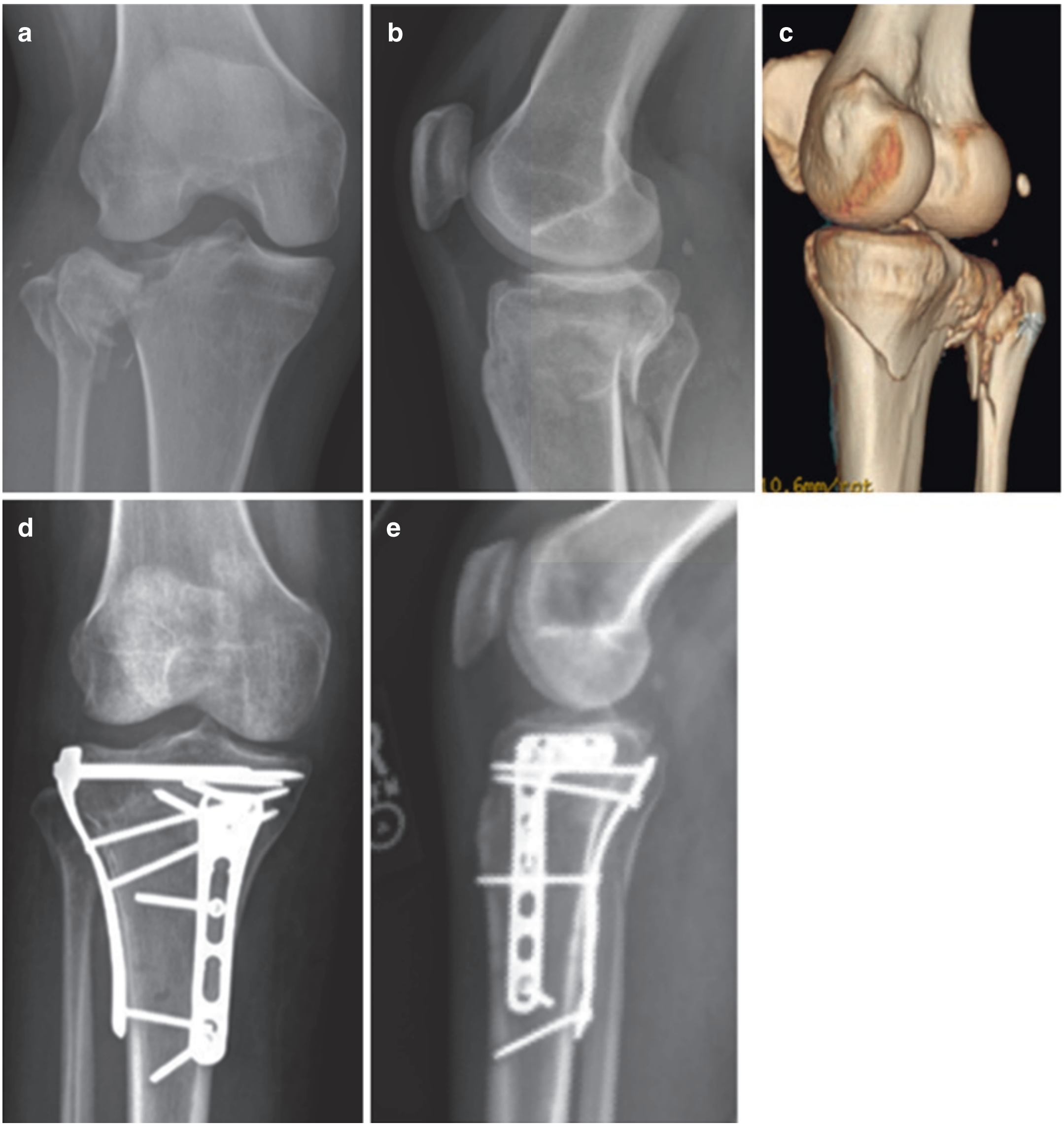

Fig. 13.11 Injury anteroposterior (**a**) and lateral (**b**) radiographs and posteromedial oblique 3D surface rendered CT image (**c**) demonstrating a right bicondylar tibial plateau fracture. Postoperative anteroposterior (**d**) and lateral (**e**) radiographs demonstrating reduction and fixation with buttress plates for both fractures. Note that although the posteromedial proximal tibial plate (DePuy Synthes) has holes with locking capabilities, only non-locking screws are utilized

in buttress mode to prevent caudal displacement (Fig. 13.11).

A 22-year-old female involved in a motor vehicle collision sustained a left supination-adduction ankle fracture. She underwent a direct medial surgical approach for open direct reduction of the medial malleolus. Due to the vertical orientation of the fracture, a one-quarter tubular plate with 2.7-mm screws was placed in antiglide mode to neutralize the shear forces and maintain the anatomical reduction. The tension failure of the fibula was stabilized with a retrograde 3.5-mm medullary cortical screw (Fig. 13.12).

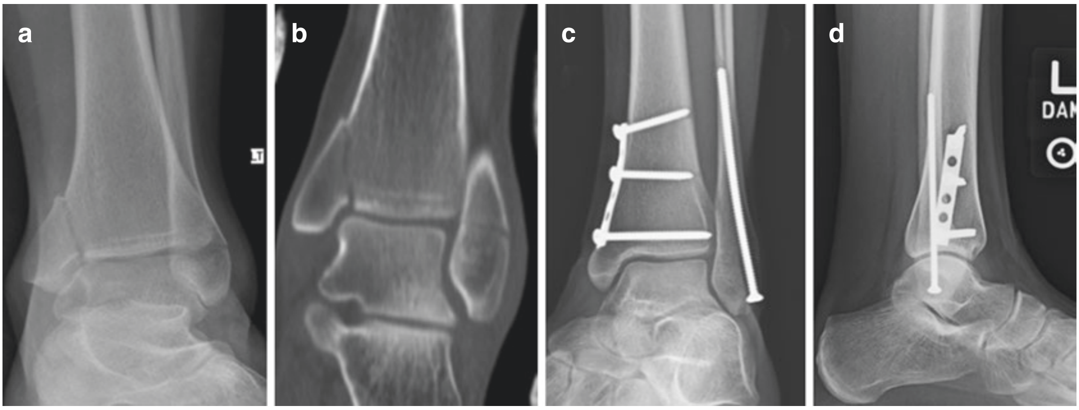

Fig. 13.12 Anteroposterior radiograph (**a**) and coronal CT (**b**) reconstruction demonstrating a left supination-adduction ankle injury with no medial marginal impaction. Postoperative AP (**c**) and lateral (**d**) radiographs demonstrate anatomical reduction and fixation utilizing a one-quarter tubular plate and 2.7-mm screws. The medial plate is functioning in antiglide mode with the second screw in the plate placed at the apex of the fracture

A 28-year-old male involved in a motorcycle collision sustained an open complex intra-articular distal femur fracture with a large medial condylar segment and lateral femoral condyle coronal plane shear fragment. This was approached with working through the open medial wounds as well as a lateral parapatellar arthrotomy for reduction and fixation. The large medial condylar block with diaphyseal extension was reduced and stabilized with 4.5-mm shaft screws for interfragmentary compression and a 3.5-mm compression plate placed directly at the apex of the fracture. In this position, the plate is functioning in antiglide mode to prevent cranial displacement of the diaphyseal fracture (Fig. 13.13).

Tension Band Plating

The main principle of tension band plating is stabilizing a fracture by placing an implant on the tension side of bones that undergo eccentric loading. The common sites where nonlocking tension band plating is performed are as follows: the lateral side of the proximal or diaphyseal femur, the anterior surface of the patella, and the posterior surface of the olecranon. The proximal humerus is another example although this commonly utilizes locking technology for proximal fixation. The forces present on the tensile side are stabilized by the implant and convert them into compressive forces on the opposite surface with physiological load from the attached muscles/ligaments—i.e., extensor mechanism of the elbow.

Successful tension band plating has certain requirements. First, the compression side of the fracture has to be compressible (simple) without comminution or at least has reconstructible comminution. If the compression side of the fracture (i.e., the olecranon articular surface) is not reconstructed, collapse will occur through the comminution during physiological load leading to potential fixation failure, malunion, or nonunion [2]. A plate and screw construct can still be used in these fractures; however, the plate would function as a bridge plate as described in the next chapter [2, 24–26]. The second requirement is that the plate has to be on the tension side of the bone and be of sufficient strength to withstand the tensile forces applied. If the plate is placed on the compressive side of the bone, it cannot resist the distractive forces. As discussed above, numerous plates can function as a tension band plate and must have the appropriate size to be able to withstand the forces of the specific anatomical location. For the femur, large fragment implants such as 4.5-mm compression and fixed angle plates can all serve as a tension band plate.

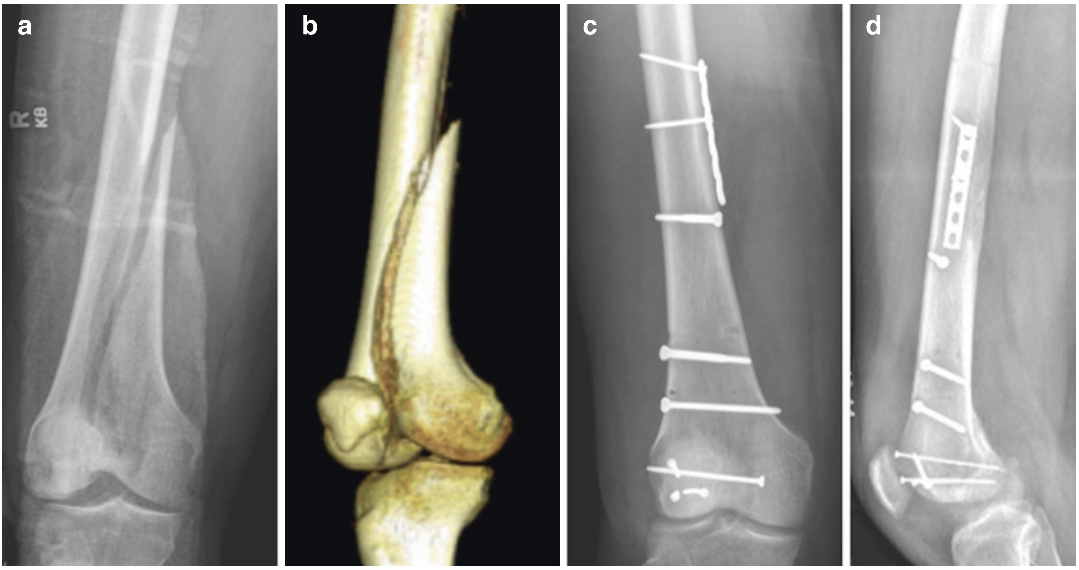

Fig. 13.13 Anteroposterior (**a**) and oblique (**b**) 3D CT reconstruction image demonstrating an intra-articular distal femur with large medial segment with diaphyseal extension. Postoperative anteroposterior (**c**) and lateral (**d**) radiographs demonstrating reduction and fixation with independent 3.5-mm lag screws, 4.5-mm shaft screws, and a medial 3.5-mm limited contact dynamic compression plate (LC-DCP) (DePuy Synthes) placed at the apex of the fracture functioning in buttress mode. Note the plate is functioning in pure buttress mode as it only has screws proximal to the fracture and no screws through the plate into the distal segment

Depending on patient size, 3.5-mm, 2.7-mm, and sometimes 2.4-mm plates can function as a tension band plate [2, 24–26].

Tension Band Plating Patient Examples

A 33-year-old male was involved in a skydiving accident sustaining a right subtrochanteric/intertrochanteric femur fracture in addition to a right associated both column acetabular fracture. This was approached through a lateral subvastus exposure with clamp reduction and initial Kirschner wire fixation to restore the proximal femur and recreate the medial (compressive) side of the proximal femur. Once adequate temporary fixation was placed, a 95° angled blade plate (DePuy Synthes) was applied. With the medial cortex restored with the reduction, the plate is on the lateral tensile cortex and will convert the tensile stresses into compressive forces. With the fracture reduced, several other implants could have been used to stabilize the fracture. A proximal femoral locking plate or proximal femur hook plate could also have been used in similar fashion but would necessitate locking screws for proximal segment fixation. A sliding hip screw implant would not be optimal as a high failure rate has been demonstrated with subtrochanteric fractures. A reconstruction or cephalomedullary nail could have also been used but was not used in this particular patient. With the large intertrochanteric fracture line, ensuring no iatrogenic fracture displacement with either type of nail insertion can be challenging, and plate stabilization was desired (Fig. 13.14).

A 32-year-old female was involved in a motor vehicle collision sustaining an open proximal transverse patella fracture. This was approached by extension of the anterior traumatic wounds to expose the anterior surface and fracture. After appropriate irrigation and debridement of the fracture, pointed clamp reduction and Kirschner wire temporary fixation were performed. The final tension band plate construct consisted of two one-quarter tubular plates and 2.7-mm

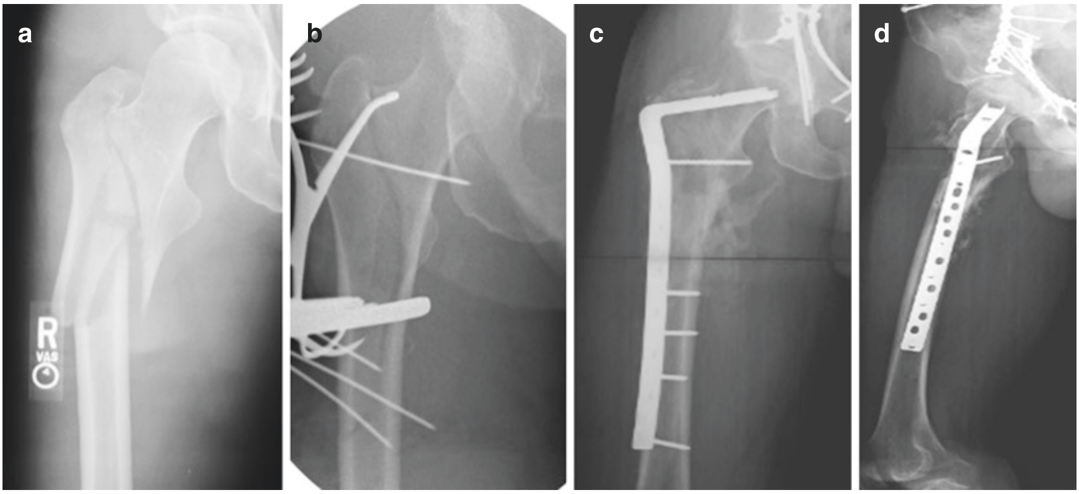

Fig. 13.14 Injury anteroposterior (**a**) radiograph of the right femur demonstrating a right subtrochanteric femur fracture with intertrochanteric extension. Intraoperative fluoroscopic (**b**) image demonstrating reduction and restoration of the medial column (compression side). The fracture was stabilized with a 95°angled blade plate (DePuy Synthes) that functions as a tension band plate. Postoperative anteroposterior (**c**) and lateral (**d**) image demonstrating healed fracture with maintenance of alignment

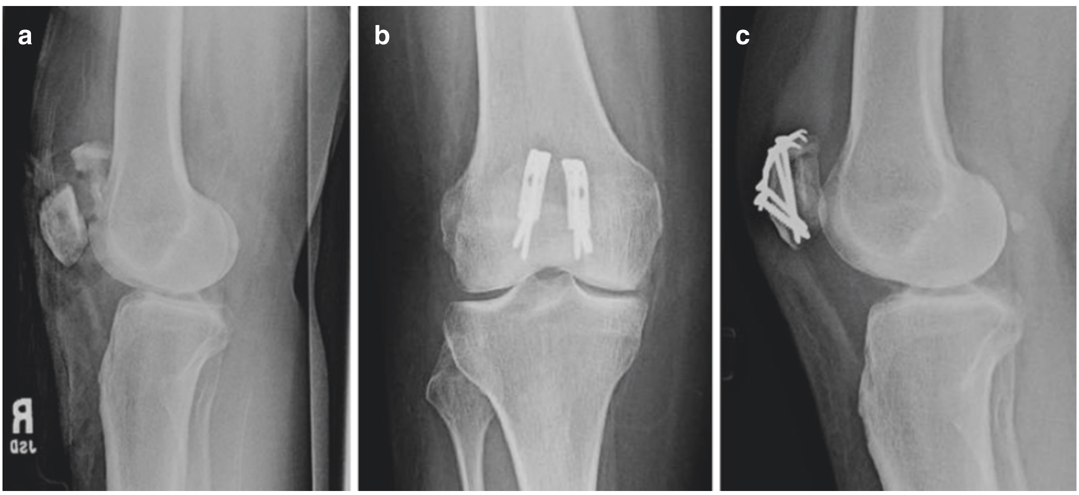

Fig. 13.15 Injury lateral radiograph (**a**) of the right knee demonstrating a displaced superior patella fracture. Postoperative anteroposterior (**b**) and lateral (**c**) images demonstrating reduction and application of two one-quarter tubular plates with 2.7-mm screws. The plating construct is functioning in tension band plate mode as it is on the anterior (tension) surface of the patella and will convert the distractive tensile forces from the extensor mechanism into compressive forces. Note the proximal fixation of the plate with a hole being cut and fashioned into a hook to maximize fixation in addition to the two proximal 2.7-mm screws placed from proximal to distal in the plate

screws. Due to the small nature of the proximal segment, the proximal hole of each plate was cut and fashioned into a hook to achieve an additional fixation. As discussed above, with the plate on the anterior tensile surface, the distractive stress will be converted into compressive forces when the knee extensor mechanism is activated (Fig. 13.15).

In conclusion, surgical stabilization of many fractures can be performed with plates using nonlocking screws. Some manufacturers are starting to only manufacture locking implants with the ability to use nonlocking screws for compression. Therefore, while there are numerous types, sizes, shapes, thicknesses, noncontoured and precontoured, locking and nonlocking plates available, the function any plate serves is dictated by where and how the surgeon applies it. The four main modes a plate can function that were discussed in this chapter were compression, neutralization, buttress, and tension band. Each function has been described and illustrated with patient examples of both appropriate and inappropriate application. Even in the era of locked plating and expanding indications of intramedullary nailing, a thorough understanding of the fracture and the accompanying patient is needed and can lead to success with nonlocking screw constructs.

References

1. Uhthoff HK, Poitras P, Backman DS. Internal plate fixation of fractures: short history and recent developments. J Orthop Sci. 2006;11(2):118–26.
2. Buckley R, Moran CG, Apivatthakakul T. AO principles of fracture management, Principles, vol. 1. 3rd ed. New York: Thieme; 2017.
3. Anderson LD, Sisk D, Tooms RE, Park WI 3rd. Compression-plate fixation in acute diaphyseal fractures of the radius and ulna. J Bone Joint Surg Am. 1975;57(3):287–97.
4. Chapman MW, Gordon JE, Zissimos AG. Compression-plate fixation of acute fractures of the diaphyses of the radius and ulna. J Bone Joint Surg Am. 1989;71(2):159–69.
5. Hadden WA, Reschaver R, Seggl W. Results of AO plate fixation of forearm shaft fractures in adults. Injury. 1982;15:448.
6. Hertel R, Pisan M, Lambert S, Ballmer FT. Plate osteosynthesis of diaphyseal fractures of the radius and ulna. Injury. 1996;27:545–8.
7. Allende C, Vanoli F, Gentile L, Gutierrez N. Minimally invasive plate osteosynthesis in humerus nonunion after intramedullary nailing. Int Orthop. 2018;42(11):2685–9.
8. Hakeos WM, Richards JE, Obremskey WT. Plate fixation of femoral nonunions over an intramedullary nail with autogenous bone grafting. J Orthop Trauma. 2011;25(2):84–9.
9. Rupp M, Biehl C, Budak M, Thormann U, Heiss C, Alt V. Diaphyseal long bone nonunions – types, aetiology, economics, and treatment recommendations. Int Orthop. 2018;42(2):247–58.
10. Patel R, Neu CP, Curtiss S, Fyhrie DP, Yoo B. Crutch weightbearing on comminuted humeral shaft fractures: a biomechanical comparison of large versus small fragment fixation for humeral shaft fractures. J Orthop Trauma. 2011;25(5):300–5.
11. Lindvall EM, Sagi HC. Selective screw placement in forearm compression plating: results of 75 consecutive fractures stabilized with 4 cortices of screw fixation on either side of the fracture. J Orthop Trauma. 2006;20(3):157–62.
12. Nauth A, McKee MD. Open reduction and internal fixation of both-bones forearm fractures. JBJS Essent Surg Tech. 2015;5(4):e28.
13. Lucas JF, Lee MA, Eastman JG. Optimizing compression: comparing eccentric plate holes and external tensioning devices. Injury. 2016;47(7):1461–5.
14. Virkus WV, Goldberg SH, Lorenz EP. A comparison of compressive force generation by plating and intramedullary nailing techniques in a transverse diaphyseal humerus fracture model. J Trauma. 2008;65(1):103–8.
15. Perren SM. Evolution of the internal fixation of long bone fractures. The scientific basis of biological internal fixation: choosing a new balance between stability and biology. J Bone Joint Surg Br. 2002;84(8):1093–110.
16. Schildhauer TA, Nork SE, Mills WJ, Henley MB. Extensor mechanism-sparing paratricipital posterior approach to the distal humerus. J Orthop Trauma. 2003;17(5):374–8.
17. Gerwin M, Hotchkiss RN, Weiland AJ. Alternative operative exposures of the posterior aspect of the humeral diaphysis with reference to the radial nerve. J Bone Joint Surg Am. 1996;78(11):1690–5.
18. Brunner CF, Weber BG. Anti-glide plate. In: Special techniques in internal fixation. Berlin/Heidelberg: Springer; 1982. p. 115–32.
19. Wegner AM, Wolinsky PR, Robbins MA, Garcia TC, Maitra S, Amanatullah DF. Antiglide plating of vertical medial malleolus fractures provides stiffer initial fixation than bicortical or unicortical screw fixation. Clin Biomech (Bristol, Avon). 2016;31:29–32.
20. Switaj PJ, Wetzel RJ, Jain NP, Weatherford BM, Ren Y, Zhang LQ, Merk BR. Comparison of modern locked plating and antiglide plating for fixation of osteoporotic distal fibular fractures. Foot Ankle Surg. 2016;22(3):158–63.
21. Ratcliff JR, Werner FW, Green JK, Harley BJ. Medial buttress versus lateral locked plating in a cadvaver medial tibial plateau fracture model. J Orthop Trauma. 2007;21(7):444–8.
22. Patel PB, Tejqani NC. The Hoffa fracture: coronal fracture of the femoral condyle a review of literature. J Orthop. 2018;15(2):726–31.
23. Barei DP, O'Mara TJ, Taitsman LA, Denbar RP, Nork SE. Frequency and fracture morphology of the posteromedial fragment in bicondylar tibial plateau fracture patterns. J Orthop Trauma. 2008;22(3):176–82.

24. Hommel GJ, Lobrano C, Ogden AL, Mukherjee DP, Anissian L, Marymont JV. A quantitative analysis of tension band plating of the femur diaphysis. Arch Orthop Trauma Surg. 2011;131(10):1325–30.
25. Duckworth AD, Clement ND, White TO, Court-Brown CM, McQueen MM. Plate versus tension-band wire fixation for olecranon fractures: a prospective randomized trial. J Bone Joint Surg Am. 2017;99(15):1261–73.
26. Zderic I, Stoffel K, Sommer C, Höntzsch D, Gueorguiev B. Biomechanical evaluation of the tension band wiring principle. A comparison between two different techniques for transverse patella fracture fixation. Injury. 2017;48(8):1749–57.

Nonlocking Plate Functions 2

14

Elizabeth B. Gausden and Timothy S. Achor

Bridge Plating

Bridge plates are used for open reduction and internal fixation (ORIF) of comminuted metaphyseal and some diaphyseal fractures. Common situations include comminuted distal femur fractures with complex articular involvement, or more simple articular fractures where the plate can also be used as a reduction tool. A common misconception is that bridge plates are locking plates. It is important to realize that bridge plating is a function of the plate and not the specific type of plate.

Bridge plates are used to restore functional reduction of the metadiaphyseal fracture component. That is to say, restoration of length, alignment, and rotation are the goals of the bridge plate. Bridging thus provides relative stability to the fracture. The desired healing that ensues is secondary healing through callus formation.

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A bridge plate is ideally used to span comminuted fractures. Performing a direct reduction and using an absolute stability construct would devitalize soft tissues and periosteum that are vital to revascularization and fracture healing in the setting of comminution. Thus, an indirect reduction technique for non-articular fracture components and preservation of periosteum, fracture hematoma, and soft tissues is generally used in concert with bridge plating.

An understanding of stress and strain is necessary in order to determine which fractures are amenable to bridge plate fixation. Stress is force divided by area, while strain is defined as the motion between fracture fragments divided by the distance between fracture fragments [1]. Fractures unite through secondary bone healing in environments with low strain [1].

An appropriately placed bridge plate results in a flexible environment that allows for motion between comminuted bony fragments with physiological loading. In fracture comminution, there is a large overall distance between fracture fragments. A flexible construct in the setting of a fracture with a large overall distance between fragments results in a low overall strain (strain=motion between fragments/distance between fragments).

Three factors influence the stability of a bridge plate: length of the plate, the working length of the plate, and the density and design of the screws used [2].

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Length of the Plate

Since stress is defined as force/area, longer plates will have a larger stress distribution. The ideal length of a bridge plate relative to the length of the comminuted fracture being spanned is debatable. A reasonable estimation is that a bridge plate should be three times the length of the comminution it spans [1]. If a bridge plate is used in a simple fracture pattern, it should be 8–10 times the length of the fracture in order to dissipate the strain over a longer working distance [2].

Working Length of the Plate

The working length of the bridge plate is defined as the distance between the two screws closest to either side of the fracture site. The shorter the working length, the less flexible the construct is. Although not proven clinically, a shortened working length increases strain and is thought to lead to a higher risk of nonunion.

Screw Design and Density

Any plate can be used as a bridge plate, including locking plates. When locking screws are used in bridge plating, the surgeon must be mindful of their effect on strain. Locking screws increase the stiffness of a construct, thereby increasing the strain by decreasing the allowable motion at the fracture. While this is advantageous in certain situations, a very stiff construct in bridge plate situations may lead to nonunion. One study of distal femur fractures identified that when all screws proximal to the fracture were locking screws, there was a 48.8% rate of nonunion compared to 25.0% when cortical screws were used as well (hybrid screw technique) [3]. In patients with adequate bone quality, cortical screws (nonlocking) can be placed proximal and distal to the spanned comminution, while in osteoporotic bone, locked screws may be necessary if there is poor cortical screw purchase—however, this is unlikely in the diaphysis.

Deciding on the appropriate number of screws on either side of the fracture (screw density) is crucial, as too many screws can lead to an overly stiff construct, creating a high strain environment and predisposing to nonunion. A general recommendation is to use a screw density of 0.5 [2], meaning that at least half of the screw holes of the plate are left empty in a bridge plate construct. Depending on the anatomic region and bone quality in question, screws can be placed much more sparingly.

Case 1: Bridge Plate

A patient is an 18-year-old male who was the victim of multiple gunshot wounds. One of the bullets caused a right comminuted distal humeral shaft fracture (Fig. 14.1). The patient was indicated for debridement as well as osteosynthesis given the distal extent of the fracture.

A triceps splitting approach was performed and the radial nerve was protected. After excisional debridement of the multiple bullet fragments and the bony fragments that were devoid of soft tissue, the fracture was grossly reduced, and no attempt was made to reduce additional fragments of comminution to minimize further periosteal damage. A 12-hole posterolateral extra-articular distal humeral locking plate was selected and positioned deep to the radial nerve. Four proximal cortical screws were placed in addition to two distal cortical screws. Three distal locking screws were used because it was a short segment for fixation distally, and bicortical nonlocking screws would be intra-articular at this level. Additional stability was obtained by applying an orthogonal contoured 3.5-mm reconstruction plate with three proximal and two distal cortical screws (Fig. 14.2). At 9 months postoperatively, the patient returned for follow-up and the fracture was healed (Fig. 14.3).

Why This Works

Secondary to the severe comminution and periosteal damage from the energy of the injury, bridge plating was chosen as the fixation strategy. The

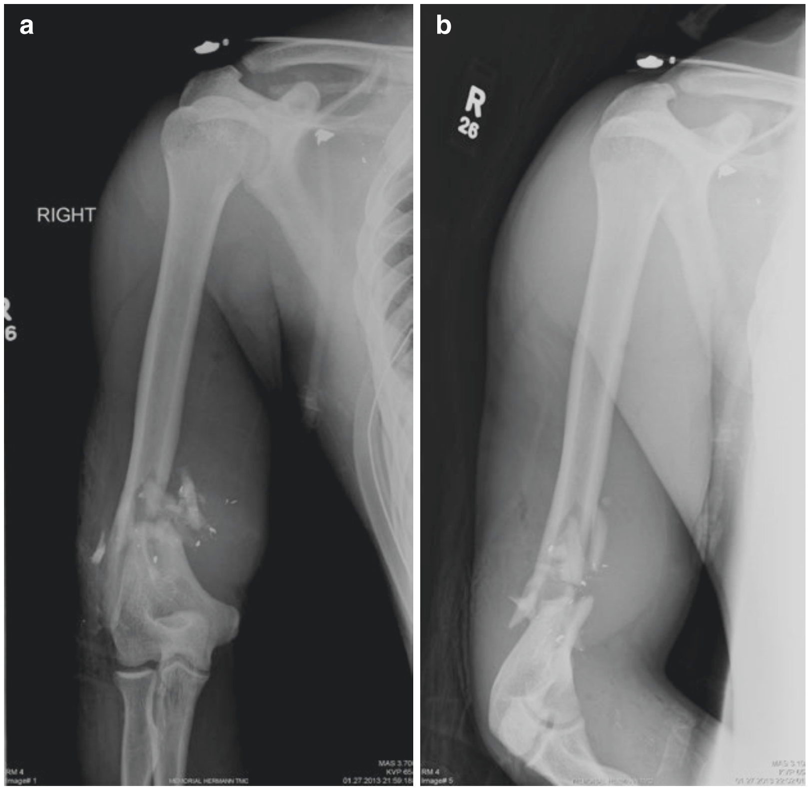

Fig. 14.1 Lateral (**a**) and anteroposterior (**b**) injury radiographs demonstrating a comminuted extra-articular distal humerus fracture

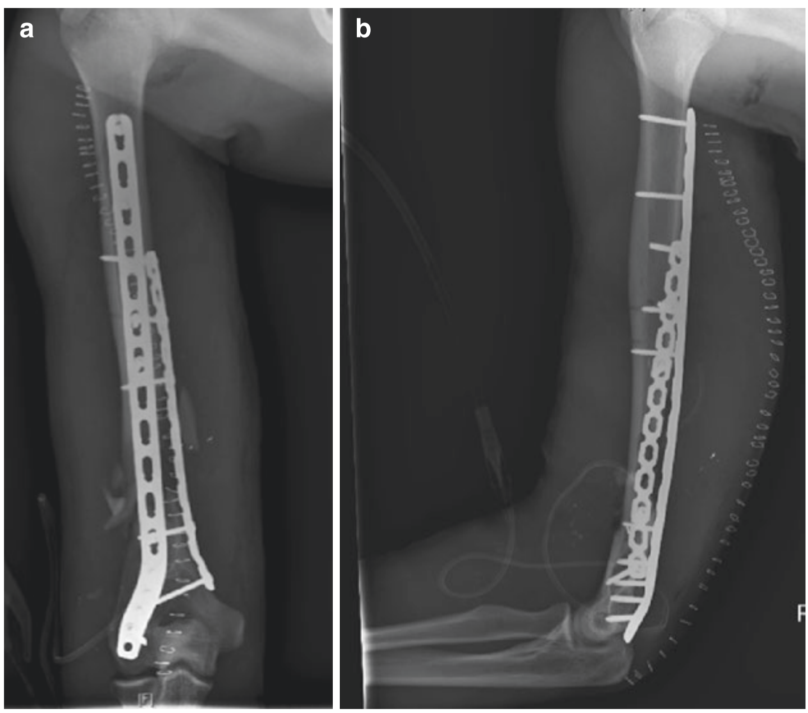

Fig. 14.2 Anteroposterior (**a**) and lateral (**b**) radiographs immediately post-fixation with bridge plating

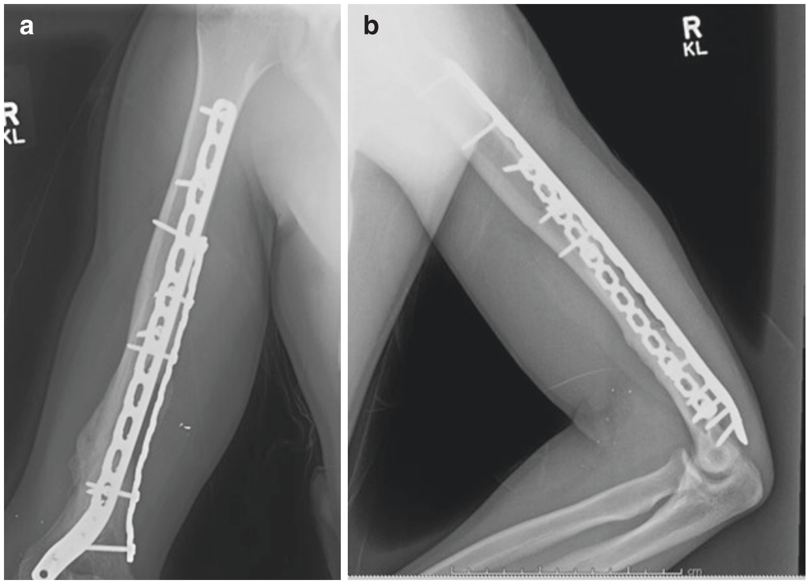

Fig. 14.3 Anteroposterior (**a**) and lateral (**b**) radiographs demonstrating secondary healing of the fracture

long posterior locking plate with both locking and cortical screws in combination with the orthogonally placed shorter reconstruction plate created a dual bridge plate construct. The orthogonal plates distributed the load over a long working distance and were strong enough to resist the torsional forces on the humeral shaft while the comminuted fracture healed.

Wave Plate

One modification of a bridge plate is the wave plate, which was originally described by Weber and Brunner in 1982 [4]. The distinctive feature of a wave plate is its central curved segment. There are three main advantages to using wave plates [5]. First, the "wave" of the plate provides improved access to the fracture or nonunion site for bone grafting. The wave also reduces the plate's contact with the bone and reduces the interruption in periosteal blood flow compared to standard plating. Finally, the wave in the plate increases the area and force distribution of the plate. As callus forms, exaggerating the convexity of the bony surface to which the wave is applied, the plate will function as a tension band, converting tension forces to compression across the fracture site. In recent literature, the wave plate has been applied successfully to long bone nonunions including the femur, humerus, ulna, and radius [6–8].

Case 2: Wave Plate

A 64-year-old polytrauma patient sustained a femoral shaft fracture in a motor vehicle collision that was initially treated at an outside institution. The patient presented 1 year following the incident with an atrophic left femoral nonunion (Fig. 14.4). After discussing treatment options, the patient elected to undergo exchange nail and augmentative wave plating. The previous nail and interlocking screws were removed, and the nail was exchanged for a larger reamed retrograde nail. Then, a lateral subvastus approach to the femur was performed. The nonunion site was identified and taken down with osteotomes and a high-speed burr. Approximately 40 cc of iliac crest bone graft was harvested. A wave plate was contoured and then applied to the nonunion site, and screws were strategically placed around the nail. The bone graft was packed underneath the wave of the plate abutting the nonunion site both anteriorly and posteriorly. At the patient's

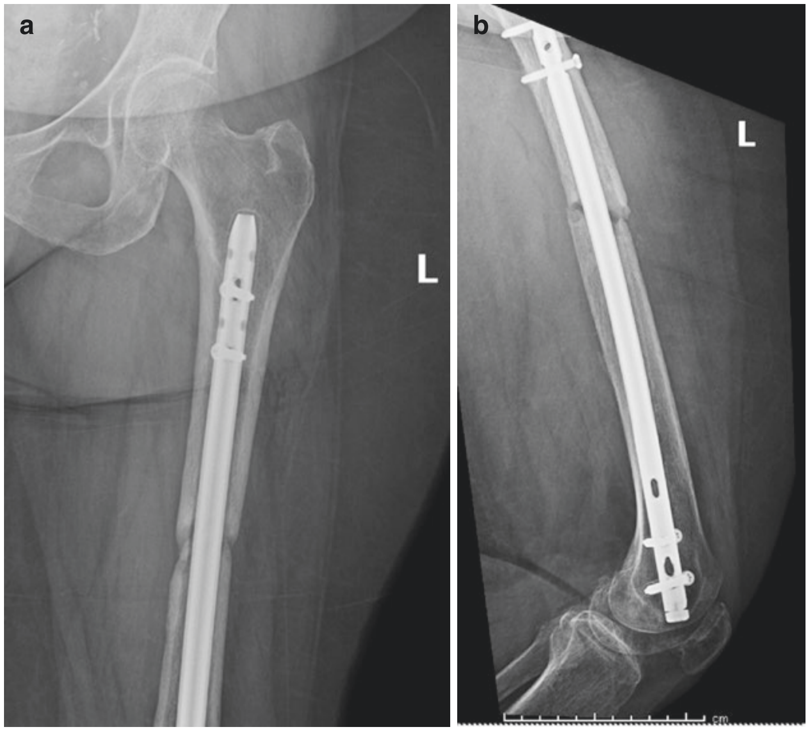

Fig. 14.4 Anteroposterior (**a**) and lateral (**b**) radiographs demonstrating an atrophic nonunion of the femoral shaft

10-month follow-up appointment, the nonunion had healed and the patient's symptoms resolved (Fig. 14.5).

Why This Works

In this case, the subtrochanteric region had already failed to heal after one attempt at intramedullary nailing. The wave plate with bone grafting was chosen as a supplement to the larger exchange nail in order to add stability to the construct as well as to capture the bone graft.

Fixed Angle Devices

A fixed angle device is any device that has a fixed angle within the implant, including blade plates, dynamic hip plates/sliding hip screws, dynamic condylar plates, and locking plates. The blade plate and the dynamic hip/condylar plate are two unique nonlocking plates that warrant additional discussion.

The blade plate was the first used fixed angle plate, and it was introduced in the 1960s [9]. The blade plate can function as a tension band, a compression plate, or a bridge plate depending on its application. The blade plate is an L-shaped plate that is fashioned from a single piece of stainless steel. The most commonly used plate has an angle of 95 degrees, but additional plates come in 110, 120, or 130 degrees. Although largely replaced by locked plating, the blade plate continues to be used in select acute fractures [10], proximal femoral osteotomies, nonunions [11], and for salvage arthrodesis [12]. The original indications of the blade plate were to treat proximal and distal femur fractures.

There is evidence of failure of proximal femoral locking plates (PFLP) that has renewed interest in blade plating [13–15]. While the PFLP are stronger biomechanically compared to blade plating [16], they have not performed as well clinically. After blade plate insertion, the articulated tensioning device (ATD) can be used to compress or "load" the fracture and perhaps further correct deformity, prior to placing shaft

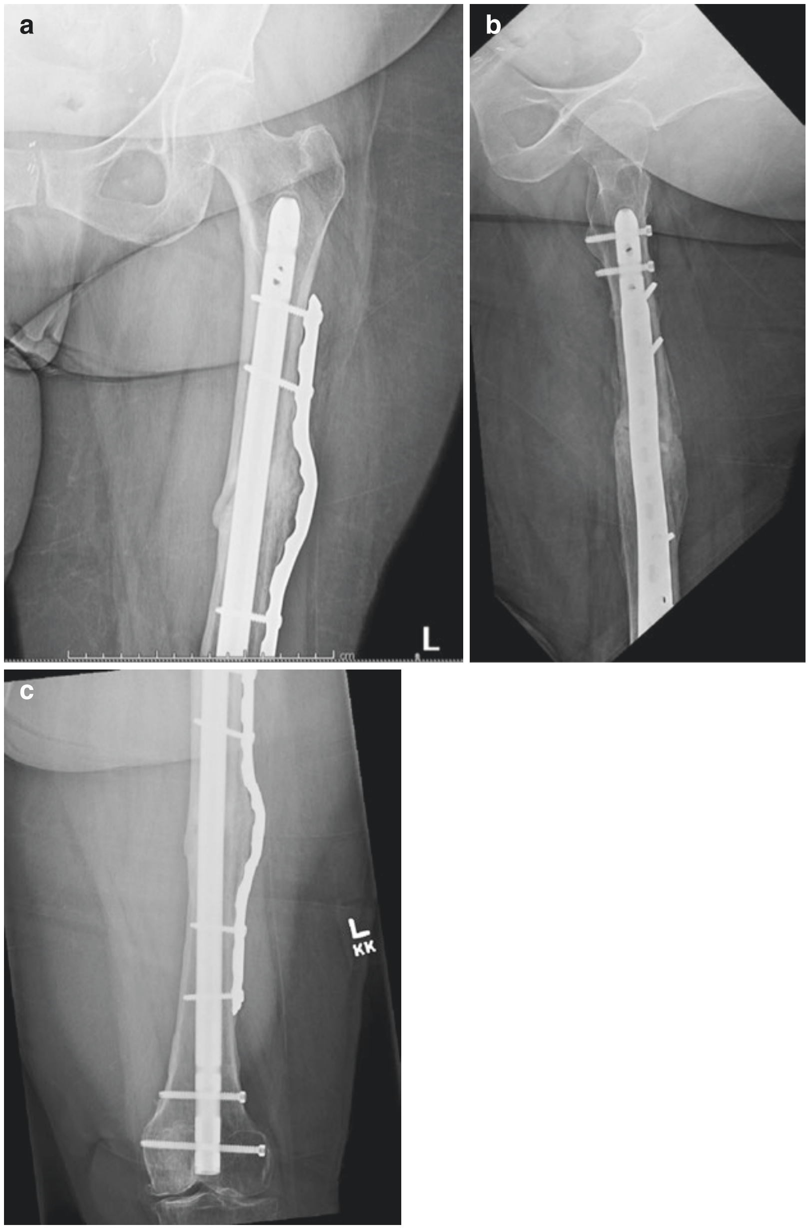

Fig. 14.5 Postoperative anteroposterior (**a**) and lateral images of the hip (**b**) and anteroposterior image of the distal femur (**c**), demonstrating healing of the femoral shaft fracture following exchange nailing with wave plating

screws. The use of the ATD and its ability to load the blade plate is one of the major advantages, and for this reason, it is particularly beneficial in fractures that are amenable to compression [10].

However, the blade plate is technically challenging because precise placement is required. Once the blade has been impacted, altering plate position will alter the reduction. Therefore, this implant requires appropriate preoperative planning and correct positioning in multiple planes and cannot be inserted in a percutaneous manner.

Case 3

A 57-year-old male was struck by a projectile from a wood chipper and sustained a complex open distal femur fracture (Fig. 14.6). There was

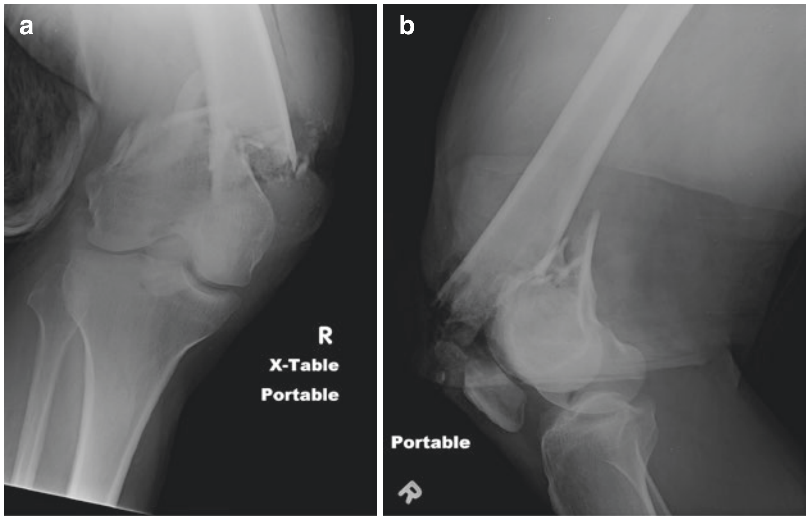

Fig. 14.6 Anteroposterior (**a**) and lateral (**b**) radiographs of the distal femur fracture

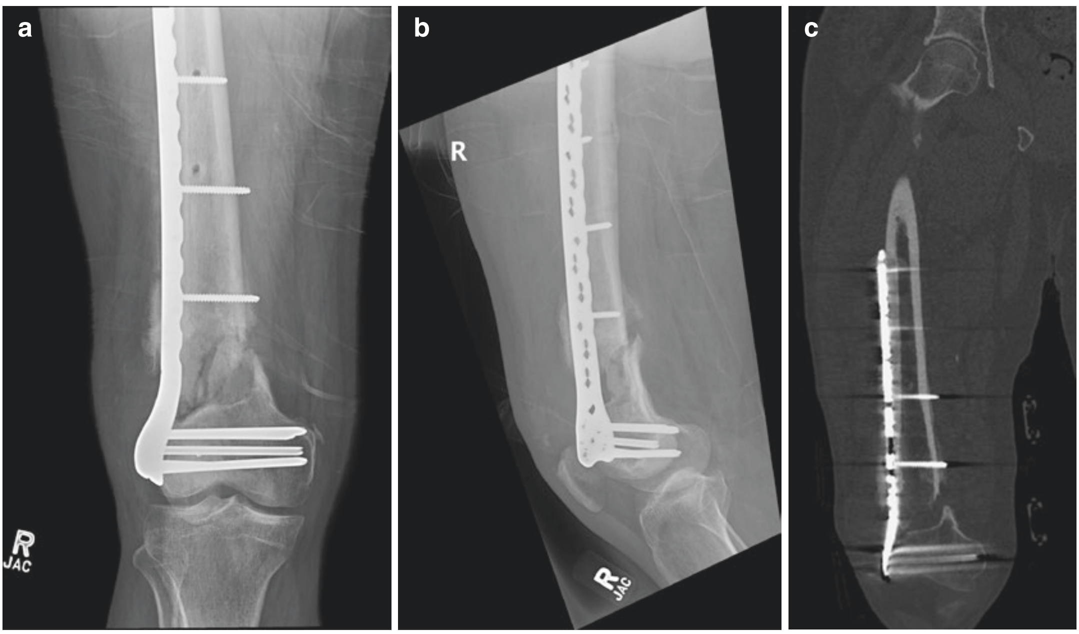

Fig. 14.7 The patient went on to nonunion of the distal femur as demonstrated on anteroposterior (**a**) and lateral (**b**) radiographs and confirmed by CT (**c**)

a large transverse wound along the anterior/distal thigh that transected his quadriceps tendon. After initial debridement and knee-spanning external fixation, definitive ORIF was performed through an anterior incision that incorporated his open wound using a distal femoral locking plate. Eight months following the initial procedure, the patient presented with continued pain and nonunion (Fig. 14.7).

A separate lateral incision was used, and a subvastus approach to the distal femur was performed. The nonunion was debrided. Then iliac crest was harvested for bone graft. The reference wire for the blade is critical and needs to be placed parallel to the joint line on the AP view (Fig. 14.8). A 95-degree blade plate was inserted. This was secured distally with cortical screws, and then the fracture was compressed with the ATD proxi-

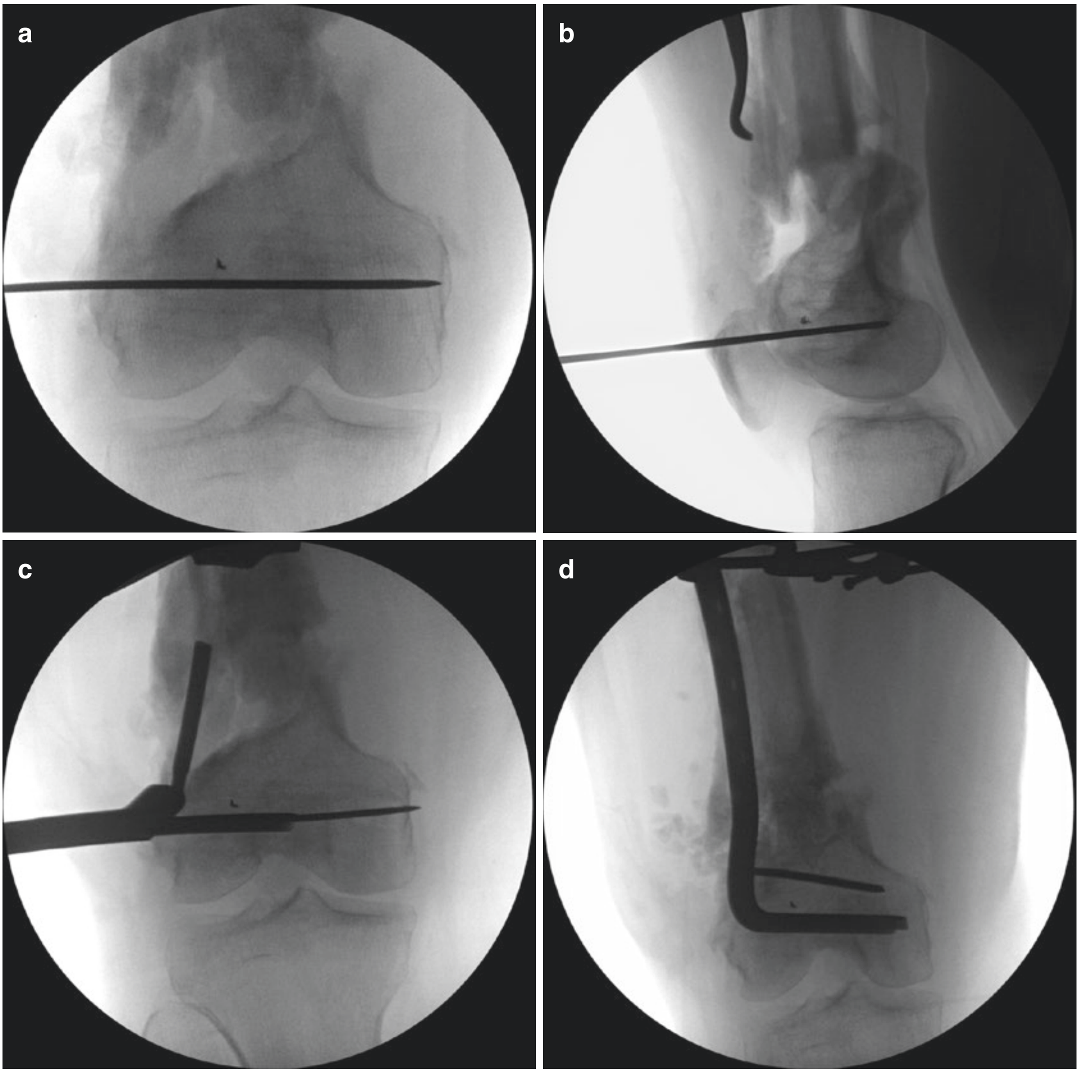

Fig. 14.8 Intraoperative images demonstrating blade plate preparation and insertion. Placement of the summation guide wire on anteroposterior (**a**) and lateral (**b**) fluoroscopy. (**c**) The chisel is introduced over the guide wire and (**d**) the blade plate is inserted

mally and then secured with additional cortical screws proximally using compression technique (Fig. 14.9). The patient returned for follow-up 9 months following nonunion repair and showed radiographic healing (Fig. 14.10).

Why This Works

This distal femoral fracture was at high risk for nonunion given the high-energy and the open nature of the injury. When the distal femoral locking plate failed, the blade plate was chosen to compress the nonunion with ATD and provide rigid, fixed angle fixation.

Similarly, the dynamic condylar screw (DCS) is a 95° fixed angle implant intended for fixation of distal femur fractures or subtrochanteric fractures that has largely been replaced by other more technically forgiving implants [17]. The DCS is traditionally considered more "forgiving" than the blade plate because once the plate is inserted, unlike the blade, it can still be adjusted in the sagittal plane to accommodate the femoral shaft. Furthermore, it can be placed percutaneously.

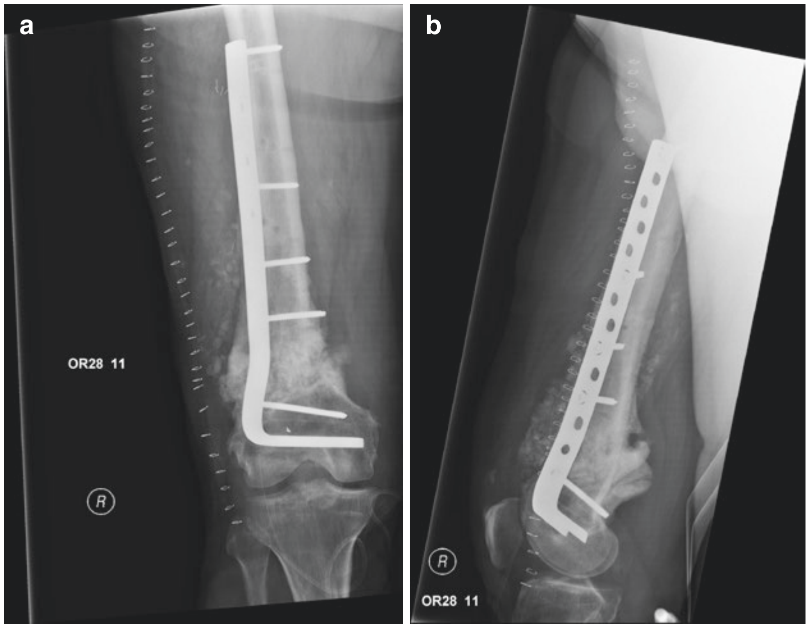

Fig. 14.9 Immediate postoperative anteroposterior (**a**) and lateral (**b**) radiographs demonstrating the final construct

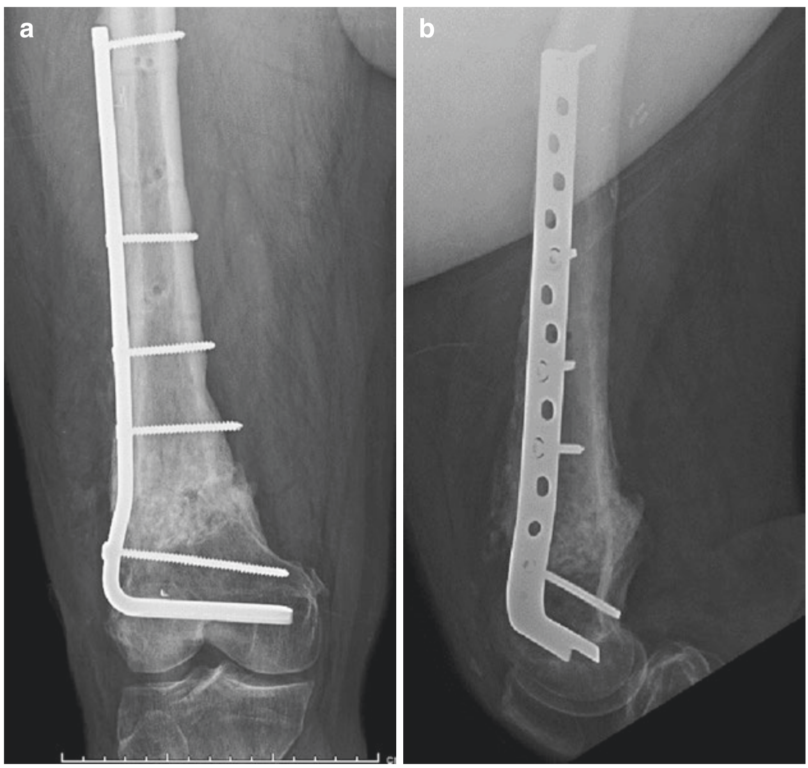

Fig. 14.10 Anteroposterior (**a**) and lateral (**b**) radiographs at 9 months following surgery

Case 4: Dynamic Condylar Screw

A 62-year-old female sustained a spiral femoral shaft fracture from a mechanical fall and subsequently underwent retrograde intramedullary nailing. At 8 weeks post-op from the retrograde nail, she was involved in a motor vehicle collision and sustained an ipsilateral unstable intertrochanteric femur fracture proximal to the previously placed nail (Fig. 14.11). A lateral approach to the hip was performed for fracture reduction. Due to the level of the retrograde nail impeding placement of a DHS, a 95° DCS implant was used to stabilize the hip fracture (Fig. 14.12).

Fig. 14.11 Intertrochanteric hip fracture proximal to a retrograde femoral nail

Why This Works

In this patient with an intertrochanteric fracture proximal to a retrograde femoral nail, preoperative templating revealed that a standard sliding hip screw would not be able to be inserted as there was interference from the nail that would block the barrel of the screw. The DCS with its 95° allowed for insertion proximal to the femoral nail. While a blade plate also could have been used, it is a more technically demanding device to insert as it cannot be rotated once inserted.

Another fixed angle plate commonly used in fracture surgery is the sliding hip screw (SHS). This is a stainless steel implant designed to treat proximal femur fractures. The device consists of a large cancellous screw that freely slides within a barrel that is attached to a side plate. The design allows for controlled collapse within a single

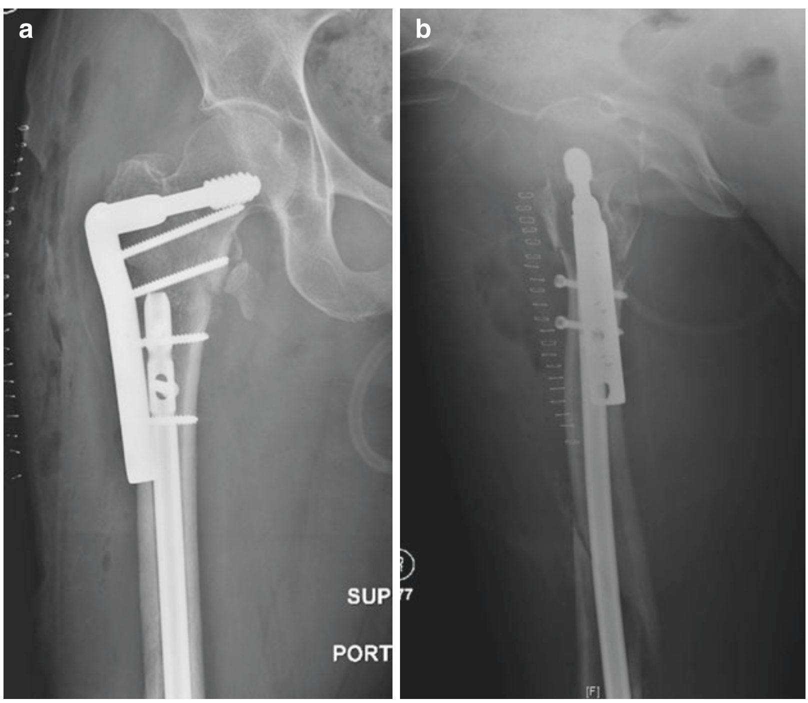

Fig. 14.12 Immediate postoperative imaging anteroposterior (**a**) and lateral (**b**) demonstrating the placement of the DCS proximal to the femoral nail

plane as the fracture heals. An anti-rotation screw can be placed prior to insertion of the SHS in order to resist rotation of the femoral head as the lag screw is inserted. However, for unstable intertrochanteric hip fractures, including reverse obliquity fractures, fractures with posteromedial comminution, subtrochanteric extension, or lateral cortex insufficiency, cephalomedullary nailing is typically chosen over SHS. These fracture patterns result in either loss of the lateral femur as a buttress or loss of resistance to medial shaft displacement. If the lateral femoral cortex is disrupted, then the sliding hip screw will fail as the telescoping along the lag screw will result in uncontrolled fracture displacement and failure.

Fig. 14.13 Widely displaced intertrochanteric hip fracture resulting from a high-energy injury

Case 5: Sliding Hip Screw

A 62-year-old male involved in a motor vehicle collision sustained an open intertrochanteric hip fracture with an 8-cm open wound (Fig. 14.13). The wound was irrigated copiously. A subvastus lateral approach to the hip was performed. The fracture was reduced using traction and direct manipulation of fragments with large pointed reduction clamps.

A sliding hip screw with an anti-rotational screw was used for fixation (Fig. 14.14).

Why This Works

The SHS, in combination with an anti-rotational screw, was successful in this high-energy intertrochanteric hip fracture as it allowed for controlled collapse along the femoral neck while maintaining fracture alignment. It should be noted that this patient had a stable, albeit high-energy, intertro-

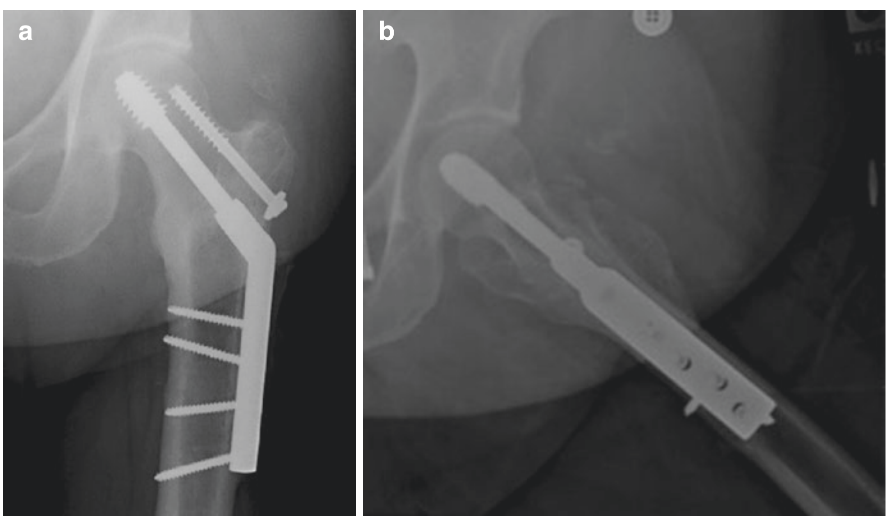

Fig. 14.14 Postoperative images including AP (**a**) and lateral (**b**) demonstrating placement of the SHS with anti-rotational screw fixation

chanteric hip fracture with an intact lateral femoral cortex that was amenable to SHS fixation. While an intramedullary device would have been another option for fixation [18], surgeons should recognize that in this high-energy patterns, closed reduction techniques are unlikely to reduce the fracture due to soft tissue injury, and open reduction will likely be necessary. Therefore, a subvastus approach was utilized to accomplish both reduction and application of fixation while providing the benefit of not violating the abductors as an intramedullary device would [3].

Conclusion

In most fractures or nonunions, surgeons have a variety of implant and techniques to choose from that can accomplish the goal of fracture healing and restoration of length, alignment, and rotation. Bridge plating can be accomplished with either locked or nonlocked plates and has the goal of providing stability while minimally disrupting soft tissues. Wave plates allow the advantage of providing access for bone grafting at fracture or nonunion sites while also providing additional stability. Fixed angle devices, including sliding hip screws, dynamic condylar screws, and blade plates, have varying degrees and ease of use, with the blade plate being the most technically challenging. They are excellent tools that orthopedic surgeons can employ and should be chosen based on the biomechanical advantages that are needed in each individual setting.

References

1. Graves M. Principles of internal fixation. In: Browner BD, Jupiter JB, Krettek C, Anderson PA, editors. Skeletal trauma: basic science, management, and reconstruction. 5th ed. Philadelphia: Elsevier; 2015. p. 221–90.
2. Gautier E. A thorough understanding of the basic concepts of fixation, the bone biology and biomechanics, remains of outstanding importance. AO Dialogue Davos Platz, Switzerland: AO Foundation. Feb 2009. https://issuu.com/aofoundation/docs/aodialogue_2009_02_14/24. Accessed 23 Oct 2019.
3. Harvin WH, Oladeji LO, Della Rocca GJ, Murtha YM, Volgas DA, Stannard JP, Crist BD. Working length and proximal screw constructs in plate osteosynthesis of distal femur fractures. Injury. 2017;48(11):2597–601.
4. Brunner CF, Weber BG. Special techniques in internal fixation. Berlin/Heidelberg: Springer; 1982.
5. Wilber JH, Baumgaertel F. Bridge plating. AO principles of fracture management. New York: Thieme; 2007. p. 287.
6. Young JP, Stover MD, Magovern B. Treatment of isolated ulnar nonunions using wave plate osteosynthesis: a report of four cases. J Orthop Trauma. 2009;23(8):595–9.
7. Ring D, Jupiter JB, Sanders RA, Quintero J, Santoro VM, Ganz R, Marti RK. Complex nonunion of fractures of the femoral shaft treated by wave-plate osteosynthesis. J Bone Joint Surg Br. 1997;79(20):289–94.
8. Ring D, Jupiter JB, Quintero J, Sanders RA, Marti RK. Atrophic unurited diaphyseal fractures of the humerus with a bony defect: treatment by wave-plate osteosynthesis. J Bone Joint Surg Br. 2000;82(6):867–71.
9. Depuy Synthes. Angled blade plate for adults. Jun 2016. Accessed 20 Oct 2018.
10. Hartline BE, Achor TS. Use of the 95-degree angled blade plate to treat a proximal femur fracture. J Orthop Trauma. 2018;32(Suppl 1):S26–7.
11. Amorosa LF, Jayaram PR, Wellman DS, Lorich DG, Helfet DL. The use of the 95-degree-angled blade plate in femoral nonunion surgery. Eur J Orthop Surg Traumatol. 2014;24(6):953–60.
12. Abebe E, Farrell DJ, Zelle B, Gruen G. Primary posterior blade plate tibiotalar arthrodesis: a salvage procedure for complex nonreconstructable pilon fractures. J Orthop Trauma. 2017;31(Suppl 3):S30–3.
13. Berkes MB, Little MT, Lazaro LE, Cymerman RM, Helfet DL, Lorich DG. Catastrophic failure after open reduction internal fixation of femoral neck fractures with a novel locking plate implant. J Orthop Trauma. 2012;26(10):E170–6.
14. Glassner PJ, Tejwani NC. Failure of proximal femoral locking compression plate: a case series. J Orthop Trauma. 2011;25(2):76–83.
15. Collinge CA, Hymes R, Archdeacon M, Streubel P, Obremskey W, Weber T, Members of the Proximal Femur Working Group of the Southeast Trauma Consortium, et al. Unstable proximal femur fractures treated with proximal femoral locking plates: a retrospective, multicenter study of 111 cases. J Orthop Trauma. 2016;30(9):489–95.
16. Crist BD, Khalafi A, Hazelwood SJ, Lee MA. A biomechanical comparison of locked plate fixation with percutaneous insertion capability versus the angled blade plate in a subtrochanteric fracture gap model. J Orthop Trauma. 2009;23(9):622–7.
17. Depuy Synthes. DHS/DCS dynamic hip screw and condylar screw system: technique guide. Westchester PA: Synthes 2012; 2018. Accessed 30 Oct 2018.
18. Niu E, Yang A, Harris AH, Bishop J. Which fixation device is preferred for surgical treatment of intertrochanteric hip fractures in the United States? A survey of orthopaedic surgeons. Clin Orthop Relat Res. 2015;473(11):3647–55.

Locked Plating

15

Jason A. Lowe

Introduction

Development of locking technology was the response to failed fracture fixation when conventional plates were used as bridge plating constructs in the setting of metaphyseal comminution or osteoporosis. Mechanical failure of non-locking constructs occurs when axial loading creates sufficient shear force to overcome the friction created by screw torque compressing plate to the bone [1]. The resulting loss of fixation results in motion, which in turn increases fracture strain and instability leading to nonunion and hardware failure. Locking constructs avoid the challenges of conventional, non-locking, screw-plate constructs because the locking mechanism creates a fixed angle between plate and screw irrespective of bone quality. Fixation depends upon the locking mechanism and not bone quality. Each successive locked screw creates a fixed angle and adds to overall construct strength. In non-locking constructs however, each screw may loosen independently and has variable contribution to the overall construct strength. Therefore, locked plating is recommended for poor bone quality (osteoporosis or bone metabolic disease) and short segment articular fractures with associated metadiaphyseal comminution or bone loss secondary to open fractures where there is an expected prolonged healing time.

Locking technology was originally presented as a component of "biological internal fixation." Biological internal fixation, as originally described, is a principle that includes preservation of biology and reducing strain to a level (2–10%). "Biology" is preserved by application of plates that are not dependent upon compression to the bone. As a result, the periosteal blood supply is preserved. By achieving relative stability with a flexible construct in the metadiaphyseal region, secondary bone healing may occur prior to hardware failure [1–3].

Locking screws, however, have drawbacks as well. Early locking screw designs were limited by their unidirectional design. Unidirectional or monoaxial locking screws can only be inserted along the locking axis without compromising screw pull-out strength. Multidirectional or multi-axial locking screws allow for the mechanical benefits of a fixed angle interface between the plate and screw as well as the ability to target the screw through a fixed arc within the screw hole that typically maximally varies between 10° and 15° in each direction and a total arc of 20–30°. The freedom to target locking screws, however, decreases the strength to failure of the screw plate locking mechanism.

Furthermore, the initial locked plating systems were designed to be stiff and resist fatigue

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in an effort to avoid the hardware failure observed with non-locked plating. A drawback of increasing stiffness, particularly axial stiffness, is the subsequent reduction in strain, which can be sufficiently reduced below <2% which is the minimum strain desired to support secondary bone healing [1, 2, 4]. As such, overly stiff constructs can result in atrophic nonunion. To address excessive stiffness, locking screw design evolved to allow for controlled axial motion. The goal of controlled axial motion is to allow for symmetric motion and symmetric fracture callus without reducing construct strength.

In summary, locked plating continues to evolve in an effort to achieve flexible and stable fixation without compromising construct fatigue strength in the setting of comminution or bone loss, short segment fixation, and poor bone quality. Understanding the biomechanical principles of strain and how to modulate strain encourages successful union when applying locking constructs. In this chapter, the biomechanical benefits, limitations, and complications of locked plating are reviewed.

Monoaxial/Unidirectional Locking Screws

Monoaxial locking screws, sometimes referred to as uniaxial or unidirectional locking screws, represent the first generation of locked plating. As exemplified by the titanium PC-Fix and LISS plates (DePuy Synthes, Paoli, PA, USA), unidirectional screws are inserted orthogonal to the plate hole's long axis. Initially, these screws were available as self-drilling unicortical or bicortical screws (Fig. 15.1). Initial studies demonstrated successful union with secondary bone healing and no plate failures, which suggested that the locked plating construct had sufficient strength to maintain fracture alignment and flexibility to allow secondary bone healing [5, 6]. It is important to consider that these reports often included constructs with unicortical screws.

Monoaxial locking screws do not rely on friction created at the plate-bone interface for stability like cortical or cancellous screws. A fixed angle is created between the screw and plate in the locking screw hole with increased strength and stiffness. Understanding a locking screw's effect on construct stiffness is important because the increased stiffness can create a fixed gap in a simple fracture leading to high strain and result in a nonunion. If there is no fracture gap and no motion, strain is reduced. In the absence of bony contact (gap >50 microns) however, primary bone healing will not occur [3]. In this example, a fracture will remain atrophic until the construct loosens enough to allow sufficient fracture motion to either stimulate secondary healing or result in hardware failure and nonunion. Depending on the plate's strength (determined by

Fig. 15.1 Anteroposterior (AP) radiograph of a healed left intercondylar distal femur fracture demonstrates successful secondary bone healing (*white arrows*) with application of unicortical self-drilling, self-tapping screws (*black arrow*). Stress reactions can be seen around each unicortical screws suggesting loosening and increased motion during the healing process. However, catastrophic fixation failure did not occur

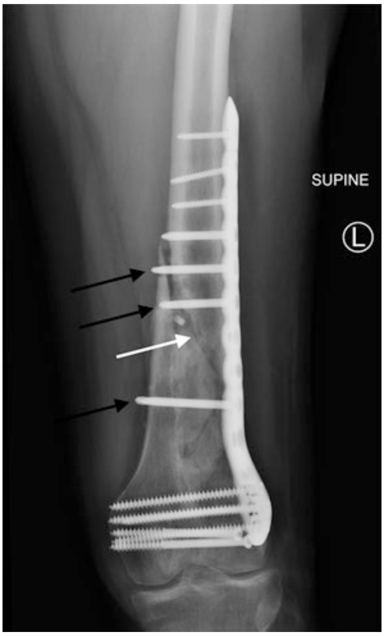

Fig. 15.2 Anteroposterior (AP) radiograph of a left supracondylar femoral atrophic nonunion. A gap is seen at the metaphysis (*white arrow*). Multiple locking screws traverse the fracture (*black arrows*) creating a fixed gap. A short working length and multiple locking screws increased stiffness and decreased motion leading to suppression of secondary bone healing and ultimate nonunion

the plate's geometric design, material, and screw configuration), loosening and subsequent callus will not form until implant fatigue begins [7, 8].

Comminuted fractures, which are inherently more mobile than simple fractures, may achieve sufficient stability with locking screws to allow early motion and secondary bone healing. It is possible, however, that the locked plating construct stiffness decreases fracture strain below 2% and suppresses secondary bone healing desired in a bridge plate construct (Fig. 15.2). Placing a screw close to the fracture site decreases the construct working length and increases construct fracture axial stability. But omitting a screw immediately adjacent to the fracture reduces axial stiffness and torsional rigidity by 64% and 36%, respectively [9]. The net effect is an increase in construct flexibility, but it also decreases fatigue strength sufficiently to increase the risk of implant failure [9]. Stoeffel et al. showed that yield strength decreased from 520 N to 350 N to 300 N as the working length was lengthened by moving the near screw away from the fracture, and cyclic load to failure decreased to 500,000 cycles vs one million. Clinically, Lujan et al. observed no difference in distal femur fracture callus nor implant failure when the working length was increased by omitting a screw immediately adjacent to the proximal extent of the fracture [10]. However, understanding working length is important when modulating construct stiffness and strength.

Adding cortical screws to a locked plating construct may reduce construct stiffness compared to an all locked construct. This concept is referred to as "hybrid fixation" and is a useful technique to affect fracture reduction by pulling the bone toward the plate. Controlling the plate bone distance is also important as stiffness and fatigue strength decrease with increased distance between the bone and plate. One benefit of not requiring plate bone compression is the preservation of periosteal blood supply. When there is minimal bone-plate distance, the strength of a hybrid and all locked construct are similar [9, 11]. Other studies have shown that hybrid constructs are 42% stronger in torsion than all locking constructs [12]. Gardner et al. showed similar biomechanical stiffness in hybrid and locked humeral fractures [13]. Preserving construct flexibility with hybrid fixation and longer working lengths should be balanced with preserving construct strength (Fig. 15.3).

Unicortical locking screws are another way to affect construct stiffness. Unicortical screws were initially used because they were easy to insert and because fixation into the second cortex offers little increase in screw pull-out strength [2, 3]. A unicortical screw's working length however is proportional to the thickness of cortical bone [8, 14]. The consequence of the shorter working length is more pronounced in osteoporotic bone due to thinner cortices. Unicortical locked constructs are 69% weaker in torsion than non-locking screws [8]. This is important for bones that see high torsional load like the humerus and forearm because they are at greater risk for screw pull-out and failure [15, 16]. Unicortical screws are also weaker in bending compared to bicortical

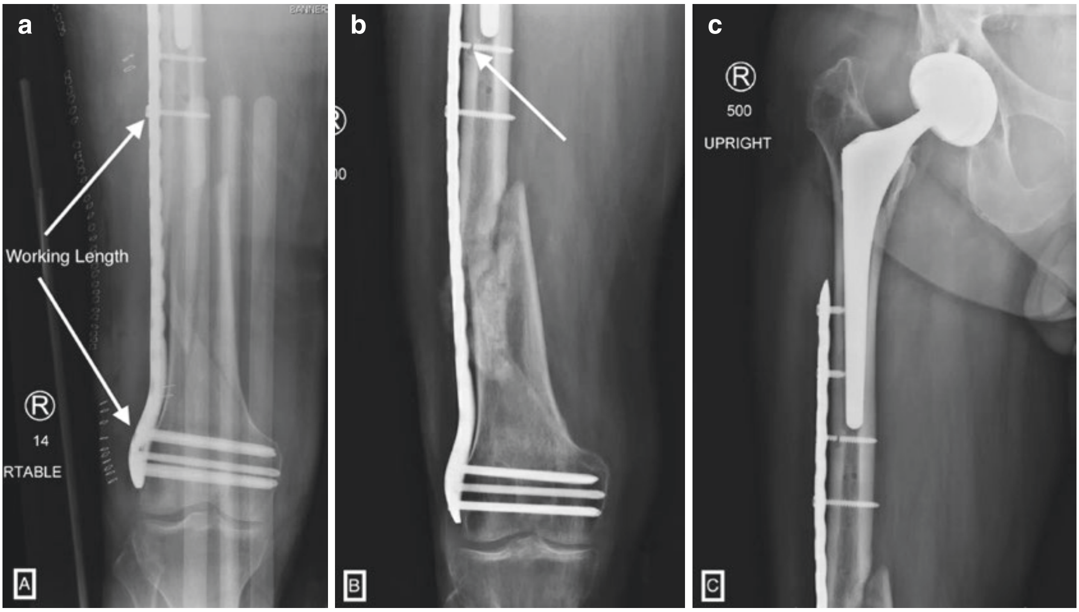

Fig. 15.3 Anteroposterior (AP) radiographs of a 70-year-old with a supracondylar fracture stabilized with a long titanium plate. (**a**) Nonanatomical reduction is noted, but length, alignment, and rotation are preserved. (**b**) At 3 months, hardware failure is noted by a fracture of the cortical screw and bending of the plate with increased varus alignment. A large working length is noted. Although early callus is observed, the implant failed. The construct's strength could have been improved with either anatomical reduction of the spiral fracture (load-sharing construct), or additional screws proximally, or a shorter working length. (**c**) Proximal hybrid fixation is seen with a broken cortical screw

screws. For this reason, unicortical locking screws are reserved for diaphyseal fixation in non-osteoporotic bone, in the setting of periprosthetic fractures or as a means to reduce construct stiffness at the end of the plate. Decreasing construct stiffness at the end of the plate in osteoporotic bone may decrease the risk of peri-implant fractures [17]. If locking screws are used in the diaphysis, placing a cortical or unicortical locking screw at the end of the plate may sufficiently reduce the plate's stiffness to decrease the risk of peri-implant fracture [18]. On the other hand, if increased bending and torsional stiffness is desired, a bicortical locking screw can be used [19].

The inability to angularly direct monoaxial screws can limit fragment specific fixation and ability to achieve locking fixation around other implants. Unidirectional screws may fail to capture key fracture fragments or may be directed toward the articular surface [20]. If used in isolation, a lateral only monoaxial, locked plating of bicondylar tibia plateau does not afford adequate purchase of the posteromedial fragment and has a statistically lower load to failure than augmenting the posteromedial fragment with a 1/3 tubular or dynamic compression plate ($p = 0.006$) [21]. In an effort to achieve fixation around implants, plates can be positioned eccentrically on the bone. However, eccentric positioning can lead to limited bone engagement or the transcortical fixation that compromises construct stability and may create a stress riser increasing the risk of peri-implant fractures.

The limitations of unidirectional locked screws became apparent. Unidirectional locked fixation supports stable fixation provided the insertion angle is <5° from neutral [22, 23]. If the unidirectional locked screw insertion angle is greater than 10° from neutral axis, the screw pull-out force decreases by 77%, and bending load to failure decreases by 69% [23]. In smaller diameter screws, it is believed that <2°of angular malalignment will self-correct, but larger implants will not self-correct and failure can occur [22, 24]. These narrow insertion angles preclude targeting around existing implants, avoiding articular surfaces, and capturing fracture fragments.

Polyaxial/Multidirectional Locking Screws

To address the limitation of unidirectional screws, plating systems have evolved to allow locked screw insertion through a predesigned arc while preserving the benefit of a fixed angle construct (Figs. 15.4 and 15.5). Multiple screw plate locking mechanisms are available for multidirectional locking including point-loading thread in, cut-in, locking cap, expansion bushing, and screw-head expansion [17, 25, 26]. However, the integrity of the locking mechanism as the insertion angle progresses from a neutral axis changes. Lenz et al. evaluated the DePuy Synthes variable angle locking screws [22, 27]. Similar failure load moments and screw failure mechanisms (screw head fracture) were observed when the screw insertion angle was between zero (coaxial with a unidirectional screw) and 10°. But when the insertion angle was at least 15 degrees, the screws disengaged from the plate and demonstrated a lower failure moment. Other studies have shown that multidirectional locking screws have a lower resistance to rotational failure and moment to failure than unidirectional locking screws. Failure moments increase linearly as the angle of inser-

Fig. 15.4 Two photos demonstrate locking screws placed at the extremes of allowable insertion angles through a variable angle distal medial humerus plate. (DePuy Synthes, Paoli, PA, USA)

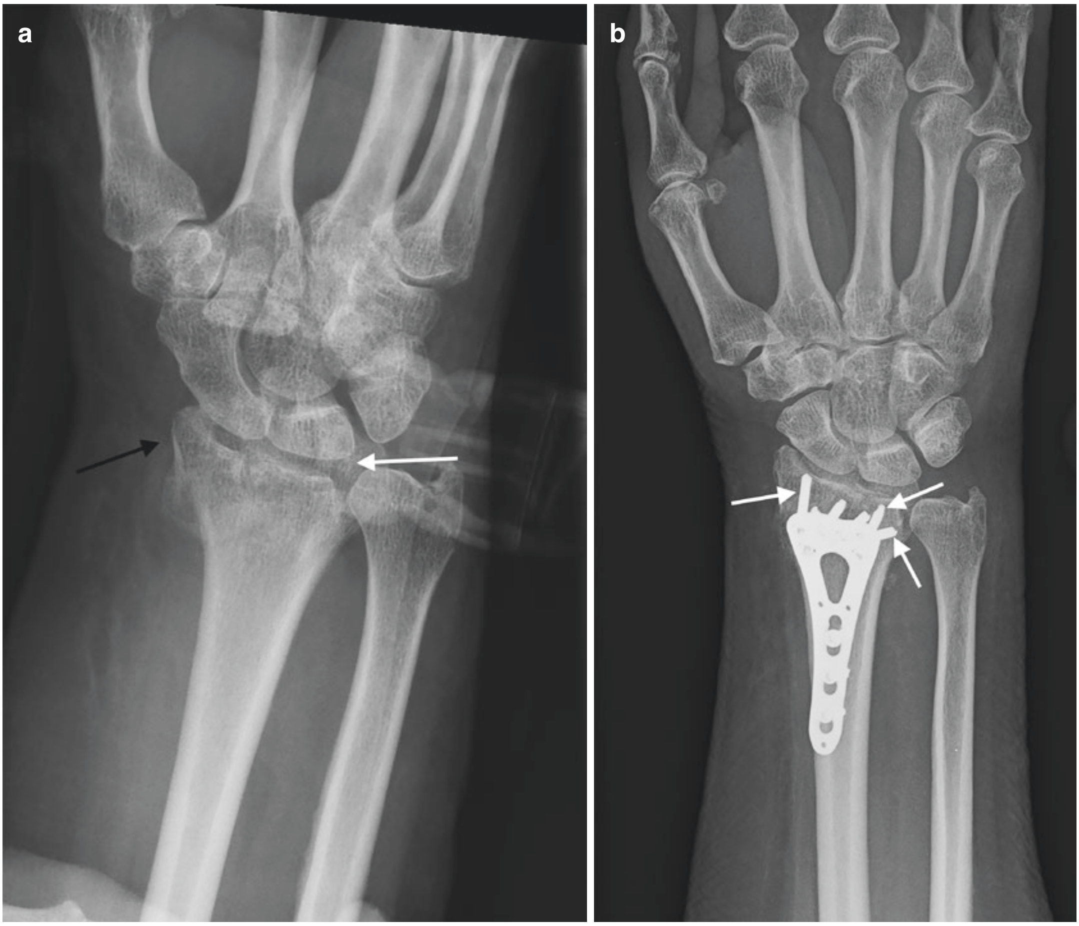

Fig. 15.5 (**a**) Posteroanterior (PA) radiographs of a left comminuted intra-articular distal radius with volar lunate fragment (*white arrow*) and radial styloid fragment (*black arrow*). (**b**) Postoperative PA radiograph post-fixation with multidirectional locking screws angled into the volar lunate fragment and radial styloid fragment (*white arrows*)

tion increases from 0° to 15° [28]. A direct comparison of three multidirectional locking systems showed significant reduction (45% and 43%) in force required to displace a Stryker VariAx (Stryker, Mahwah, NJ, USA) and Smith and Nephew PERI-LOC (Smith and Nephew Memphis, TN, USA) screw as screw insertion angle increased. The Zimmer NCB (Zimmer Biomet, Warsaw, IN, USA) demonstrated increased load to displacement, but larger variability between samples [17]. At present, the literature on biomechanical properties of multidirectional locking screws is limited, but there appears to be consensus that as the angle of insertion increases from zero, the strength of the screw plate locking mechanism decreases.

Multidirectional locking plate technology allows the ability to target specific fracture fragments (see Fig. 15.5) or angle locking screws around existing implants. In doing so, however, fixation strength can be compromised. Interestingly, the loss of fixation observed with multidirectional locked screws parallels failures observed with non-locking implants. Currently, it is recommended to limit the insertion angle of multidirectional locking screws to <10° from the central axis. There may be a mechanical benefit to locking cap stabilization over other mechanisms, but failures still occur [17]. Relying on targeted screws to maintain fracture fixation, particularly at the extremes of angular insertion, may predispose the construct to mechanical failure [29, 30].

Flexible Locking

Following initial reports of success with monoaxial locking systems, nonunions and late hardware failures were reported [10, 31–33]. The intended construct stiffness benefits of locking constructs led to unintended consequences like promoting asymmetric callus formation with resorption of the bone under the plate due to stress shielding. Lujan et al. observed this phenomenon in the distal femur [10]. The stiffness pendulum swung too far, and flexible locking was developed in an attempt to allow for controlled axial motion to promote symmetric fracture callus. Like polyaxial locking screws, several different flexible locking constructs were developed [34, 35].

Flexible locking constructs derive their flexibility from either screw or plate design. Far Cortical Locking (FCL) screws (Zimmer Biomet) represent flexible locking through screw design. This implant reduces construct stiffness and promotes symmetric fracture motion by locking into the plate and the far cortex. Axial motion is permitted by a reduced screw shaft diameter relative to the screw head and distal threads allowing for elastic deformation of the screw during loading, as well as by the diameter of the collar adjacent to the screw head. This design allows for up to 0.54–0.6 mm of motion across the fracture correlating to symmetric callus formation and greater mineral content within the callus—44% more in FCL compared to standard locking plate [36–38]. Bottlang showed that FCL screws reduced stiffness in the axial (88%), torsional (58%), and bending (29%) stiffness compared to standard locking constructs without reducing axial strength [34]. Additionally, he noted that bending and torsional strength of the FCL constructs were increased in both osteoporotic and non-osteoporotic models.

An alternative FCL design is Dynamic Locking Screws (DLS) (DePuy Synthes). DLS permitted axial motion through a mechanical sleeve within the screw. As opposed to the FCL screw, the DLS anchored in both the near and far cortical bone. Richter et al. showed that fractures fixed with DLS, compared to standard locking, had a greater maximum failure moment, greater periosteal callus volume at the near cortex and intercortical region, and greater torsional stiffness (84% vs 58% of intact tibia; $p = 0.027$) with homogenous interfragmentary strain in an ovine model [35]. As constructs integrate motion into locked constructs, there is a risk for implant failure or the possibility of introducing too much motion (strain) leading to hypertrophic nonunion. For example, the DLS was removed from the US market in 2013 due to implant failures [39].

Dynamic stabilization with “Active” locking plates represents another form of flexible locking fixation. As the name implies, motion occurs through the plate screw hole design using alternately spaced holes, which, by design, incorporates a sliding element that is suspended in a silicone envelope that allows for 1.5 mm of axial motion. Ovine fracture models demonstrated increased fracture callus ($p < 0.001$), 81% of initial torsional strength, and 399% stronger in torsion than statically locked constructs [40].

To date, one study has compared flexible locking fixation constructs. Using synthetic bone models, Henschel et al. compared unidirectional locking, non-locked, bridged (omitting two screws adjacent to the fracture), Far Cortical Locking, and Active locking plates [41]. Nonlocking constructs had similar stiffness and axial motion to unidirectional locking plates. Bridge plating constructs reduced stiffness by 45%, but the interfragmentary motion was shear and associated with nonunion. FCL and Active locking designs significantly reduced stiffness by 62% and 75%, respectively ($p < 0.001$). Clinical data supporting the biomechanical benefits of flexible locking constructs is currently lacking.

Summary

Locking technology, together with techniques for moderating strain and construct strength (plate material length, screw density, screw spread) discussed in other chapters, is necessary to create a biomechanical environment that will promote the desired mode of healing. Locking technology was developed to improve fracture stability in complex fractures that otherwise were prone to the complications of implant loosening and loss of fixation. Transitioning to a system that relied on screw-plate interface for stability, locking technology allowed for load-bearing constructs that were successful in maintaining fixation of complex periarticular and osteoporotic fractures. The increased construct stiffness over non-locking fixation could sufficiently reduce strain enough to inhibit bone healing in comminuted or simple fractures leading to nonunion and implant failure. However, locking plate technology is evolving. Multidirectional locking screws afford the increased ability to direct locking screws, but increasing insertion angles compromise the screw plate locking mechanism. Although techniques to reduce locking construct strain have developed, additional studies are required to completely understand the clinical impact and potential complications.

References

1. Egol KA, Kubiak EN, Fulkerson E, Kummer FJ, Koval KJ. Biomechanics of locked plates and screws. J Orthop Trauma. 2004;18(8):488–93.
2. Perren SM, Matter P. Evolution of AO philosophy. Acta Chir Orthop Traumatol Cechoslov. 2003;70(4):205–6.
3. Perren SM. Backgrounds of the technology of internal fixators. Injury. 2003;34(Suppl 2):B1–3.
4. Perren SM, Matter P, Ruedi R, Allgower M. Biomechanics of fracture healing after internal fixation. Surg Annu. 1975;7:361–90.
5. Haas N, Hauke C, Schutz M, Kaab M, Perren SM. Treatment of diaphyseal fractures of the forearm using the Point Contact Fixator (PC-Fix): results of 387 fractures of a prospective multicentric study (PC-Fix II). Injury. 2001;32(Suppl 2):B51–62.
6. Tepic S, Remiger AR, Morikawa K, Predieri M, Perren SM. Strength recovery in fractured sheep tibia treated with a plate or an internal fixator: an experimental study with a two-year follow-up. J Orthop Trauma. 1997;11(1):14–23.
7. Beltran MJ, Collinge CA, Gardner MJ. Stress modulation of fracture fixation implants. J Am Acad Orthop Surg. 2016;24(10):711–9.
8. Gautier E, Sommer C. Guidelines for the clinical application of the LCP. Injury. 2003;34(Suppl 2):B63–76.
9. Stoffel K, Dieter U, Stachowiak G, Gachter A, Kuster MS. Biomechanical testing of the LCP–how can stability in locked internal fixators be controlled? Injury. 2003;34(Suppl 2):B11–9.
10. Lujan TJ, Henderson CE, Madey SM, Fitzpatrick DC, Marsh JL, Bottlang M. Locked plating of distal femur fractures leads to inconsistent and asymmetric callus formation. J Orthop Trauma. 2010;24(3):156–62.
11. Ahmad M, Nanda R, Bajwa AS, Candal-Couto J, Green S, Hui AC. Biomechanical testing of the locking compression plate: when does the distance between bone and implant significantly reduce construct stability? Injury. 2007;38(3):358–64.
12. Doornink J, Fitzpatrick DC, Boldhaus S, Madey SM, Bottlang M. Effects of hybrid plating with locked and nonlocked screws on the strength of locked plating constructs in the osteoporotic diaphysis. J Trauma. 2010;69(2):411–7.
13. Gardner MJ, Griffith MH, Demetrakopoulos D, Brophy RH, Grose A, Helfet DL, et al. Hybrid locked plating of osteoporotic fractures of the humerus. J Bone Joint Surg Am. 2006;88(9):1962–7.
14. Pater TJ, Grindel SI, Schmeling GJ, Wang M. Stability of unicortical locked fixation versus bicortical non-locked fixation for forearm fractures. Bone Res. 2014;2:14014.
15. Sommer C, Babst R, Muller M, Hanson B. Locking compression plate loosening and plate breakage: a report of four cases. J Orthop Trauma. 2004;18(8):571–7.
16. Fitzpatrick DC, Doornink J, Madey SM, Bottlang M. Relative stability of conventional and locked plating fixation in a model of the osteoporotic femoral diaphysis. Clin Biomech (Bristol, Avon). 2009;24(2):203–9.
17. Hebert-Davies J, Laflamme GY, Rouleau D, Canet F, Sandman E, Li A, et al. A biomechanical study comparing polyaxial locking screw mechanisms. Injury. 2013;44(10):1358–62.
18. Bottlang M, Doornink J, Byrd GD, Fitzpatrick DC, Madey SM. A nonlocking end screw can decrease fracture risk caused by locked plating in the osteoporotic diaphysis. J Bone Joint Surg Am. 2009;91(3):620–7.
19. Dougherty PJ, Kim DG, Meisterling S, Wybo C, Yeni Y. Biomechanical comparison of bicortical versus unicortical screw placement of proximal tibia locking plates: a cadaveric model. J Orthop Trauma. 2008;22(6):399–403.
20. Barei DP, O'Mara TJ, Taitsman LA, Dunbar RP, Nork SE. Frequency and fracture morphology of the posteromedial fragment in bicondylar tibial plateau fracture patterns. J Orthop Trauma. 2008;22(3):176–82.
21. Yoo BJ, Beingessner DM, Barei DP. Stabilization of the posteromedial fragment in bicondylar tibial plateau fractures: a mechanical comparison of locking and nonlocking single and dual plating methods. J Trauma. 2010;69(1):148–55.
22. Lenz M, Wahl D, Gueorguiev B, Jupiter JB, Perren SM. Concept of variable angle locking–evolution and mechanical evaluation of a recent technology. J Orthop Res. 2015;33(7):988–92.
23. Kaab MJ, Frenk A, Schmeling A, Schaser K, Schutz M, Haas NP. Locked internal fixator: sensitivity of

screw/plate stability to the correct insertion angle of the screw. J Orthop Trauma. 2004;18(8):483–7.
24. Schneider K, Oh JK, Zderic I, Stoffel K, Richards RG, Wolf S, et al. What is the underlying mechanism for the failure mode observed in the proximal femoral locking compression plate? A biomechanical study. Injury. 2015;46(8):1483–90.
25. Gueorguiev B, Lenz M. Why and how do locking plates fail? Injury. 2018;49(Suppl 1):S56–60.
26. Schoch B, Hast MW, Mehta S, Namdari S. Not all polyaxial locking screw technologies are created equal: a systematic review of the literature. JBJS Rev. 2018;6(1):e6.
27. Lenz M, Wahl D, Zderic I, Gueorguiev B, Jupiter JB, Perren SM. Head-locking durability of fixed and variable angle locking screws under repetitive loading. J Orthop Res. 2016;34(6):949–52.
28. Tidwell JE, Roush EP, Ondeck CL, Kunselman AR, Reid JS, Lewis GS. The biomechanical cost of variable angle locking screws. Injury. 2016;47(8):1624–30.
29. Tank JC, Schneider PS, Davis E, Galpin M, Prasarn ML, Choo AM, et al. Early mechanical failures of the synthes variable angle locking distal femur plate. J Orthop Trauma. 2016;30(1):e7–e11.
30. Otto RJ, Moed BR, Bledsoe JG. Biomechanical comparison of polyaxial-type locking plates and a fixed-angle locking plate for internal fixation of distal femur fractures. J Orthop Trauma. 2009;23(9):645–52.
31. Button G, Wolinsky P, Hak D. Failure of less invasive stabilization system plates in the distal femur: a report of four cases. J Orthop Trauma. 2004;18(8):565–70.
32. Cole PA, Zlowodzki M, Kregor PJ. Treatment of proximal tibia fractures using the less invasive stabilization system: surgical experience and early clinical results in 77 fractures. J Orthop Trauma. 2004;18(8):528–35.
33. Weight M, Collinge C. Early results of the less invasive stabilization system for mechanically unstable fractures of the distal femur (AO/OTA types A2, A3, C2, and C3). J Orthop Trauma. 2004;18(8):503–8.
34. Bottlang M, Doornink J, Fitzpatrick DC, Madey SM. Far cortical locking can reduce stiffness of locked plating constructs while retaining construct strength. J Bone Joint Surg Am. 2009;91(8):1985–94.
35. Richter H, Plecko M, Andermatt D, Frigg R, Kronen PW, Klein K, et al. Dynamization at the near cortex in locking plate osteosynthesis by means of dynamic locking screws: an experimental study of transverse tibial osteotomies in sheep. J Bone Joint Surg Am. 2015;97(3):208–15.
36. Bottlang M, Lesser M, Koerber J, Doornink J, von Rechenberg B, Augat P, et al. Far cortical locking can improve healing of fractures stabilized with locking plates. J Bone Joint Surg Am. 2010;92(7):1652–60.
37. Bottlang M, Feist F. Biomechanics of far cortical locking. J Orthop Trauma. 2011;25(Suppl 1):S21–8.
38. Dobele S, Horn C, Eichhorn S, Buchholtz A, Lenich A, Burgkart R, et al. The dynamic locking screw (DLS) can increase interfragmentary motion on the near cortex of locked plating constructs by reducing the axial stiffness. Langenbeck's Arch Surg. 2010;395(4):421–8.
39. U.S. Food and Drug Administration. FDA Home. Medical Devices. Databases. Class 2 Device Recall Synthes 3.7 and 5.0mm Dynamic Locking Screwn (DLS). https://www.accessdata.fda.gov/scripts/cdrh/cfdocs/cfres/res.cfm?id=122920. Accessed 4 Jul 2019.
40. Bottlang M, Tsai S, Bliven EK, von Rechenberg B, Klein K, Augat P, et al. Dynamic stabilization with active locking plates delivers faster, stronger, and more symmetric fracture-healing. J Bone Joint Surg Am. 2016;98(6):466–74.
41. Henschel J, Tsai S, Fitzpatrick DC, Marsh JL, Madey SM, Bottlang M. Comparison of 4 Methods for dynamization of locking plates: differences in the amount and type of fracture motion. J Orthop Trauma. 2017;31(10):531–7.

Part V

Intramedullary Nailing Principles with Case Examples

16 Biomechanics of Intramedullary Nails Relative to Fracture Fixation and Deformity Correction

Justin C. Woods and Gregory J. Della Rocca

Introduction

Modern intramedullary nailing began in the World War II era with Gerhard Küntscher's hypothesizing that his intramedullary implant would act as an internal splint, allowing a fractured bone to heal [1]. As with many advancements in medicine, Küntscher's idea was initially met with skepticism; standard of care at the time was to treat long bone fractures with external immobilization such as plaster casts or skeletal traction. His original design was a V-shaped stainless-steel nail, replaced several years later by the more universally recognized cloverleaf-shaped nail. Multiple design advances followed, with the result that decades later, the intramedullary nail has become the workhorse in the armamentarium of the orthopedic trauma surgeon for long bone fracture and deformity care. Current nail design features that have an effect on nail biomechanics include the following: material (metal) properties, cross-sectional shape, diameter, curvature of the nail, and the ability to place interlocking devices (such as bolts). Extrinsic factors, such as reaming of the medullary canal, inherent fracture stability (fracture pattern, including comminution or spiral configuration), and the use of adjuncts for stability (such as blocking screws), also affect biomechanics of fracture fixation. This chapter will describe biomechanics of intramedullary nailing, with illustrative examples.

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Biomechanics

Young's modulus of elasticity is defined as the stress (force per unit area) divided by the strain (change in length divided by the original length) [2]. This concept is thought to be important in orthopedic care as it relates the stiffness of an implant to that of human bone. Intramedullary nails have varying degrees of flexibility, based upon nail geometry, material, and size. This flexibility can directly affect fracture behavior and has an influence on callus formation at the site of fracture. The current type of stainless steel found in orthopedic implants is 316L, which is composed of molybdenum (3%), nickel (16%), and low levels of carbon [3]. The modulus of elasticity of 316L stainless steel is 193 GPa, which is significantly higher than that of human cortical bone, which is approximately 18.6 GPa [4]. The most common titanium alloy (Ti6Al4V) used in orthopedic implants today is composed of titanium, aluminum (6%), and vanadium (4%) and has a Young's modulus of elasticity of 115 GPa [5]. Perhaps given the fact that the Young's modulus of elasticity for titanium alloy implants is much closer to cortical bone (and much less rigid

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than 316L stainless steel), most modern intramedullary nails are made of titanium alloy.

The shape of intramedullary nails has changed throughout the years as our understanding of anatomy and biology has improved. Küntscher's nails had a longitudinal slot that allowed the nail to be relatively flexible and conform to the endosteal surface of the bone. This was important, as the femur is not a straight bone, and the tibia requires an eccentric start point (resulting in a nail pathway that is not straight). The conformity of the nail to the endosteal surface also provided a friction fit of the nail, allowing it to maintain fracture reduction. Long bones experience significant bending and torsional forces, and an intramedullary implant must be able to resist these stresses during fracture healing. The slot in Küntscher's nails decreased the torsional rigidity of the implant, which was felt perhaps to be suboptimal. Solid nails were then developed; however, these were felt to be too rigid, and they lacked the ability to be inserted over a guide wire. Cannulated nails were then developed and are used to this day; this confers the benefit of increased torsional rigidity (over slotted nails) and the convenience of being placed over a guide wire. The rigidity of a round intramedullary nail is directly proportional to its radius: torsional rigidity is proportional to the fourth power of the radius of the nail (although this is also proportional to the strength of interlocking bolts with modern nail designs when considering fracture stabilization), and bending rigidity is proportional to the third power of the radius of the nail.

Intramedullary devices (unlocked) provide coronal and sagittal plane bending stability, but there is minimal resistance to change in length or rotation of the fractured bone. Küntscher's nails overcame this problem via friction fit, but later-developed nails lost the ability to obtain a friction fit for all but isthmic diaphyseal fractures with minimal comminution. Grosse and Kempf are credited with introducing the concept of interlocking screws, which connect the bone to the nail in locations both proximal and distal to the fracture [6]. Interlocking solved these two major limitations of intramedullary nailing by minimizing shortening and rotational instability of a properly reduced fracture treated with such devices. As many long bone fractures lack cortical contact between the proximal and distal segments, due to comminution, the nailed fracture fixation construct may rely entirely on the interlocking screw-nail interface for stability.

There is some inherent flexibility of the construct when a fracture is treated with intramedullary nailing. Given that intramedullary nails act as an internal splint, micromotion is expected at the fracture site. No matter the degree of comminution, callus formation is expected to occur as the bone heals.

Reaming the intramedullary canal allows for a larger-diameter intramedullary nail to be inserted, increasing construct strength and rigidity. This may have important advantages for long bone fracture fixation. First, current-generation nails may be larger than the native medullary canal and may become incarcerated during insertion without prior reaming. Second, a larger nail may be inserted after reaming, increasing nail longevity and strength (longevity is important, especially when bone healing progresses slowly). Third, current-generation intramedullary nails may have variations in interlocking screw diameter, with larger interlocking screws intended for use with larger-diameter nails (and attendant increases in durability and torsional rigidity associated with larger-diameter interlocking screws).

Fractures and Deformities

Most long bones in the human body can be treated with an intramedullary nail. The most common bones treated with nailing are the tibia, femur, and humerus. Intramedullary nailing of the clavicle, radius, and ulna has been described; these constructs may or may not be interlocking, based upon nail design utilized (see [7] and [8] for reviews). The serpiginous nature of the medullary canals of the clavicle, radius, and ulna makes for increased difficulty in maintaining anatomical reductions using intramedullary devices; plate-and-screw fracture fixation constructs tend to be favored for all of these bones except in some circumstances (such as pediatric forearm fractures [9]).

Biologically, intramedullary nails may offer advantages for long bone fracture fixation that are not afforded by plate-and-screw constructs. In many cases, the fracture is stabilized via direct access to the bone at a distance from the fracture site itself. Theoretically, the fracture site remains relatively undisturbed, allowing the fracture hematoma and its associated growth factors to remain in place, in addition to the putative benefit of depositing endosteal reamings at the fracture site when reamed nailing is used. Nails and associated interlocking screws are typically placed percutaneously through smaller incisions. If a clamp is used, it can typically be placed percutaneously or through small incisions that are less likely to cause periosteal injury than that which might occur during fracture treatment with plate-and-screw fixation.

Intramedullary nailing is not without its challenges. Although most nailing procedures minimize further injury to the periosteal blood supply at the fracture site, nailing +/− reaming of the intramedullary canal disrupts the endosteal blood supply. Malalignment and malreduction may also occur following nailing of certain fracture patterns. For example, intramedullary nails do not achieve good endosteal fit at a metaphyseal fracture site; for this reason, reduction *prior* to nailing is imperative, and consideration to the use of fracture fixation augments (such as blocking screws) may be beneficial to assist fracture reduction during surgery and prevent loss of reduction during healing. Articular fractures of long bones may not be adequately stabilized by intramedullary nails for a variety of reasons: loss of reduction may occur during canal preparation and nail insertion, interlocking screws may not provide sufficient stability for fractures orthogonal to their insertion, and (for example) coronal plane articular fractures may not be captured by interlocking screws. For these reasons, intramedullary nails are either avoided for stabilization of articular fractures, or they are used in conjunction with independent screws and/or plate-and-screw constructs for fracture stabilization.

The intramedullary nail is a powerful tool that an orthopedic surgeon may employ to treat malunions, nonunions, bone defects, or other osseous malformations. Malunions may be corrected with an appropriate osteotomy followed by intramedullary nailing. If a fracture nonunion occurs after intramedullary nailing, one option for treatment is to perform an exchange nailing, during which a larger-sized nail is often placed after reaming. This provides the theoretical benefits of providing autograft at the site of the nonunion (via reaming) as well as increasing overall stability of the construct. For fracture deformities, bone transport or lengthening procedures may be performed utilizing an intramedullary nail as a “rail” over which the bone is transported, providing the benefit of using the nail as a guide for correction [10].

Intramedullary nailing for long bone fixation, particularly in the lower extremities, has been highly successful in the younger patient with high-quality bone. Many studies, however, have noted that fixation failures occur more often in osteoporotic bones—particularly in the proximal and distal aspect of the bones where nail interlocking sites are located (cancellous bone quality in these locations is often compromised). Several strategies have been developed to address this problem, such as modifying the geometry of intramedullary nails used for diaphyseal fracture fixation, decreasing bone removal during interlocking by using blades instead of screws, using blocking screws adjacent to the nail (which can function as “cortical substitutes”), and using fixed angle interlocking screws [11–15].

The proximal geometry of most nails has been modified over the years, along with the bow of nails for femoral nailing, to help decrease the stress on the intramedullary canal during insertion and seating of the nail and during fracture healing. A great deal of work was done early after the advent of intramedullary nailing in order to understand insertion stresses and the importance of location of the entry point, particularly in the proximal femur. With the larger proximal diameter nails used for cephalomedullary nailing, it has become increasingly important to not start them too anterior in the intertrochanteric region to prevent hoop stresses that could lead to bursting of the proximal segment [16].

The human femur is not a straight bone. The adult femur has a radius of curvature that approximates 1.2 m [17]. Early iteration nails (as described previously) were straight but flexible and able to “bend” to accommodate the curvature of the femur. As nail design changed, increasingly rigid nails were less likely to accommodate the curvature of the femur, potentially leading to proximal femoral bursting. In osteoporotic patients, a straight nail may perforate the anterior distal femoral metaphysis, leading to iatrogenic distal femur fracture [18–20]. Decreasing the radius of curvature allows for the nail tip to seat more posteriorly in the distal femur, potentially reducing the likelihood of this problem.

Stability of nails used for fixation of fractures is partially related to the bone quality at interlocking locations. Numerous modifications of interlocking techniques have led to theoretical improvements in the ability of nails to stabilize fractures in osteoporotic patients. Some devices utilize interlocking blades, as opposed to screws/bolts, near the insertion site of the nail – ostensibly, to reduce the amount of bone removal that occurs when drilling and placing screws or bolts [21]. Multiaxial interlocking options also may improve the stability and longevity of interlocking nail constructs, as compared to interlocking simply in the coronal plane with 1–2 interlocking devices [22]. Angular stability of interlocking screws can be enhanced with locking end caps [23, 24] or by placement of sleeves within the interlocking holes to minimize screw toggle [25].

Blocking (“poller”) screws are highly useful to minimize angular deformities that may occur after nailing of metadiaphyseal fractures, especially in patients with osteoporotic bone. First described in the literature by Krettek and colleagues [26, 27], these devices (most often screws) are placed adjacent to the nail in order to prevent angular deformity from occurring after nailing and interlocking have been completed. Osteoporotic patients may be more susceptible to these angular deformities, as their metaphyseal bone quality is inferior to that of non-osteoporotic patients. In some, the cortices may represent practically the only bone potentially in contact with the nail and interlocking screws (younger patients have higher-quality trabecular bone, and the nail theoretically sits in a “tunnel” of this bone, which may help prevent post-nailing deformity). Screws can be placed adjacent to the nail, either in the coronal or sagittal plane (or both), to resist deformities that are not countered adequately by the nail itself (which is often much smaller than the canal in which it is seated at the metaphysis) or by interlocks.

Case Examples

Case 1: Transverse Midshaft Femur Fracture (Fig. 16.1)

A 17-year-old male fell asleep behind the wheel while driving home late at night. The car left the roadway and struck a tree. The patient awoke with severe left thigh pain and was brought to the hospital with an isolated, closed, minimally comminuted, transverse left femoral shaft fracture at the isthmus of the medullary canal. He was placed into skeletal traction at the time of admission and had no associated injuries. There was no evidence of a femoral neck fracture. On the subsequent day, he was transported to the operating room for antegrade, trochanteric-entry intramedullary nailing of his left femur. A trochanteric entry site was selected to minimize risk to the femoral head blood supply, in light of the patient’s young age and (likely) recent skeletal maturity. The fracture was reducible without accessory incisions near the fracture site, and anatomical alignment was noted after nailing. A single static interlocking screw was placed proximally, and a single static interlocking screw was placed distally. The patient was allowed to weight bear as tolerated following surgery.

Why This Worked

First, the fracture is located in the mid-diaphysis, allowing for good fit between the nail and the canal at the site of the fracture. Second, the fracture was properly reduced, and reduction was maintained, during nail placement. Third, the fracture was minimally comminuted and was essentially transverse, allowing for the bone to share the load with the proximal and distal interlocking screws when resisting compression during weight-bearing. Since the nail had good fit, and the fracture pattern and reduction allowed for good cortical contact to share the mechanical load, only one locking screw was placed proximally and distally. If the fracture was comminuted with limited cortical contact and the nail did not have good diaphyseal fit, more locking screws would have been used. Fourth, design of the nail allowed for an eccentric starting point (proximal nail bend) along with accommodation of the anterior bow of the femur (curved nail along the bulk of its length) so as not to induce a deformity at the fracture site after nail insertion and to allow for relaxation of stress at the proximal femur after nail seating.

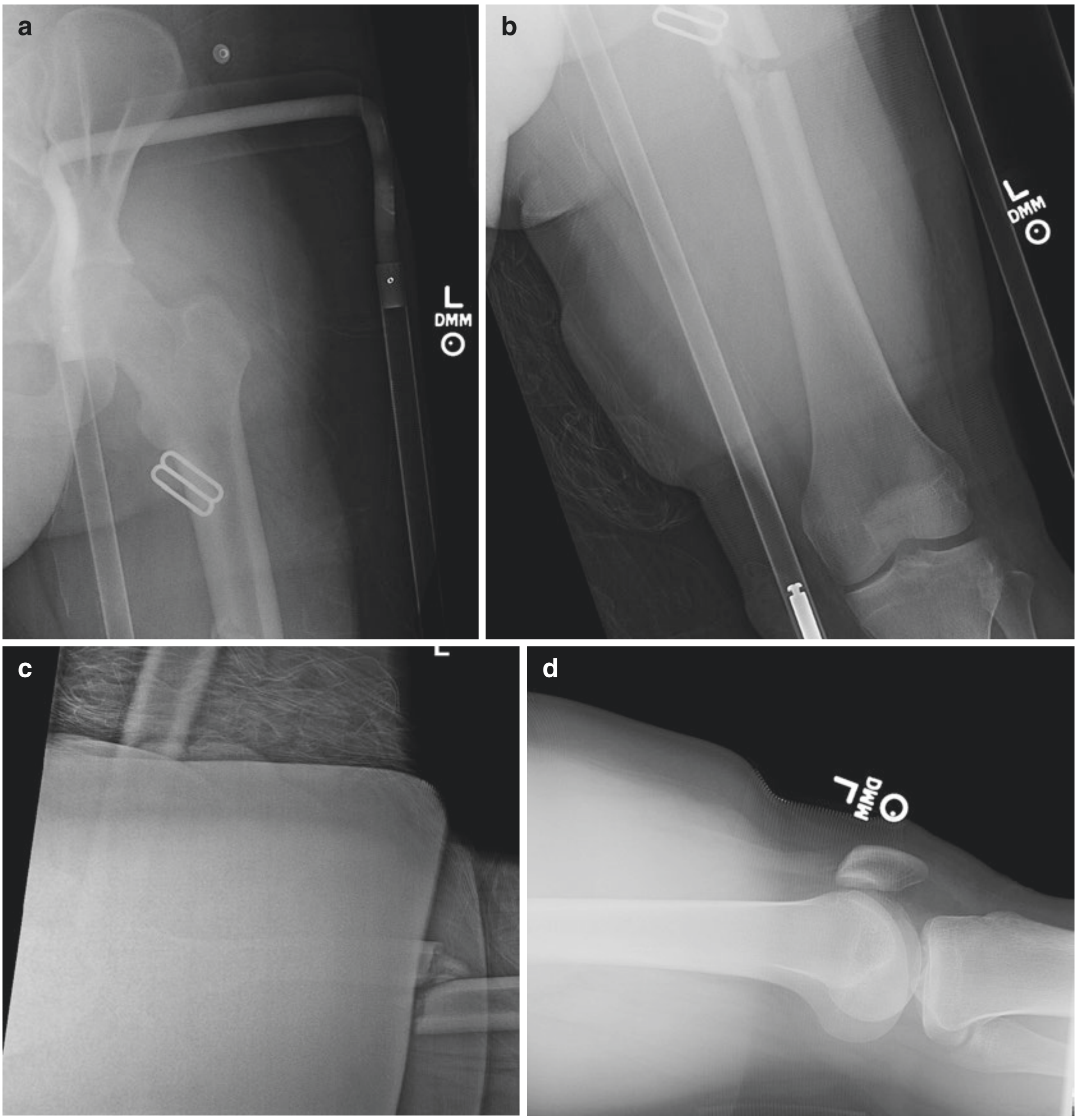

Fig. 16.1 Anteroposterior (**a**, **b**) and lateral (**c**, **d**) radiographs of a midshaft, minimally comminuted left femur fracture in a 17-year-old patient. Anteroposterior (**e**, **f**) and lateral (**g**, **h**) radiographs obtained postoperatively of the same patient's left femur after antegrade intramedullary nailing. Anteroposterior (**i**) radiograph obtained 6 months after surgical repair, demonstrating complete healing

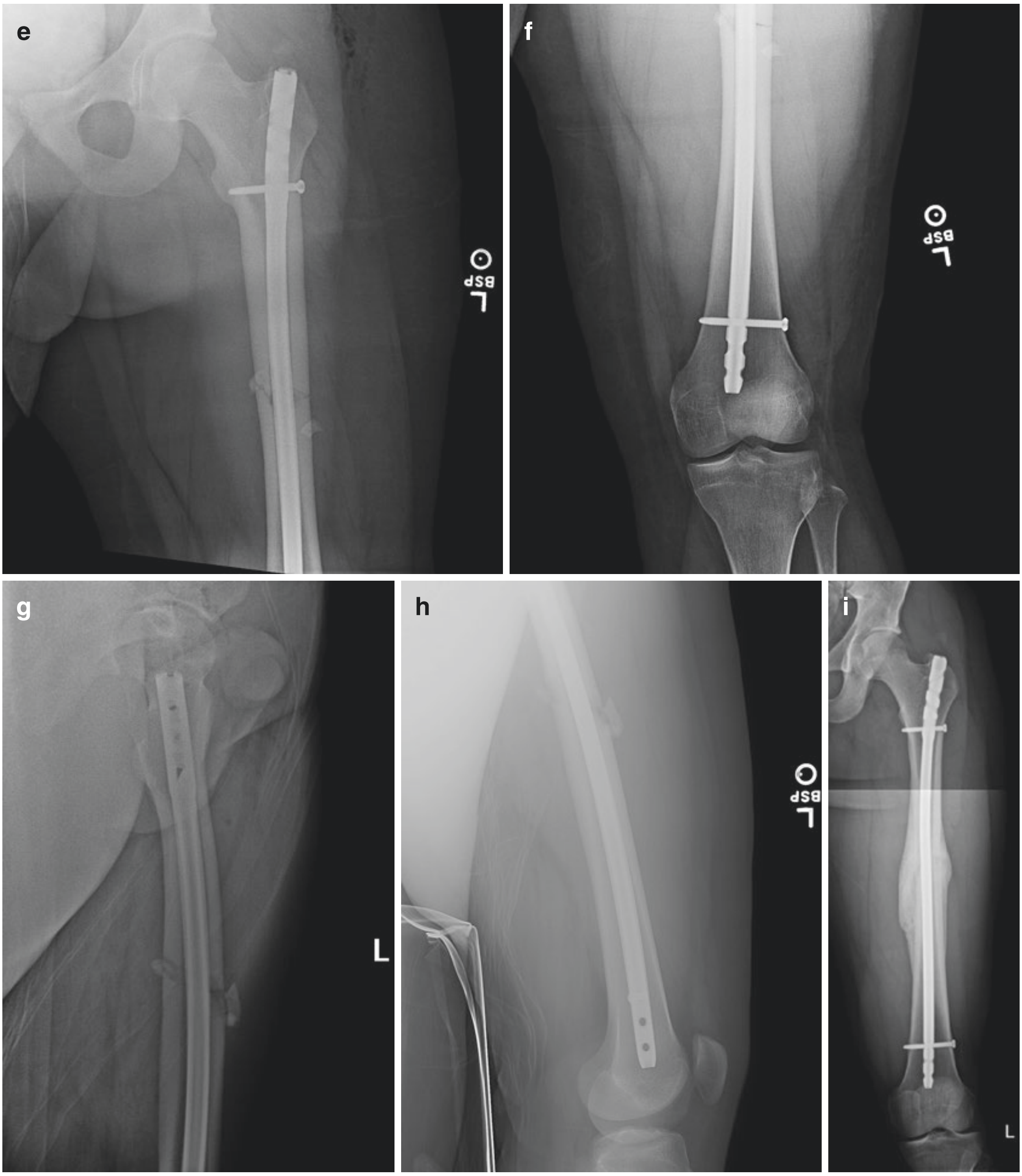

Fig. 16.1 (continued)

Case 2: Comminuted Distal Femoral Shaft Fracture (Fig. 16.2)

A 44-year-old male was involved in a motorcycle crash, during which he sustained multiple musculoskeletal injuries including an open left femoral shaft fracture. The fracture was highly comminuted. At the time of admission, and after appropriate resuscitation, he was placed into proximal tibial traction and was transported to the operating room for wound care and external fixation of his left lower extremity. He was returned to the operating room 36 hours later for repeat wound debridement, external fixator removal, and retrograde intramedullary nailing of his left femur fracture. Intraoperatively, the

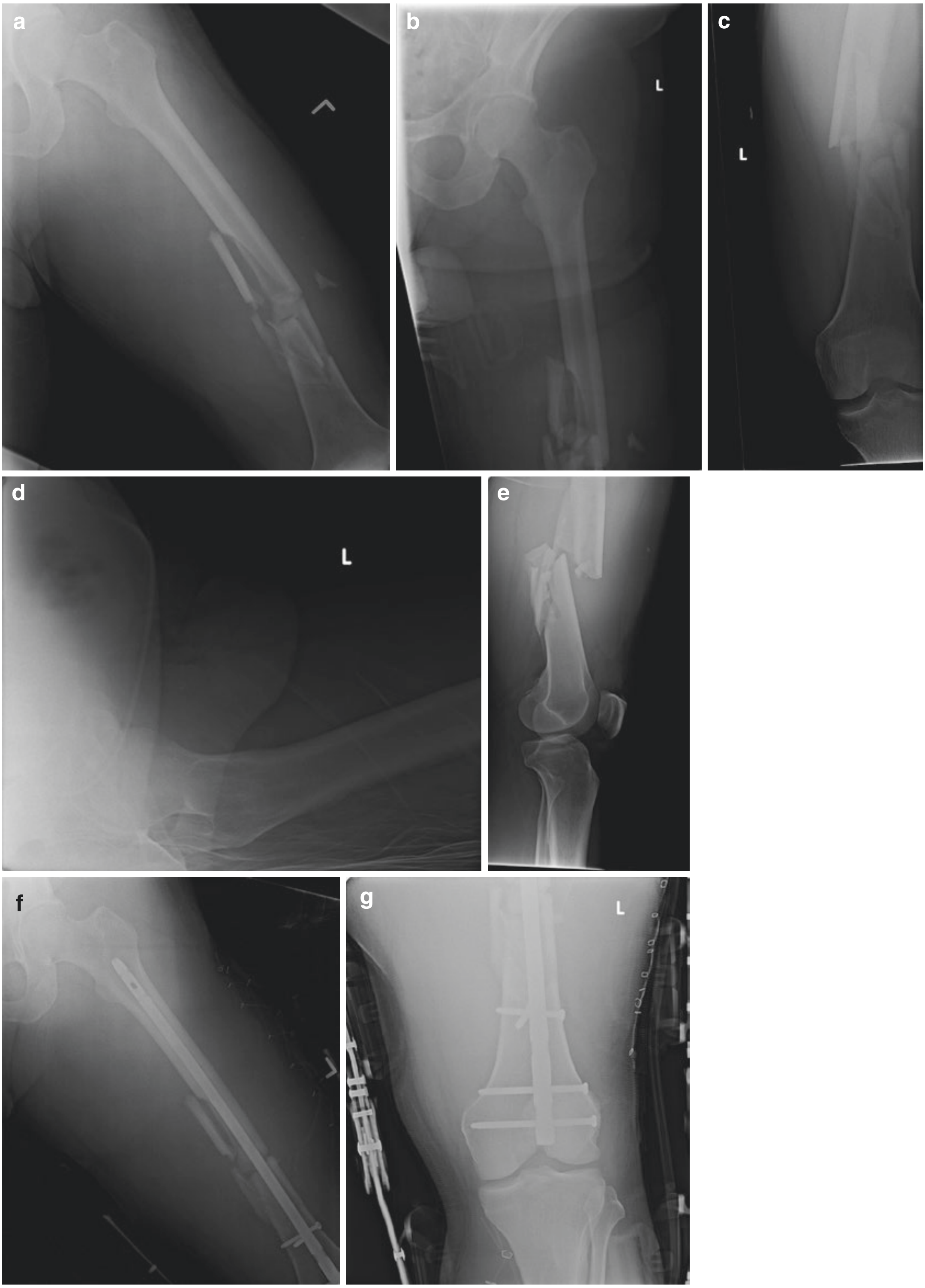

Fig. 16.2 Anteroposterior (**a**–**c**) and lateral (**d**, **e**) radiographs of a highly comminuted, open distal left femoral shaft fracture in a 44-year-old patient. Anteroposterior (**f**, **g**) and lateral (**h**, **i**) radiographs obtained postoperatively of the same patient's left femur after retrograde intramedullary nailing, using two distal blocking screws for supplemental fixation

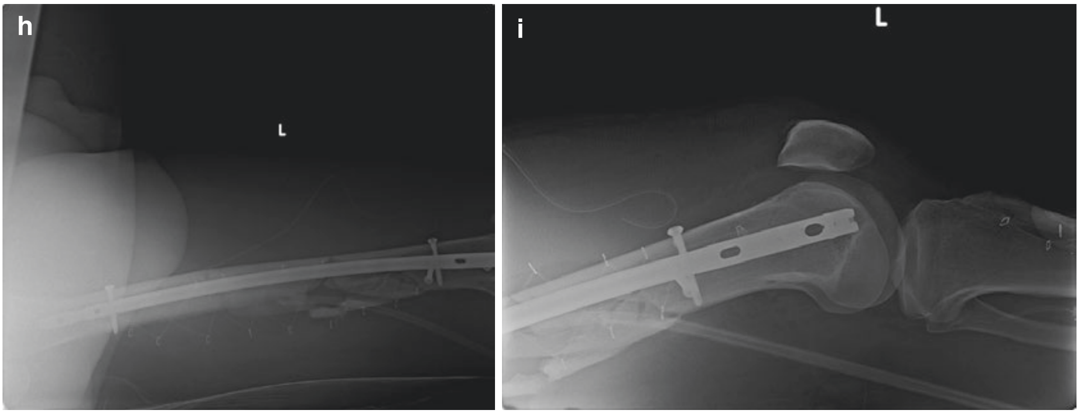

Fig. 16.2 (continued)

fracture was reduced with the assistance of sterile intraoperative traction and direct manipulation by rotating the leg and using a ball-spike pusher to maintain reduction during reaming and nailing. A reamed, retrograde intramedullary nail was placed. As the fracture was comminuted in the metaphysis, two blocking screws were placed into the distal segment, with one posterior to the nail (comminution extended very distal posteriorly) and with one medial to the nail (comminution extended very distal medially).

Why This Worked

A retrograde nail was chosen for fixation as the fracture extended very distal, and a greater "working length" of the nail would be present in the distal segment with retrograde nailing as compared to antegrade nailing (i.e., the nail will seat more distally if inserted retrograde, allowing for increased fixation length in the short distal segment). Owing to comminution which extended distally in the posteromedial aspect of the fracture, blocking screws were placed that would assist in preventing fracture deformity from weight-bearing or muscle pull. The blocking screws "substituted" for the "missing" cortices present at the site of maximal comminution. A single proximal interlocking screw was inserted, as the nail had good endosteal fit in the mid-diaphysis (at the isthmus), and anterolateral cortical contact was achieved during fracture reduction, allowing for increased axial stability after nailing.

Case 3: Comminuted Femoral Shaft Fracture with Long Lateral Butterfly Fragment (Fig. 16.3)

A 17-year-old female was involved in a motor vehicle crash, during which she sustained an isolated, closed right femoral shaft fracture. She was taken to the operating room, where she received an antegrade, reamed, trochanteric-entry intramedullary nail for treatment of her right femoral shaft fracture. During surgery, reduction was maintained with sterile skeletal traction. Owing to the large size of the lateral butterfly fragment, a single laterally based blocking screw was placed into the distal femoral segment adjacent to the reamed pathway, prior to nail insertion. Two proximal and two distal static interlocking screws were used in the setting of the long butterfly fragment, to provide additional coronal plane fracture stability.

Why This Worked

Due to the patient's young age, a trochanteric-entry nail was selected. This allows the surgeon to avoid the femoral head blood supply, which may be at risk with a piriformis fossa nail entry site in a patient recently having reached skeletal maturity. Static interlocking screws allow for maintenance of axial length and are supported by medial cortical contact. The laterally based blocking screw prevents a valgus deformity from ensuing during healing and weight-bearing by providing a "cortex" against which the nail will abut, preventing the valgus deformity. The actual lateral cortex is on the butterfly fragment, which is displaced.

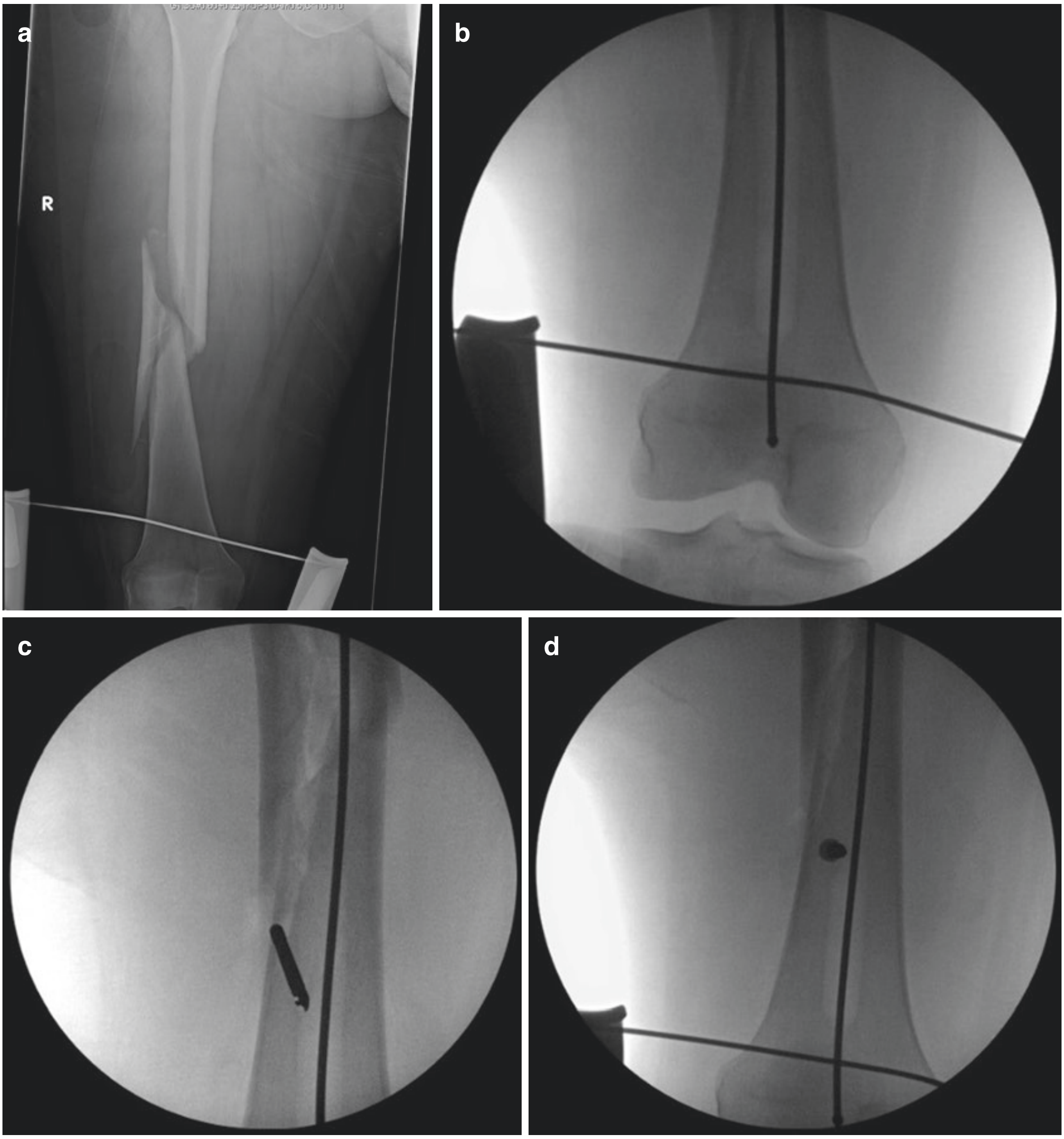

Fig. 16.3 Anteroposterior radiograph (**a**) of a comminuted right femoral shaft fracture in a 17-year-old patient, showing a large lateral butterfly fragment. Intraoperative fluoroscopic radiographs follow, demonstrating the intended nail path (**b**)—note the lucency about the ball-tipped reaming rod, drill placement for lateral blocking screw (**c**), blocking screw placement prior to nail placement (**d**), placement of nail past blocking screw (**e**), and final appearance of the fracture after completion of nail interlocking (**f**). Anteroposterior radiograph (**g**) of the same patient's right femur after completion of intramedullary nailing, using a single laterally based blocking screw for supplemental fixation. Anteroposterior radiograph (**h**) of the same patient's right femur 2 years after surgical repair, demonstrating complete healing

Fig. 16.3 (continued)

Case 4: Segmental Tibia Fracture, Including Proximal Metaphyseal Fracture Line (Fig. 16.4)

A 65-year-old male sustained an isolated, open, segmental right tibia fracture during a motorcycle crash. He was transported to the operating room shortly after arrival for debridement and definitive fixation of his right tibia fracture. The open wound was noted to be located at the distal tibial shaft fracture site. A direct reduction with provisional plate-and-screw fixation was performed at that location. Proximally, a posteromedial approach was used for exposure and reduction of the proximal quarter tibia fracture, which was difficult to control secondary to the small size of the proximal segment and secondary to the pull of the quadriceps through the patellar ligament. The fracture was reduced and

Fig. 16.4 Anteroposterior and lateral radiographs of the right tibia (**a**, **b**), right knee (**c**, **d**), and right ankle (**e**, **f**) in a 65-year-old patient with an open segmental right tibia fracture. Intraoperative fluoroscopic radiographs follow, demonstrating plate-and-screw fixation after reduction of the proximal quarter tibia fracture (**g**, **h**), direct reduction of the distal tibia diaphyseal fracture (**i**, **j**), provisional plate fixation of the distal tibia diaphyseal fracture (**k**), nailing guide wire starting point localization (**l**), and final appearance of right tibia after completion of intramedullary nailing and removal of the distal monocortical plate (**m–t**). Note that the proximal plate has been retained in place. Anteroposterior and lateral radiographs of the same tibia following completion of surgical right tibia fracture repair (**u**, **v**). Anteroposterior radiograph (**w**) of the same patient's right tibia 1 year after surgical repair, demonstrating complete healing

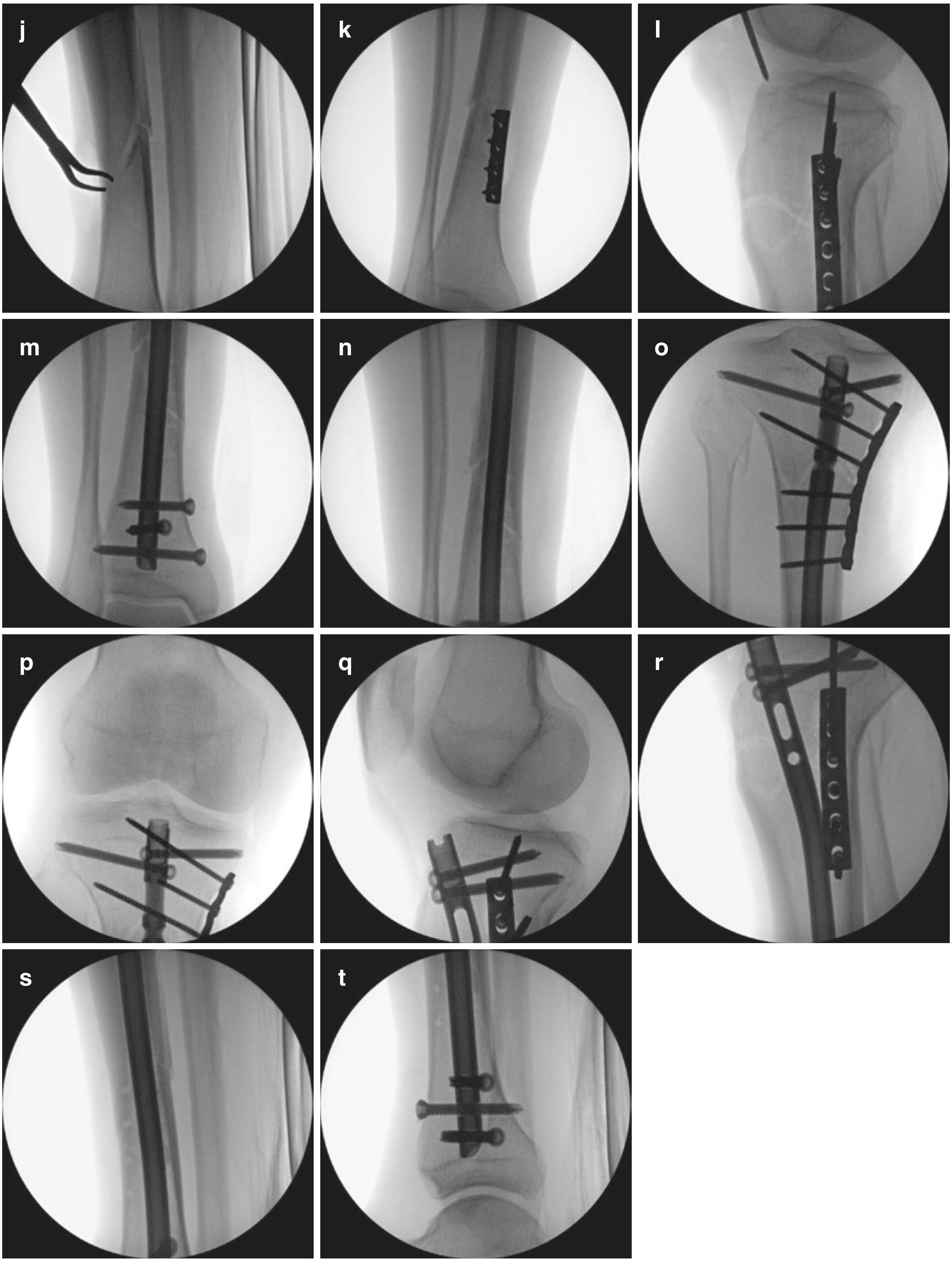

Fig. 16.4 (continued)

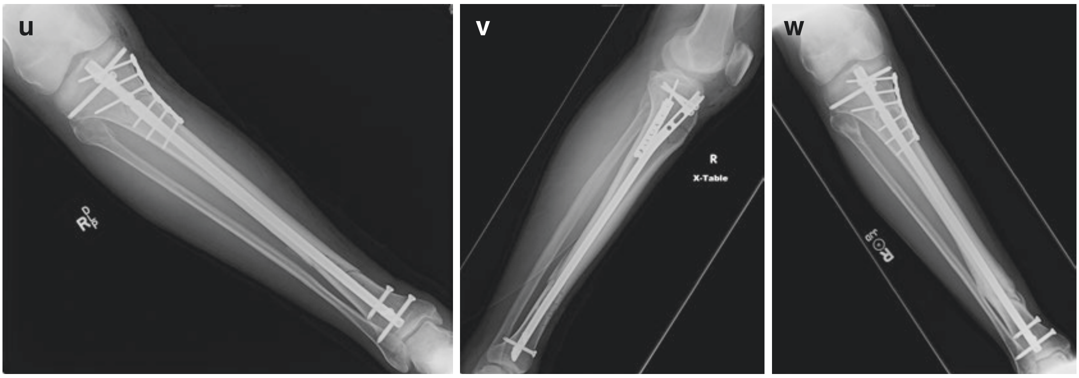

Fig. 16.4 (continued)

stabilized with a posteromedial plate-and-screw construct. After uncomplicated intramedullary nailing of the fracture, the distal provisional plate was removed, and the proximal plate was left in place. Multiple multiaxial proximal interlocking screws were utilized through the intramedullary nail for added stability of the construct.

Why This Worked

Obtaining and maintaining reduction of a proximal quarter tibia fracture may be difficult secondary to the small size of the proximal fragment and also secondary to the pull of the patellar ligament. A direct reduction of this fracture may be necessary. The plate-and-screw construct was left in place as supplemental fixation in a patient with potentially inferior bone quality (65 years old, metaphyseal fracture location). Use of multiple interlocking screws, placed through the nail in multiple directions, also increases the stability of the construct. Distally, the provisional monocortical plate was used to maintain reduction, but it was removed after nailing and interlocking. The distal fracture was diaphyseal and was easily maintained in a reduced position after nailing, and the plate would have been prominent if it had been left in place medially, potentially irritating the overlying soft tissues.

Case 5: Humerus Fracture Nonunion After Nailing (Fig. 16.5)

A 91-year-old woman presented to the orthopedic clinic complaining of upper left arm pain. She had sustained a proximal left humerus fracture approximately 11 months prior to presentation. She was treated with an intramedullary nail, with one locking blade placed proximally and one static interlocking screw placed distally. She failed to heal her fracture and was referred for evaluation and treatment recommendations. At time of presentation, she was noted to have minimal callus formation at the site of her fracture. The distal interlocking screw was noted to be loose, with an expansile lesion around the screw tract and bone formation around the screw head. Nonunion repair was recommended, and she elected to proceed. Nail removal was mildly difficult secondary to bone growth over the head of the distal interlocking screw. Repair was performed using lag screws and a long proximal humerus locked plating construct, intended to bypass the large cavitary bone defect left behind after removal of the distal interlocking screw.

Why This May Not Have Worked

Unlike lower extremity long bones, the humerus sees minimal axial loading forces but sees substantial torsional forces with daily activities. The long working length of the humeral nail reduces the torsional rigidity that might have been noted with screws placed closer to the fracture site, such as with a plate-and-screw construct. Also, fracture reduction was not optimal at the time of nailing. Finally, the small (and single) distal interlocking screw was all that was resisting the torsional loads put on the patient's humerus with regular activities. Perhaps, the nonunion could have been avoided with the use of more distal interlocking screws. The expansile bone lesion

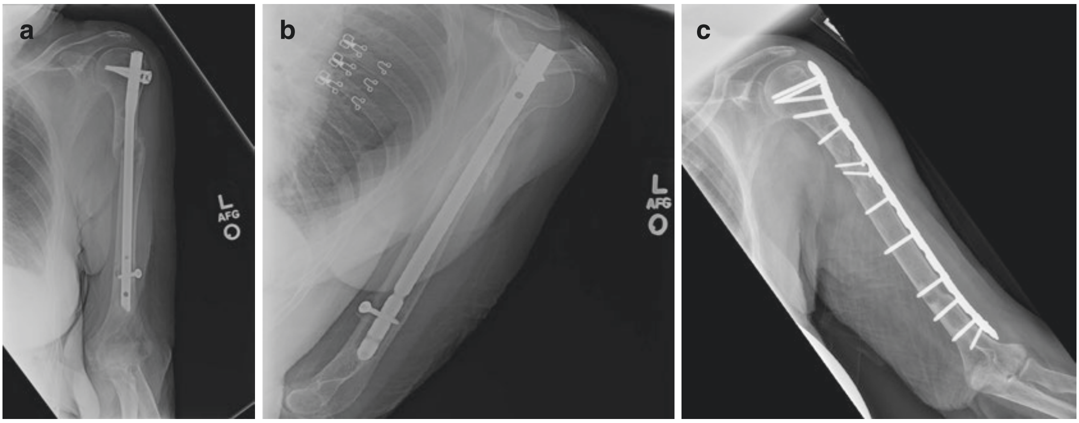

Fig. 16.5 Anteroposterior (**a**) and lateral (**b**) radiographs of the left humerus in a 91-year-old patient with a proximal humeral shaft fracture nonunion approximately 11 months post-intramedullary nailing of the fracture. Note the "soap bubble" appearance of loosening of the distal interlocking screw, with formation of callus about the head of the screw. Anteroposterior radiograph (**c**) of the same humerus a few months after nonunion repair, using compression technique with lag screws and neutralization plate

has been noted, in the authors' experience, primarily in humeral fracture nonunions (regardless of method of treatment) and not in the femur or tibia, ostensibly due to the extreme torsional forces that are transmitted through the humerus with activities of daily living (unlike the femur and tibia).

References

1. Küntscher G. The intramedullary nailing of fractures. Clin Orthop Relat Res. 1968;60:5–11.
2. University of Washington. Young's modulus. https://depts.washington.edu/matseed/mse_resources/Webpage/Biomaterials/young%27s_modulus.htm. Accessed 13 Oct 2019.
3. AZO Materials. Stainless Steel – Grade 316L – Properties, Fabrication and Applications (UNS S31603). 18 Feb 2004. Sydney, Australia. https://www.azom.com/article.aspx?ArticleID=2382. Accessed 13 Oct 2019.
4. Cuppone M, Seedhom BB, Berry E, Ostell AE. The longitudinal Young's modulus of cortical bone in the midshaft of human femur and its correlation with CT scanning data. Calcif Tissue Int. 2004;74(3):302–9.
5. AZO Materials. U.S.Titanium Industry Inc. Titanium Alloys – Ti6Al4V Grade 5. 30 Jul 2002. https://www.azom.com/properties.aspx?ArticleID=1547. Accessed 13 Oct 2019.
6. Kempf I, Grosse A, Beck G. Closed locked intramedullary nailing. Its application to comminuted fractures of the femur. J Bone Joint Surg Am. 1985;67(5):709–20.
7. Eichinger JK, Balog TP, Grassbaugh JA. Intramedullary fixation of clavicle fractures: anatomy, indications, advantages, and disadvantages. J Am Acad Orthop Surg. 2016;24(7):455–64.
8. Zhao L, Wang B, Bai X, Liu Z, Gao H, Li Y. Plate fixation versus intramedullary nailing for both-bone forearm fractures: a meta-analysis of randomized controlled trials and cohort studies. World J Surg. 2017;41(3):722–33.
9. Pace JL. Pediatric and adolescent forearm fractures: current controversies and treatment recommendations. J Am Acad Orthop Surg. 2016;24(11):780–8.
10. Bernstein M, Fragomen A, Rozbruch SR. Tibial bone transport over an intramedullary nail using cable and pulleys. JBJS Essent Surg Tech. 2018;8(1):e9.
11. Markolf KL, Cheung E, Joshi NB, Boguszewski DV, Petrigliano FA, McAllister DR. Plate versus intramedullary nail fixation of anterior tibial stress fractures: a biomechanical study. Am J Sports Med. 2016;44(6):590–6.
12. Märdian S, Schaser KD, Duda GN, Heyland M. Working length of locking plates determines interfragmentary movement in distal femur fractures under physiological loading. Clin Biomech (Bristol, Avon). 2015;30(4):391–6.
13. Ma JX, Wang J, Xu WG, Yu JT, Yang Y, Ma XL. Biomechanical outcome of proximal femoral nail antirotation is superior to proximal femoral locking compression plate for reverse oblique intertrochanteric fractures: a biomechanical study of intertrochanteric fractures. Acta Orthop Traumatol Turc. 2015;49(4):426–32.
14. Augat P, Rapp S, Claes L. A modified hip screw incorporating injected cement for the fixation of osteoporotic trochanteric fractures. J Orthop Trauma. 2002;16(5):311–6.

15. Ito K, Hungerbühler R, Wahl D, Grass R. Improved intramedullary nail interlocking in osteoporotic bone. J Orthop Trauma. 2001;15(3):192–6.
16. Tencer AF, Sherman MC, Johnson KD. Biomechanical factors affecting fracture stability and femoral bursting in closed intramedullary rod fixation of femur fractures. J Biomech Eng. 1985;107(2):104–11.
17. Egol KA, Chang EY, Cvitkovic J, Kummer FJ, Koval KJ. Mismatch of current intramedullary nails with the anterior bow of the femur. J Orthop Trauma. 2004;18(7):410–5.
18. Bazylewicz DB, Egol KA, Koval KJ. Cortical encroachment after cephalomedullary nailing of the proximal femur: evaluation of a more anatomic radius of curvature. J Orthop Trauma. 2013;27(6):303–7.
19. Roberts JW, Libet LA, Wolinsky PR. Who is in danger? Impingement and penetration of the anterior cortex of the distal femur during intramedullary nailing of proximal femur fractures: preoperatively measurable risk factors. J Trauma Acute Care Surg. 2012;73(1):249–54.
20. Collinge CA, Beltran CP. Does modern nail geometry affect positioning in the distal femur of elderly patients with hip fractures? A comparison of otherwise identical intramedullary nails with a 200 versus 150 cm radius of curvature. J Orthop Trauma. 2013;27(6):299–302.
21. Strauss E, Frank J, Lee J, Kummer FJ, Tejwani N. Helical blade versus sliding hip screw for treatment of unstable intertrochanteric hip fractures: a biomechanical evaluation. Injury. 2006;37(10):984–9.
22. Tejwani N, Polonet D, Wolinsky PR. Controversies in the intramedullary nailing of proximal and distal tibia fractures. J Am Acad Orthop Surg. 2014;22(10):665–73.
23. Mehling I, Hoehle P, Sternstein W, Blum J, Rommens PM. Nailing versus plating for comminuted fractures of the distal femur: a comparative biomechanical in vitro study of three implants. Eur J Trauma Emerg Surg. 2013;39(2):139–46.
24. Freeman AL, Craig MR, Schmidt AH. Biomechanical comparison of tibial nail stability in a proximal third fracture: do screw quantity and locked, interlocking screws make a difference? J Orthop Trauma. 2011;25(6):333–9.
25. Höntzsch D, Schaser KD, Hofmann GO, Pohlemann T, Hem ES, Rothenbach E, et al. Evaluation of the effectiveness of the angular stable locking system in patients with distal tibial fractures treated with intramedullary nailing: a multicenter randomized controlled trial. J Bone Joint Surg Am. 2014;96(22):1889–97.
26. Krettek C, Miclau T, Schandelmaier P, Stephan C, Möhlmann U, Tscherne H. The mechanical effect of blocking screws ("Poller screws") in stabilizing tibia fractures with short proximal or distal fragments after insertion of small-diameter intramedullary nails. J Orthop Trauma. 1999;13(8):550–3.
27. Krettek C, Stephan C, Schandelmaier P, Richter M, Pape HC, Miclau T. The use of Poller screws as blocking screws in stabilising tibial fractures treated with small diameter intramedullary nails. J Bone Joint Surg Br. 1999;81(6):963–8.

17 Diaphyseal Fractures

John D. Adams Jr. and Shea B. Ray

Introduction

Treatment of diaphyseal femoral and tibial fractures with intramedullary nail (IMN) fixation is currently the gold standard [1–4]. Compared to open reduction and internal fixation (ORIF) with a plate and screw construct, IMN fixation can involve less soft tissue dissection, potentially preserving the periosteum and the fracture hematoma [5]. Reduction followed by intramedullary stabilization leads to secondary bone healing [2]. While there are many factors involved in the eventual union of a diaphyseal long bone fracture that cannot be controlled (high rate of open injuries in the tibia, patient comorbidities and nutritional status, concomitant injuries, etc.), there are many factors within the surgeon's control that can be modified to obtain the best possible clinical outcome. Fracture union is dependent upon optimizing the biomechanical environment in which it is placed [6]. The biologic milieu is influenced by the quality of fracture reduction, presence of a fracture gap, and surrounding soft tissue environment. Surgical technique and implants are two of the primary factors that influence the mechanical environment. Specific to IMN fixation in diaphyseal fractures, the mechanics are impacted by nail diameter, reamed or unreamed technique, locking bolt configuration, and the type of bolts placed. Locking bolt configuration affects the working length. A nail's working length is the distance between the proximal locking bolt and the distal locking bolt. This is biomechanically important, because the torsional rigidity of the implant is inversely proportional to the working length, and the bending rigidity is inversely proportional to the square of the working length [7].

Implant-related factors such as material, nail length, and geometry of the nail also play a role. Each of these factors is within the surgeon's power to modify in order to create a biomechanically sound environment that is conducive to fracture healing.

Even with all factors optimized, nonunions and delayed unions still occur. Tibia fractures treated with IMNs have a 16.7% nonunion and delayed union rate [2], and femur fractures have a lower rate of nonunion that is typically less than 5–10% [3].

The following section reviews several clinical cases that illustrate the biomechanical principles of IMN in diaphyseal long bone fractures.

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Case 1

Background

A 19-year-old healthy male sustained a crush injury to his left femur (Fig. 17.1) after a trailer rolled over him. This was further complicated by thigh compartment syndrome.

Treatment

After fasciotomies, a reamed, antegrade femoral IMN was placed. The nail size was 10 × 420 millimeters (mm). The fracture was reduced with cortical contact, and the IMN had a good isthmic fit. Two locking bolts were placed proximally and distally (Fig. 17.2). Two days later, the patient underwent split thickness skin grafting of his fasciotomy site. After surgery, he was allowed to weight bear as tolerated. By the 8-week postoperative mark, he had bridging callus of all four cortices and experienced an uneventful recovery (Fig. 17.3).

Discussion

This case is an example of optimization of biomechanical factors leading to a satisfactory outcome. In general, the IMN has an adequate length and diameter to provide the appropriate

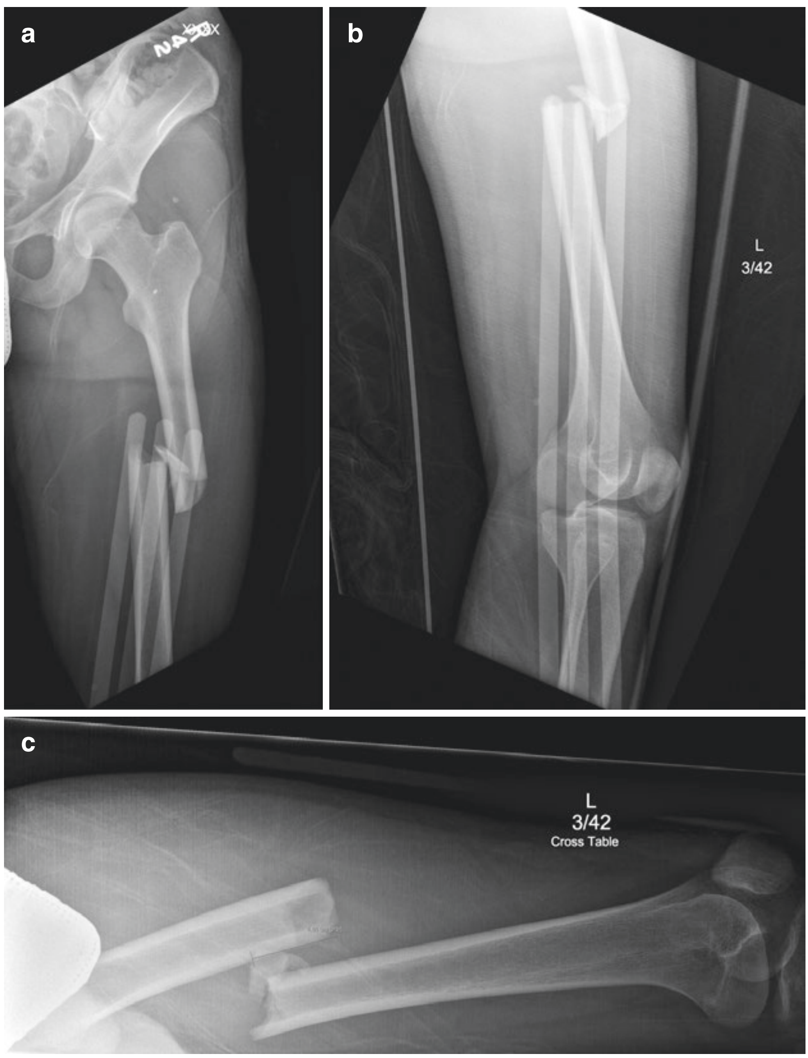


Fig. 17.1 Initial injury radiographs: (**a**) anteroposterior, (**b**) lateral, and (**c**) femur, showing a comminuted, oblique, midshaft, femur fracture

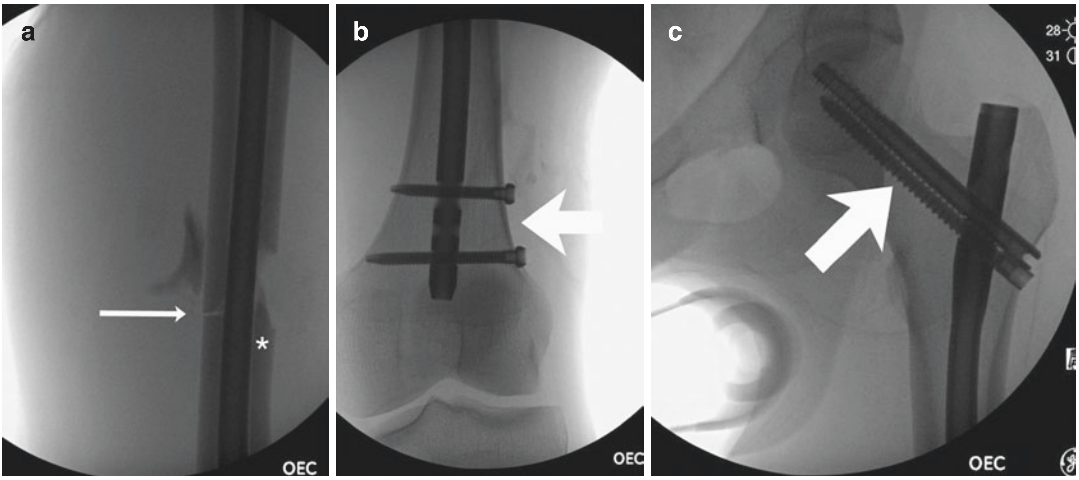


Fig. 17.2 Index procedure intraoperative fluoroscopy: (**a**) there is good isthmal fit (*asterisk*) and cortical contact (*thin arrows*) on the anteroposterior (AP) femoral fluoroscopic view; (**b**) AP knee view and (**c**) AP hip view showing the use of two distal interlocking screws placed proximally into the femoral head to decrease toggling of the bone around the nail at the fracture site (*thick arrows*)

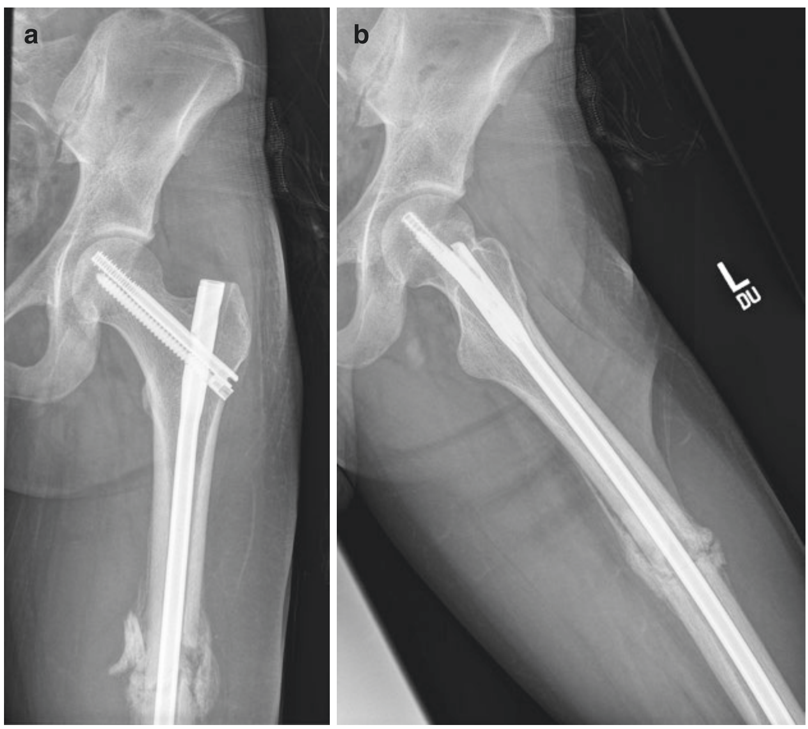


Fig. 17.3 Two-month postoperative radiographs: (**a**) anteroposterior (AP) and (**b**) lateral proximal femur, and (**c**) AP and (**d**) lateral distal femur. There is good secondary bone healing evident with bridging callus on all radiographs

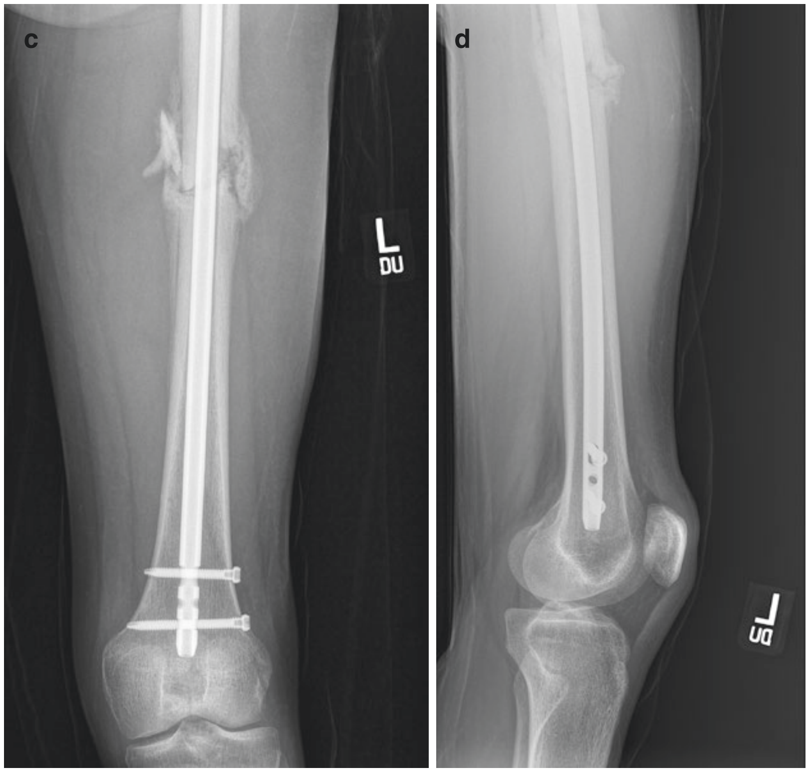


Fig. 17.3 (continued)

amount of stability and strain needed at the fracture site to induce secondary bone healing. As a general principle, fracture stability dictates the type of bone healing, with primary bone healing occurring when interfragmentary strain is <2% and secondary bone healing when the strain is less than 10% [8]. By definition, strain is

$$\varepsilon\left(strain\right)=\frac{\Delta l}{l_0}$$

where l is length and l_0 is the initial length. When applied to the biomechanical environment of an instrumented fracture, this value (reflected as a percentage) reflects the rigidity and stability of the construct. As a result, if the construct allows a lot of motion at the fracture site, the interfragmentary strain will be much higher than if a more rigid construct were applied. If the strain exceeds 10%, a nonunion will likely occur.

In this case, the nail diameter was large enough to obtain good isthmic fit after medullary reaming. The tightest spot within the femur is the isthmus. By enlarging the isthmus with reaming, the nail has a longer area of cortical contact with the endosteal bone and improves stability (Fig. 17.4). Reaming also allows for the ability to place a larger nail, which also increases biomechanical stability of the construct. Two bolts were used proximally and distally, which further contributed to stability at the fracture site with regard to rotation and axial stability. Fracture reduction with cortical contact provides the most construct stability and decreases the load seen by the implant. Additionally, this construct allows for early weight bearing through load sharing of the implant. By placing the nail down the anatomical axis of the femur, the force vector of load bearing through the femur is the same as that of load bearing through the implant, allowing for load sharing. Although a reconstruction locking option is not necessary to control this particular fracture, it was used to protect the femoral neck. There is some concern that future injury along with osteopenia near the top of the IMN may increase the risk of a future femoral neck fracture if the reconstruction screw option is not used.

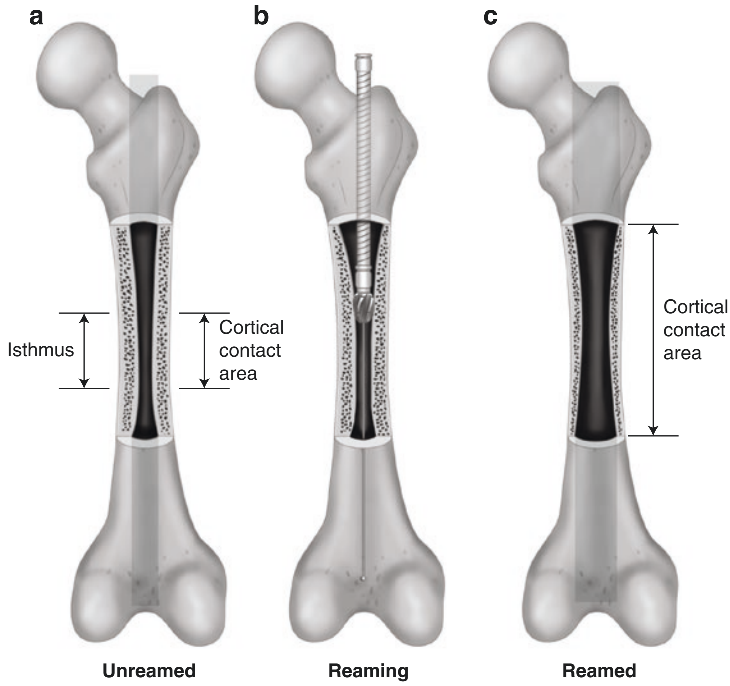


Fig. 17.4 Effect of reaming on cortical contact. Placing a nail following reaming (**b**) creates a greater amount of cortical contact (**c**), as compared to the amount of cortical contact area (**a**) at the level of the isthmus prior to reaming

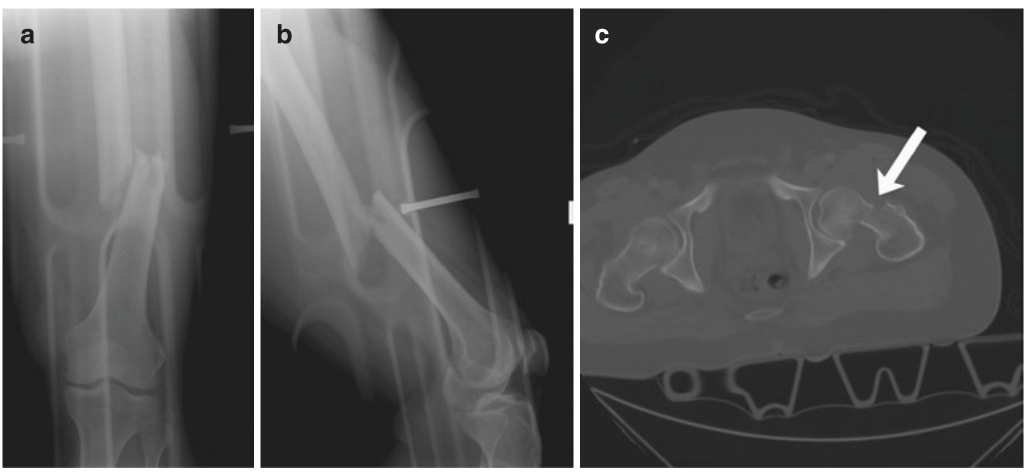


Fig. 17.5 Injury radiographs and representative computed tomography (CT) scan images. (**a**) Anteroposterior and (**b**) lateral radiographs of the femur show an oblique midshaft femur fracture; (**c**) axial cut computed tomography (CT) scan of the pelvis shows an ipsilateral femoral neck fracture (*arrow*)

Case 2

Background

A 39-year-old male with no past medical history was an unrestrained driver involved in a motor vehicle accident. He sustained a combined femoral neck and closed left femoral shaft fracture (Fig. 17.5).

Treatment

Reduction and fixation of the left femoral neck fracture with a two-hole side plate and compression screw was carried out, followed by retrograde IMN fixation of the femoral shaft fracture. The nail was locked with two distal interlocking screws and only one proximal locking bolt (Fig. 17.6).

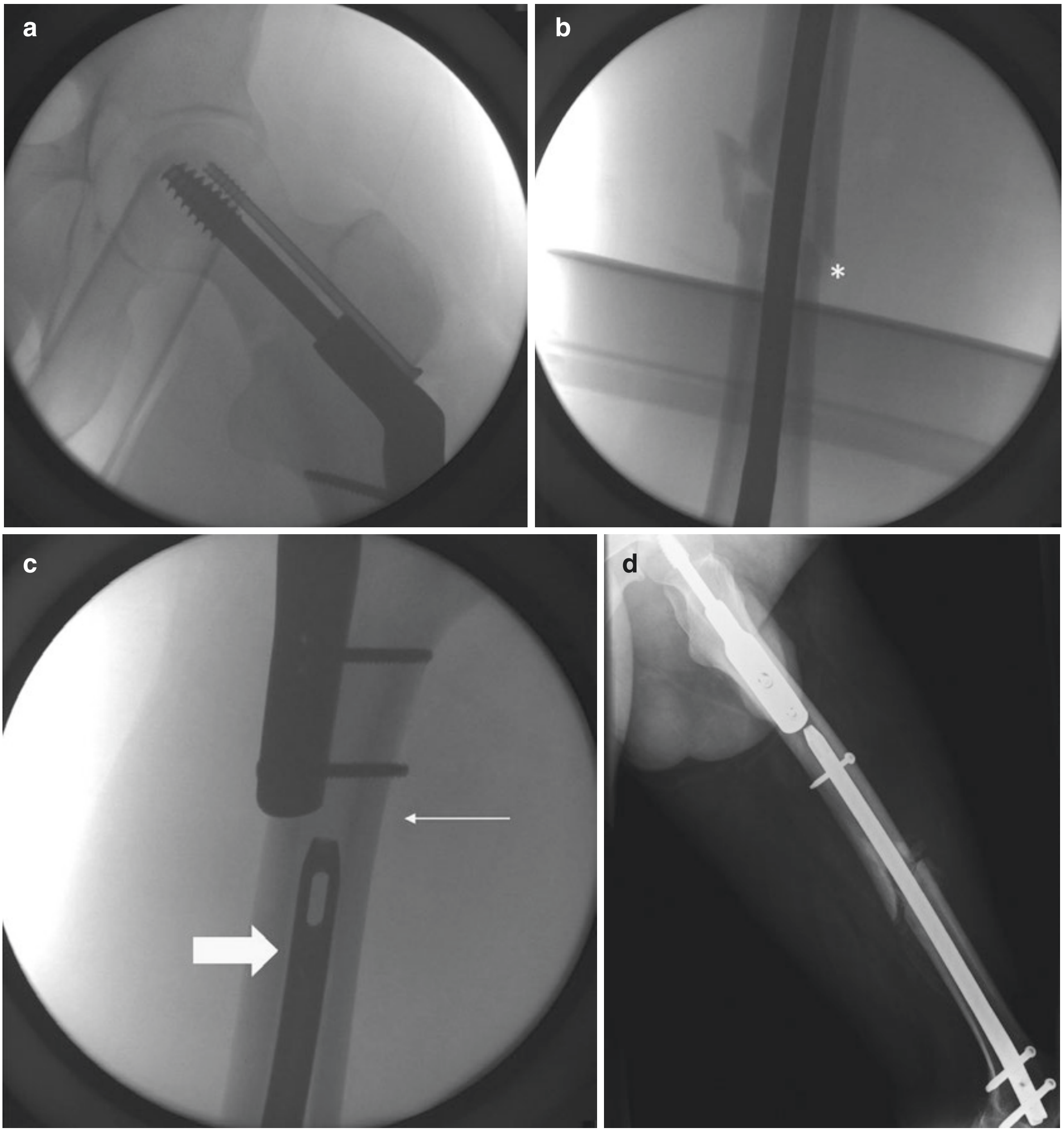


Fig. 17.6 Index procedure (**a**–**d**). (**a**) Anteroposterior (AP) hip fluoroscopic view. There are several factors here that ultimately led to a hypertrophic nonunion: an undersized nail (**b**) (AP femur view) with space between the nail and cortices (*asterisk*); (**c**) (AP proximal femur) and (**d**) (lateral femoral radiograph) short nail in the proximal segment (*thin arrow*) and placement of only one proximal interlocking screw (*thick arrow*)

The nail was also quite short, with a length of 280 mm. Radiographic follow-up showed that the femoral shaft fracture progressed to a hypertrophic nonunion (Figs. 17.7 and 17.8). Roughly 1 year after the index procedure, the patient returned for removal of the intramedullary implant and exchange nailing. The canal was reamed up to accommodate a larger 14-mm nail, and only one proximal screw placed (in the dynamic position) (Fig. 17.9). The patient nonunion persisted (Fig. 17.10) and required a second revision with open compression plating. This eventually led to union after fixation with absolute stability obtained with compression plating (Fig. 17.11).

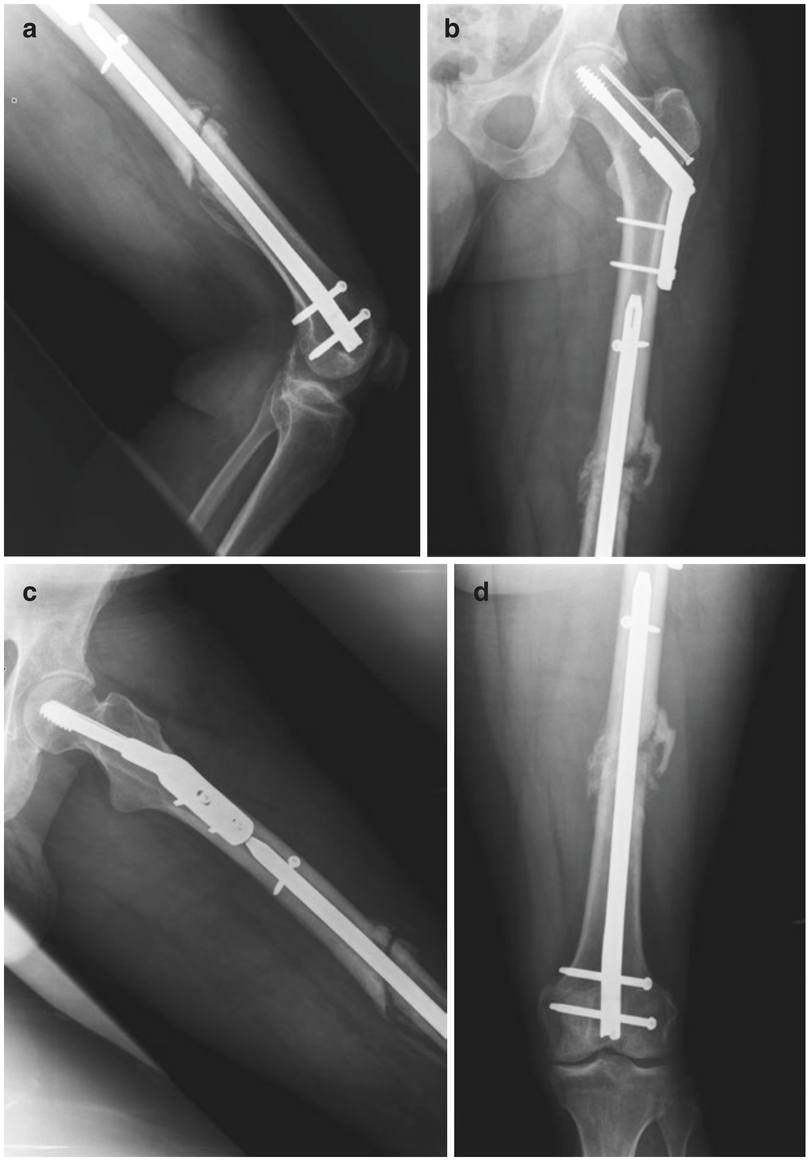


Fig. 17.7 Twelve-week postoperative radiographs show fracture callous starting to form. (**a**) Lateral distal femur, (**b**) anteroposterior (AP) proximal femur, (**c**) lateral proximal femur, (**d**) AP distal femur

Discussion

This patient had a recalcitrant hypertrophic nonunion of a femoral shaft fracture in the setting of a concomitant ipsilateral femoral neck fracture. The femoral neck fracture united without incident, and the patient's infectious and metabolic workups were negative, pointing to a mechanical cause for his femoral diaphyseal nonunion.

Several factors led to the hypertrophic nonunion. First, the initial nail was short in the proximal segment and quite small in diameter. The short length of the nail in the proximal segment decreased both the cortical contact and the working length of the nail in the proximal segment, leading to less fracture stability. Although the femur was reamed, the nail was undersized. The combination of the nail being undersized

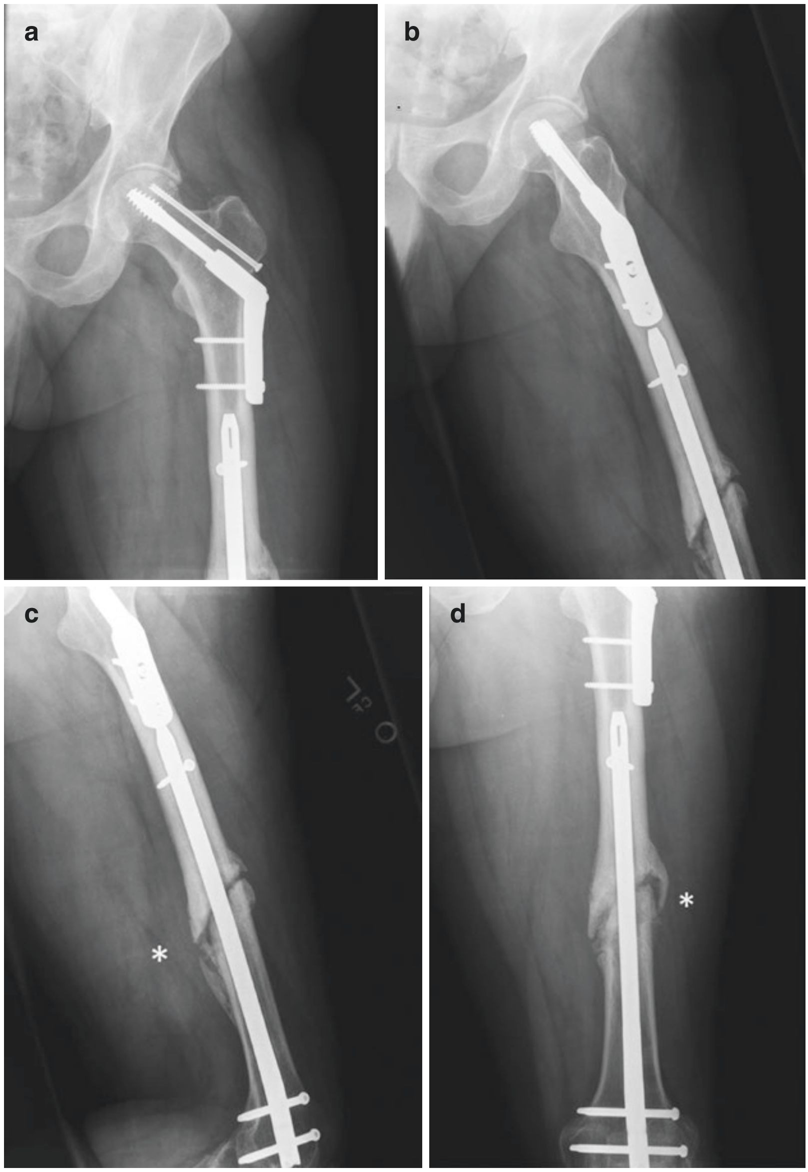


Fig. 17.8 Ten-month postoperative radiographs show hypertrophic femoral shaft nonunion. (**a**) Anteroposterior (AP) proximal femur, (**b**) lateral proximal femur, (**c**) lateral distal femur, and (**d**) AP distal femur—there is a hypertrophic nonunion present (*asterisk*)

and the short length of the nail in the proximal segment led to decreased stability (see Case 3, Fig. 17.14c). Second, only one locking bolt was placed proximally. This single locking bolt only provided stability in one plane. In this case, the proximal fragment could rotate around the one screw in the coronal plane. The amount of motion allowed in the coronal plane is directly related to the fit of the IMN against the cortex. For example, if there is 2 mm of space on each side of the nail relative to the cortex, the proximal fragment can move 2 mm in either direction. Two things can be done to eliminate this coronal plane motion—narrow the distance between the nail and the cortex (select a larger nail) and place two proximal locking bolts (Fig. 17.12). In this case, the short, undersized nail with only one proximal locking bolt led to increased motion and, subsequently, a hypertrophic nonunion.

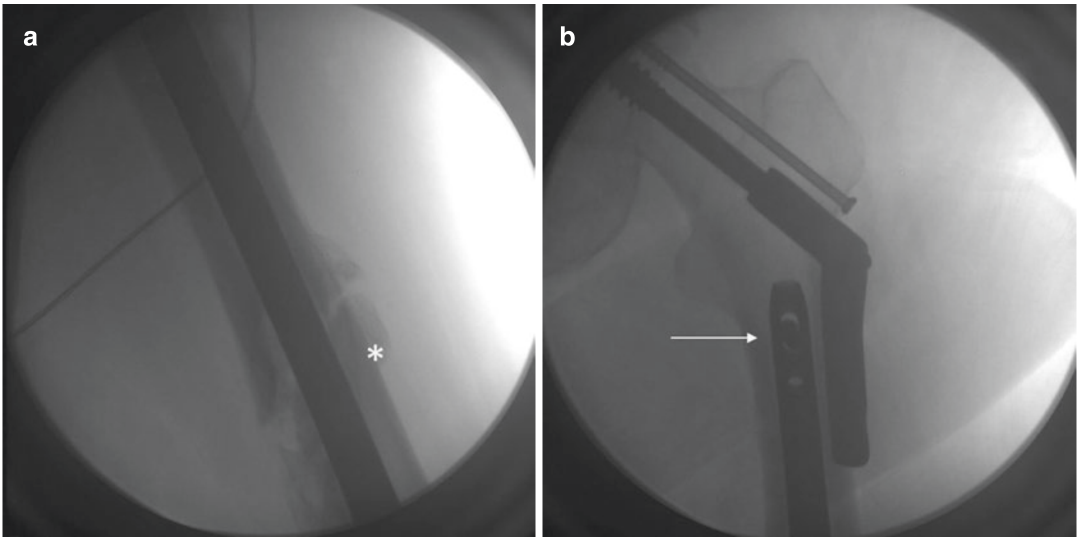
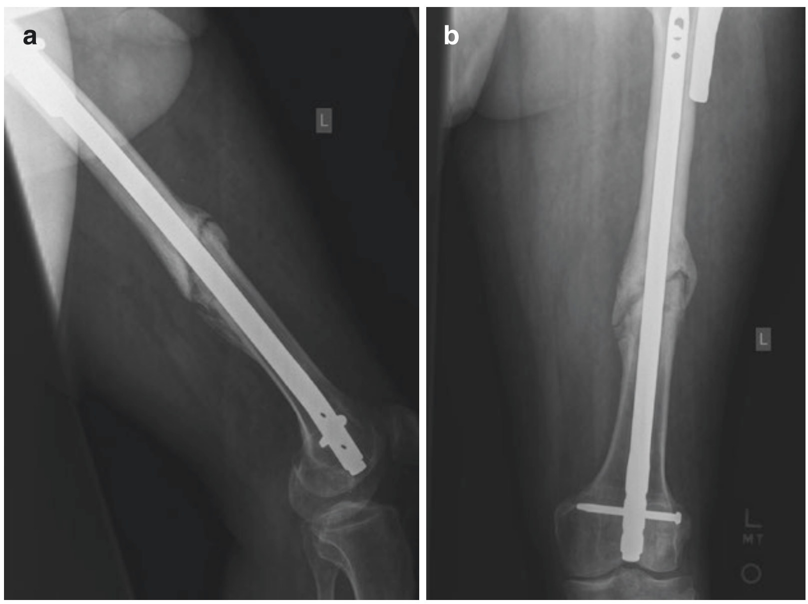


Fig. 17.9 First femur fracture nonunion surgery. (**a**) Lateral femur fluoroscopic view: in this case, the nail diameter was upsized, allowing for more cortical contact (*asterisk*). (**b**) Anteroposterior hip view: note that the proximal interlocking screw (*arrow*) was placed in the dynamic position, allowing for more motion at the fracture site rather than providing additional stability

Fig. 17.10 Five-month post-nonunion repair radiographs. A persistent hypertrophic nonunion is present on the (**a**) lateral and (**b**) anteroposterior femur



Unfortunately, the revision was also not well planned. In most circumstances, a hypertrophic nonunion is treated with more stability. In this case, the nail needed to be increased in length and diameter with two statically locked proximal bolts. Unfortunately, only one of the three things was done. The diameter of the nail was increased, which was good, but the nail was not lengthened, and again only one bolt was placed proximally. Also, the bolt was placed in the dynamic position. The goal for dynamically locking the nail is to allow more fracture motion to induce cortical contact and healing. However, this patient needed increased stability, not dynamization.

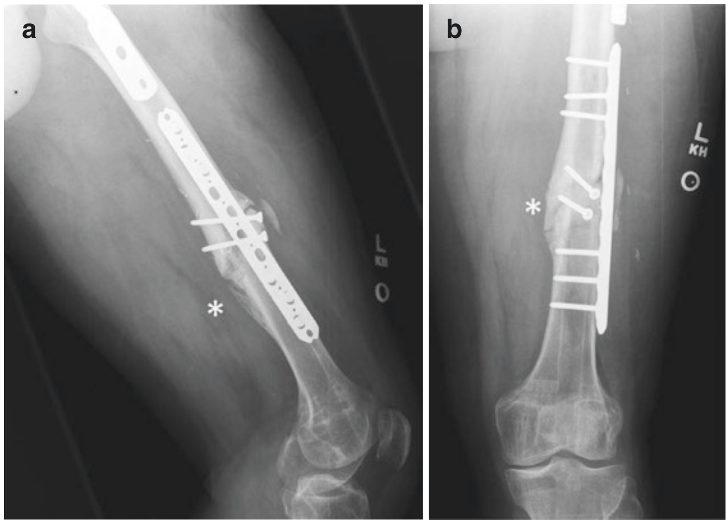


Fig. 17.11 (**a**) Lateral and (**b**) anteroposterior femur radiographs. Fracture union was achieved (*asterisks*) after a second nonunion repair to achieve absolute stability with the use of lag screws for fracture compression and a neutralization plate

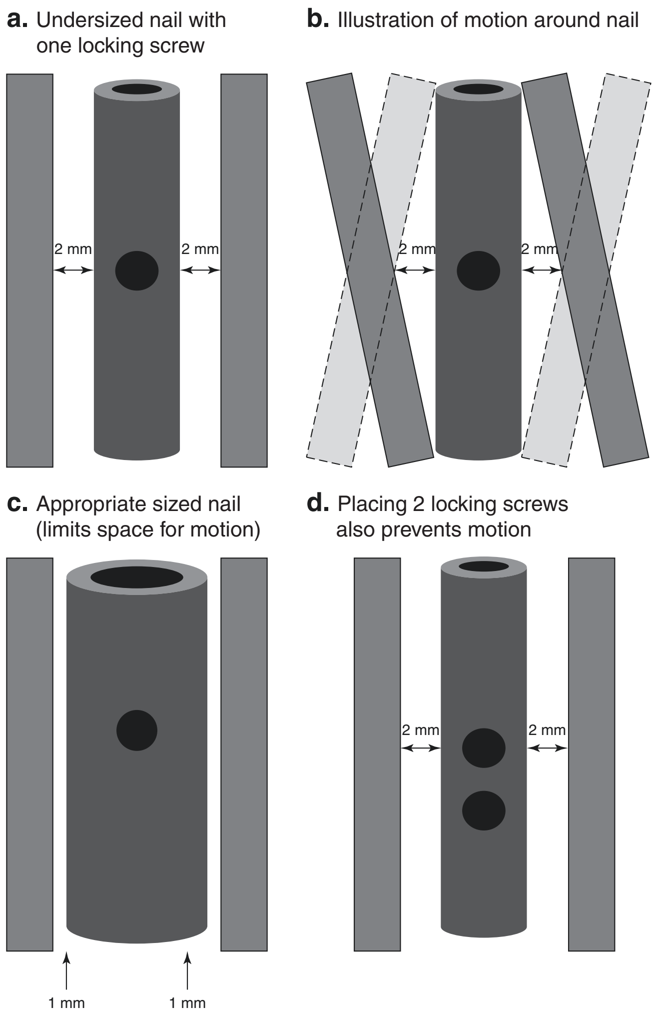


Fig. 17.12 Illustration of the effect that interlocking screw configuration and nail size have on construct stability. If there is a 2-mm gap between the nail and the cortex on each side, one single interlocking screw will allow for motion perpendicular to the orientation of the screw (2 mm on either side of the nail) (**a**, **b**). This can be remedied by placing a larger nail (**c**) to reduce the distance between the nail and the cortex, or by placing a second interlocking screw that reduces fracture motion (**d**)

Unfortunately, the revision nailing did not heal. This was not surprising, as dynamization created less instead of more stability.

Finally, one additional biomechanical area of concern in this case was the lack of overlap of the side plate and the nail. This space of unshielded bone in between two areas of increased stiffness secondary to implant presence acts as a stress riser that could increase the risk of inter-implant fracture. Ideally, these two constructs should overlap, eliminating the stress riser and the subsequent risk of fracture at this site.

Case 3

Background

This patient was a previously healthy 24-year-old male who was involved in a pedestrian versus motor vehicle collision. He was initially evaluated and provisionally treated at an outside facility. He was found to have isolated orthopedic injuries, which included a closed right femoral shaft fracture (Fig. 17.13) and a right sacroiliac joint disruption. He was placed into skeletal traction prior to transfer.

Treatment

Retrograde IMN of the right femur was performed with a 10 × 380 mm nail, with placement of two proximal and one distal static locking bolt (Fig.17.14). He underwent sacroiliac screw fixation for his pelvic ring injury in the same setting and was kept non-weight-bearing on the right lower extremity postoperatively due to his pelvic ring injury. He was allowed to weight bear on the right lower extremity 3 months postoperatively. Six months after surgery, his right femur fracture was considered a delayed union (Fig. 17.15). In preoperative discussion with the patient regarding exchange nailing versus a less invasive procedure with dynamization, he elected to pursue dynamization of the nail with removal of the proximal locking bolts (Fig. 17.16). Despite these interventions, he went on to a recalcitrant hypertrophic nonunion, with slight varus positioning at the fracture site (Fig. 17.17). Roughly 18 months after the dynamization procedure, he underwent removal of hardware and antegrade exchange nailing with a piriformis entry reconstruction nail (Fig. 17.18). His alignment was corrected and an antegrade 13 × 400 mm nail was placed, with two proximal screws, including femoral head fixation, and two distal static locking bolts.

Discussion

Similar to the previous case, the retrograde nailing technique led to a decreased working length proximally and increased motion at the fracture site. Having a shorter segment of nail on one side of the fracture should always raise concerns for the development of a hypertrophic nonunion. If retrograde nailing is performed on a fracture that is proximal to the isthmus, it is extremely important to increase the nail diameter to provide more cortical contact in the proximal fragment. Once a delayed hypertrophic nonunion was diagnosed, dynamization of the nail should have been avoided, because it decreased stability and led to excess motion and continued hypertrophic nonunion. The eventual placement of a reamed antegrade nail solved many of the biomechanical issues associated with the first procedure that led to nonunion. The patient went on to heal the fracture nonunion because the overall construct stability improved. This was accomplished with the antegrade nail by increasing control of the proximal fragment with increasing the nail working length and fixation into the proximal femur and increasing cortical contact by using a larger diameter nail.

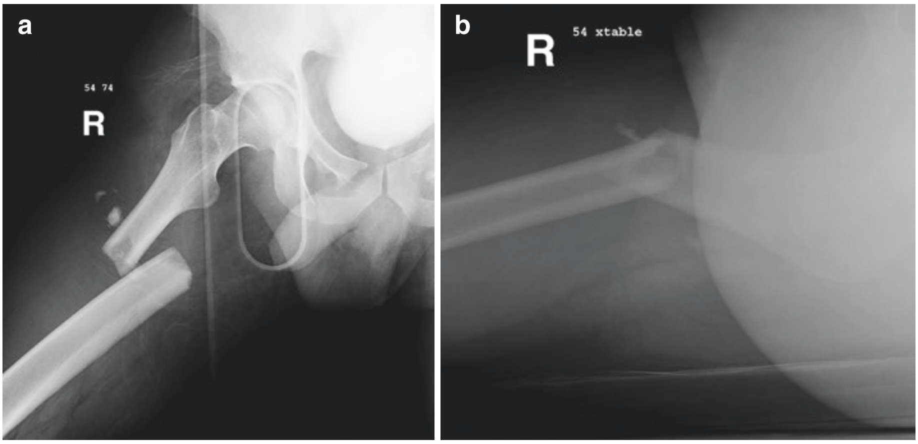


Fig. 17.13 Injury radiographs showing a transverse subtrochanteric femur fracture: (**a**) attempted anteroposterior and (**b**) lateral femur views

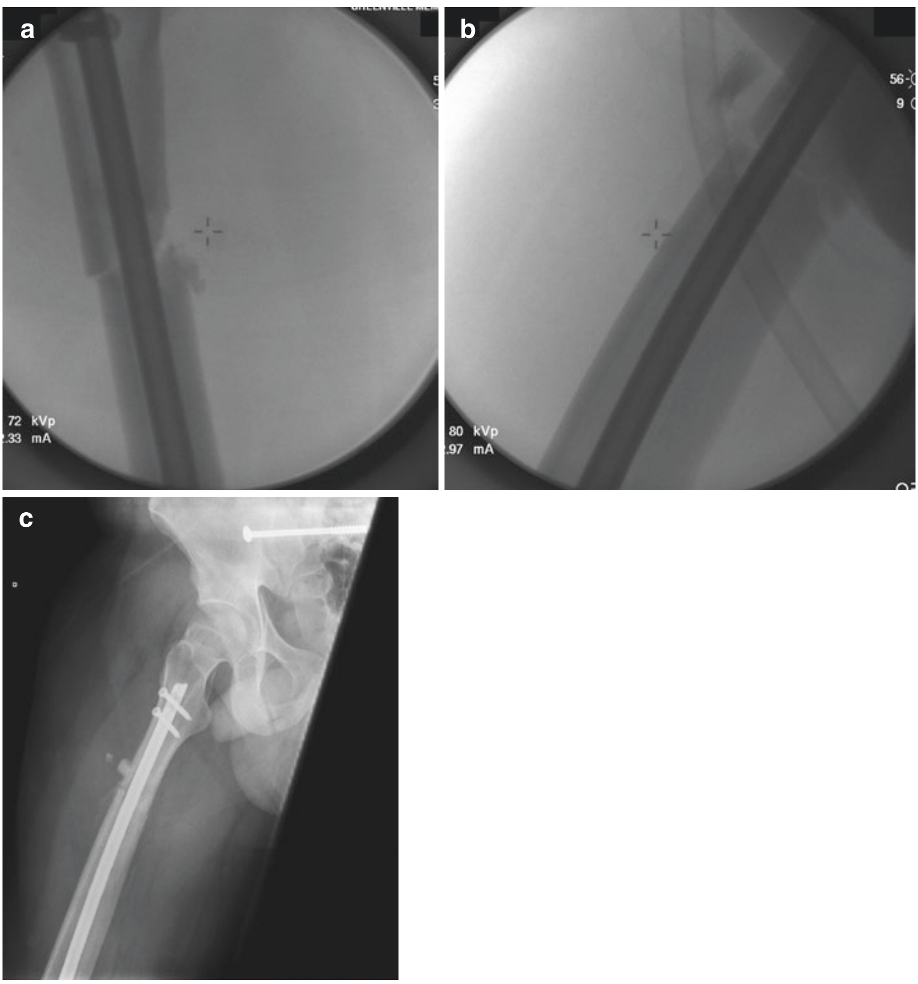


Fig. 17.14 (**a**) Anteroposterior and (**b**) lateral femur fluoroscopic views: an undersized retrograde femoral medullary nail was placed to address this subtrochanteric femur fracture with a relatively short working length due to the limited amount of nail in the proximal femur. (**c**) Postoperative lateral femoral radiograph

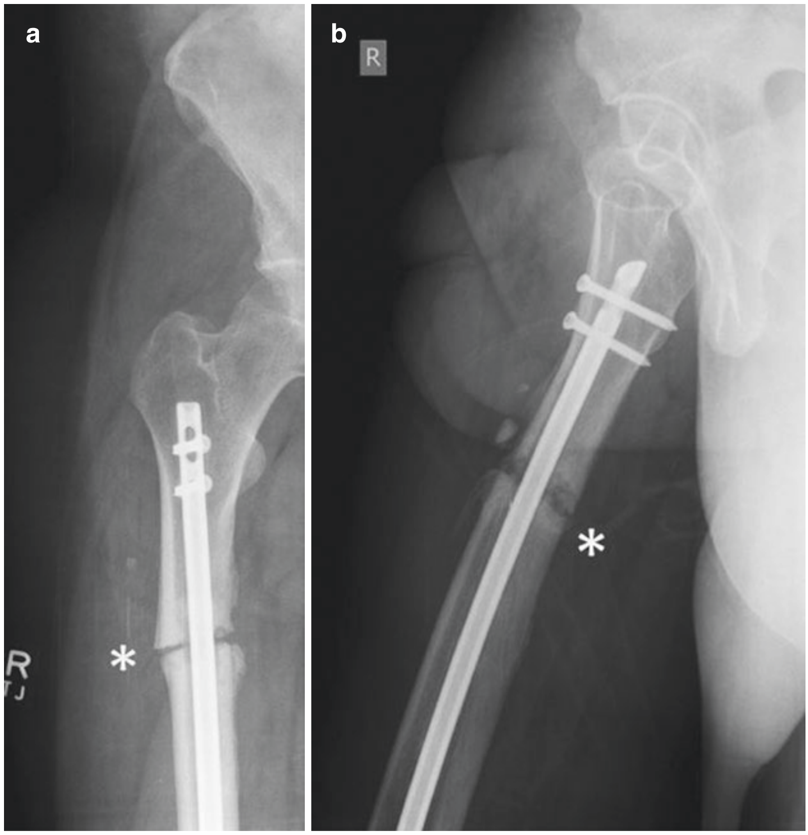


Fig. 17.15 Five-month postoperative anteroposterior (**a**) and lateral (**b**) femur radiographs showing a delayed atrophic nonunion (*asterisks*)

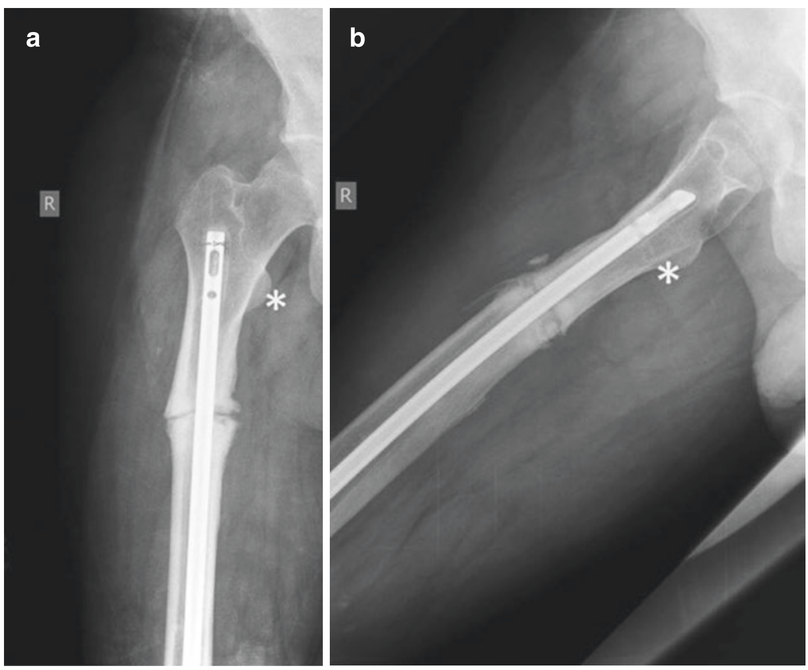


Fig. 17.16 (**a**) Anteroposterior and (**b**) lateral femur radiographs 1 month after the nail is dynamized by removing the proximal locking bolts (*asterisks*)

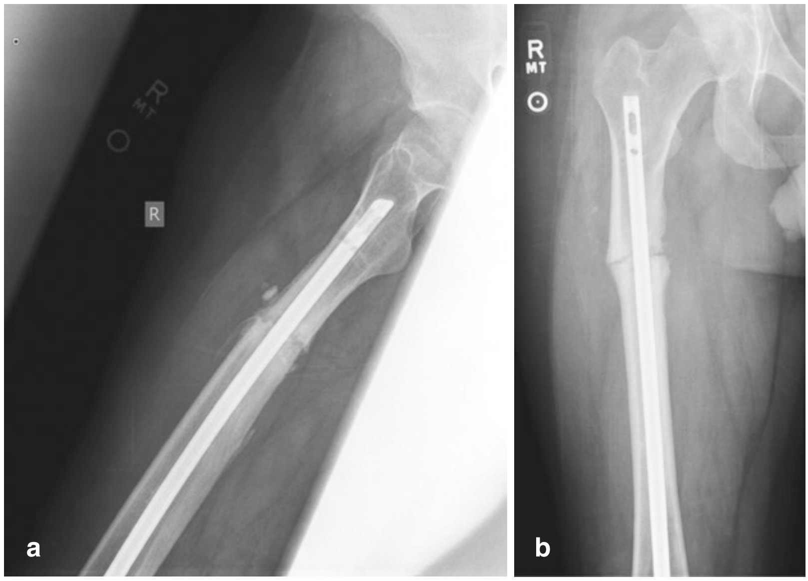


Fig. 17.17 (**a**) Lateral and (**b**) anteroposterior femur radiographs several months after the dynamization procedure indicating a persistent nonunion

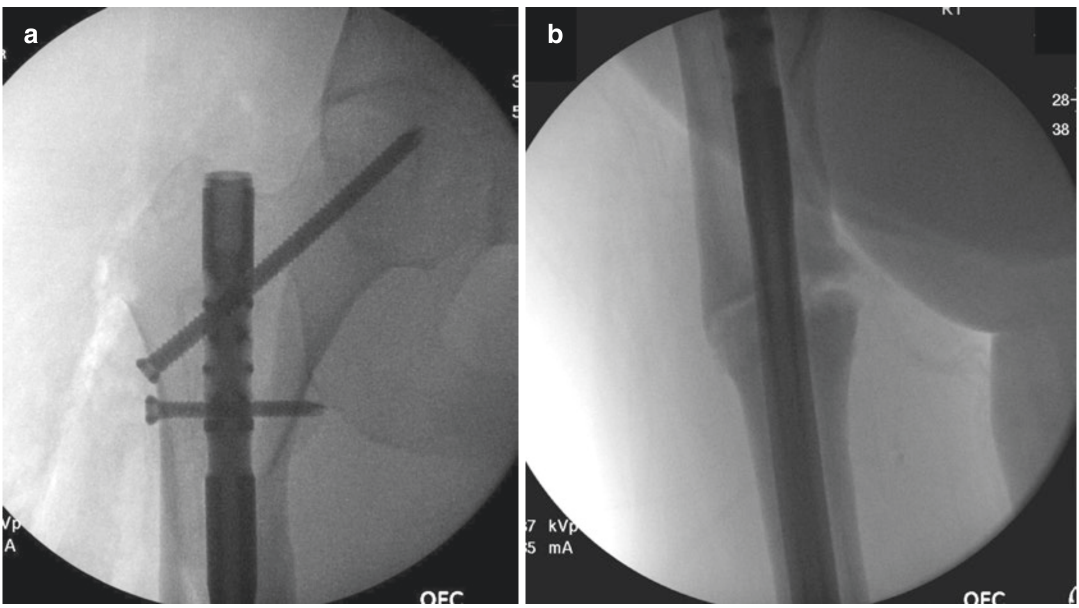


Fig. 17.18 (**a**) Intraoperative anteroposterior hip and (**b**) femur fluoroscopic views at the time of the antegrade reconstruction piriformis entry exchange femoral nailing

References

1. Samiezadeh S, Avval PT, Fawaz Z, Bougherara H. Biomechanical assessment of composite versus metallic intramedullary nailing system in femoral shaft fractures: a finite element study. Clin Biomech. 2014;29(7):803–10.
2. Hierholzer C, Friederichs J, Glowalla C, Woltmann A, Bühren V, von Rüden C. Reamed intramedullary exchange nailing in the operative treatment of aseptic tibial shaft nonunion. Int Orthop. 2017;41(8):1647–53.
3. Ricci WM, Gallagher B, Haidukewych GJ. Intramedullary nailing of femoral shaft fractures: current concepts. J Am Acad Orthop Surg. 2009;17(5):296–305.

4. Shih KS, Hsu CC, Hsu TP. A biomechanical investigation of the effects of static fixation and dynamization after interlocking femoral nailing: a finite element study. J Trauma Acute Care Surg. 2012;72(2):E46–53.
5. Wähnert D, Gehweiler D. Complications of intramedullary nailing—evolution of treatment. Injury. 2017;48(2017):S59–63.
6. Rosa N, Marta M, Vaz M, Tavares SM, Simoes R, Magalhães FD, Marques AT. Recent developments on intramedullary nailing: a biomechanical perspective. Ann N Y Acad Sci. 2017;1408(1):20–31.
7. Kyle RF. Biomechanics of intramedullary fracture fixation. Orthopedics. 1985;8(11):1356–9.
8. Perren SM. Fracture healing: fracture healing understood as the result of a fascinating cascade of physical and biological interactions. Part I. an attempt to integrate observations from 30 years AO research. Acta Chir Orthop Traumatol Cechoslov. 2014;81(6):355–64.

Periarticular and Intra-articular Fractures

18

Seong-Eun Byun, Michael Maher, Jihyo Hwang, Joshua A. Parry, and Cyril Mauffrey

Introduction

Intramedullary (IM) fixation is widely used as a reliable option for long bone fractures because it preserves periosteal blood supply [1, 2] and facilitates early weight-bearing with its load-sharing function [3]. However, anatomical characteristics of the periarticular region present complicating factors for treatment of peri- and intra-articular fractures compared to diaphyseal fractures when IM nailing (IMN) is used. The wider medullary canal at the level of the metaphysis, deforming forces from soft tissue attachments, and complex geometry around joints make it difficult to obtain satisfactory reduction and fixation solely using IMN techniques [4]. Even if fracture reduction is successful, the thin nature of periarticular cortical bone and limited fixation in commonly short articular segments make IM fixation biomechanically disadvantageous when used in isolation.

Despite inherent limitations of IM fixation, its use has increased in peri- and intra-articular fractures with improvements in technology due to its soft tissue sparing nature and the ability to withstand implant fatigue. Increased options in size, number, and orientation of interlocking bolts, as well as fixed angle designs, provide greater versatility and strength of IM constructs. Supplemental surgical techniques like poller/blocking or interfragmentary screws and provisional or supplemental plating improve the results of IM fixation.

However, these improvements cannot overcome all of the limitations of IM techniques in the peri- and intra-articular fractures. The basic principles of internal fixation, such as preserving blood supply, restoring alignment, and preventing excessive fracture gap, should be emphasized to obtain good results. In this chapter, the function of IM nails in peri- and intra-articular fractures—when they work and when they don't—will be discussed using clinical cases.

Key Concepts for Intramedullary Nailing in Peri- and Intra-articular Fracture

Advantages and Disadvantages

IM fixation can be advantageous for several reasons. The periosteal blood supply is often preserved with IMN since significant exposure of the fracture site is often not necessary [1, 2]. IMN

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can be load sharing, since they are close to the center of rotation and limit the axial bending moment arm. Therefore, IM fixation distributes axial forces evenly and facilitates early weight-bearing [3]. In addition, a smaller torsional moment arm leads to higher resistance to torsional loading compared with an extramedullary device [5]. However, these advantages are often negated in the treatment of peri- and intra-articular fractures that require direct visualization and reduction of the articular surface, when involved, and variable biomechanical properties due to limited fixation options.

IM fixation has several disadvantages in the setting of peri- and intra-articular fractures. The increased diameter of the metaphyseal space and cancellous bone limits mechanical contact between nail and bone in the typically short segments of peri- and intra-articular fractures. A wide canal contributes to malalignment which is important for stability, healing, and functional outcome (Figs. 18.1 and 18.2). Furthermore, the working length of the periarticular fragment is short, and the diaphyseal segment is long. This difference of working length between two fragments can cause a discrepancy of micromotion of each fragment and result in a negative effect on union. In the periarticular segment, the role of interlocking bolts is more important for stability, but it is difficult to gain sufficient purchase of the thin cortical bone in metaphysis (Fig. 18.3).

Additionally, the deforming forces around a joint are strong. Therefore, restoration of alignment is more difficult with closed or indirect (minimally invasive) reduction techniques in peri- and intra-articular fracture when compared to diaphyseal fractures [6].

Reduction

The main goals for treating an intra-articular fracture are to achieve anatomical reduction and absolute stability of the articular surface. These goals do not change when IMN is used in the treatment of articular fractures. During open reduction and internal fixation (ORIF) of intra-articular fractures, care should be taken to minimize the damage to the periosteal blood supply of metaphysis. If care is not taken, secondary bone healing of the metaphysis with the IMN will not proceed. Non-displaced articular fractures can be fixed with screws using minimally invasive technique before IM nailing as long as the articular surface is verified to be anatomically reduced (Fig. 18.4).

Precise entry point is crucial for obtaining correct alignment in IM fixation in peri- and intra-articular fractures [7, 8]. An improper entry point exaggerates fracture deformity and decreases stability. For example, the proximal tibial fracture is prone to valgus and procurvatum deformities, so the entry point should be located slightly laterally and posteriorly compared to the entry point for treatment of a diaphyseal fracture [4, 9].

In addition to the entry point, alternative positioning of the extremity significantly affects the deforming force of the muscles and tendons. For example, when using a semi-extended rather than a flexed position of the knee during IMN of the tibia, a reduction may be obtained with less difficulty in fractures of the proximal and distal tibia [10] by neutralizing the effect of the deforming forces of the patellar tendon/extensor mechanism on the fracture site.

Regarding reduction technique, indirect manipulation is often sufficient for reduction of the extra-articular fracture component. For example, for a spiral-type metaphyseal fracture, percutaneous clamping with fracture-reducing forceps or wiring can be attempted with minimal disruption to soft tissues (see Fig. 18.4). Use of indirect reduction techniques and careful soft tissue handling to preserve periosteal blood supply increases the likelihood of fracture union.

Choosing the appropriate fracture type to use IMN is also important. For periarticular fractures, nailing is considered when two or more interlocking bolts can be used in the short segment. For intra-articular fractures, diaphyseal or metaphyseal fractures with a minimal or non-displaced intra-articular fracture component are a possible indication for nailing (see Fig. 18.4). IM fixation is gaining popularity for open fractures with segmental bone loss in the metaphysis, if the articular surface can be reconstructed with nail fixation with or without supplemental fixation,

Fig. 18.1 Preoperative radiographs of left femur with fracture on distal metaphyseal area (**a**, **b**). Postoperative radiographs with valgus malalignment (**c**, **d**). Radiographs 1 year after surgery showing fracture healing with worsened valgus malalignment and broken interlocking screws (**e**–**g**)

Fig. 18.2 Preoperative radiographs of right tibia with fracture on distal metaphyseal area (**a**, **b**). A valgus malalignment and a fracture gap at the medial aspect of tibia were noted on postoperative radiographs (**c**, **d**). Radiographs taken 11 months after operation show non-union (**e**, **f**)

Fig. 18.3 Preoperative radiographs show right distal femoral fracture (**a**, **b**). Postoperative radiographs with good alignment (**c**, **d**). Radiographs taken 5 months after operation show nonunion with loosening of implant (**e**, **f**)

Fig. 18.4 Preoperative radiographs showing right distal femoral fracture with extension of fracture line to the intra-articular (**a**, **b**). Intramedullary nailing was done with cannulated screw fixation for intra-articular fracture and percutaneous wiring for metaphyseal fracture (**c**, **d**). Postoperative 1-year radiographs show healing of fracture with good alignment (**e**, **f**)

since the nail is central to the anatomical axis and is less likely to fatigue when compared to plates with the expected prolonged healing times.

Methods to Overcome Biomechanical Disadvantages

The biomechanical disadvantages discussed earlier hinder healing of peri- and intra-articular fractures treated with IMN. To overcome these biomechanical disadvantages, both technique and implant strategies have improved.

Advances in Interlocking Bolts

Interlocking bolts, introduced by Klemm and Schellmann in 1972, provide resistance to rotational, axial, and bending forces by interlinking nail and bone [11].

In peri- and intra-articular fractures, three-point fixation cannot be provided by the nail due to discrepancy of diameter of the nail and medullary canal. There is no cortical contact with the nail in the metaphysis. Therefore, in peri- and intra-articular fractures, resistance of interlocking bolts against deforming stress is much more important than in diaphyseal fractures. Therefore, nail designs have changed to increase the number and directions of interlocking fixation to improve stability. Kneifel and Buckely reported an approximately 60% failure rate in distal tibia fractures with one distal interlocking bolt, but a 5% failure rate when two were used [12]. Laflamme et al. reported that the addition of two oblique bolts improved the mechanical stability of nail-bone construct compared with the conventional two mediolateral locking bolts in the proximal tibia [13]. In the distal tibia, Attal et al. showed that fibular plating was unnecessary when multidirectional interlocking bolts were used [14].

More recent nail designs also include options closer to the end of the nail to increase locking options. Interlocking bolts are available within 5 mm from the distal end in tibial nails, including four bolts within 40 mm from the distal end [15] (Fig. 18.5).

Another important advancement in interlocking fixation technology is stable angular fixation. Decreased movement at the screw-nail interface resulting from tight fit between interlocking bolt and nail hole contributes to mechanical stability. For this purpose, the Angular Stable Locking System (ASLS, Synthes®, Solothurn, Switzerland) using a resorbable sleeve to fill the screw-nail interface was introduced [16]. End caps [15], threaded interface in the nail, and alternative geometry locking devices are other options to create angle-stable constructs.

Biomechanical studies have shown a higher construct stiffness and reduced gap movement under axial and torsional loading with placement of angular stable interlocking bolts when compared to conventional interlocking in the distal tibia [17, 18]. However, there are also biomechanical studies showing no advantage [19, 20]. Similar to biomechanical studies, clinical studies have also reported conflicting results for the use of angle-stable fixation in the tibia [21, 22]. Although the results of studies do not provide definitive evidence, it is possible to consider using the angular stable interlocking screws as one way to overcome the biomechanical disadvantage of using IM nailing in peri- and intra-articular fractures.

Poller/Blocking Screws Poller or blocking screws, first described by Krettek et al. in 1999, have been used to correct malalignment and to increase stability of the bone-implant construct in treating peri- and intra-articular fractures using IM fixation [23]. It narrows the medullary canal in the metaphyseal area by creating a "pseudo" cortex and provides mechanical contact for nail as a cortex does in a diaphyseal fracture. Biomechanically, inserting poller screws in the proximal and distal tibia osteotomy model decreased the bone-implant construct deformation by 25% and 57%, respectively [23]. This improved biomechanical stability is the reason why people advocate for not removing them or using blocking drill bits or wires.

The original technique described by Kretteck et al. recommended inserting a screw at the concave side of displacement of the proximal and distal fragment [23]. Subsequently, others have

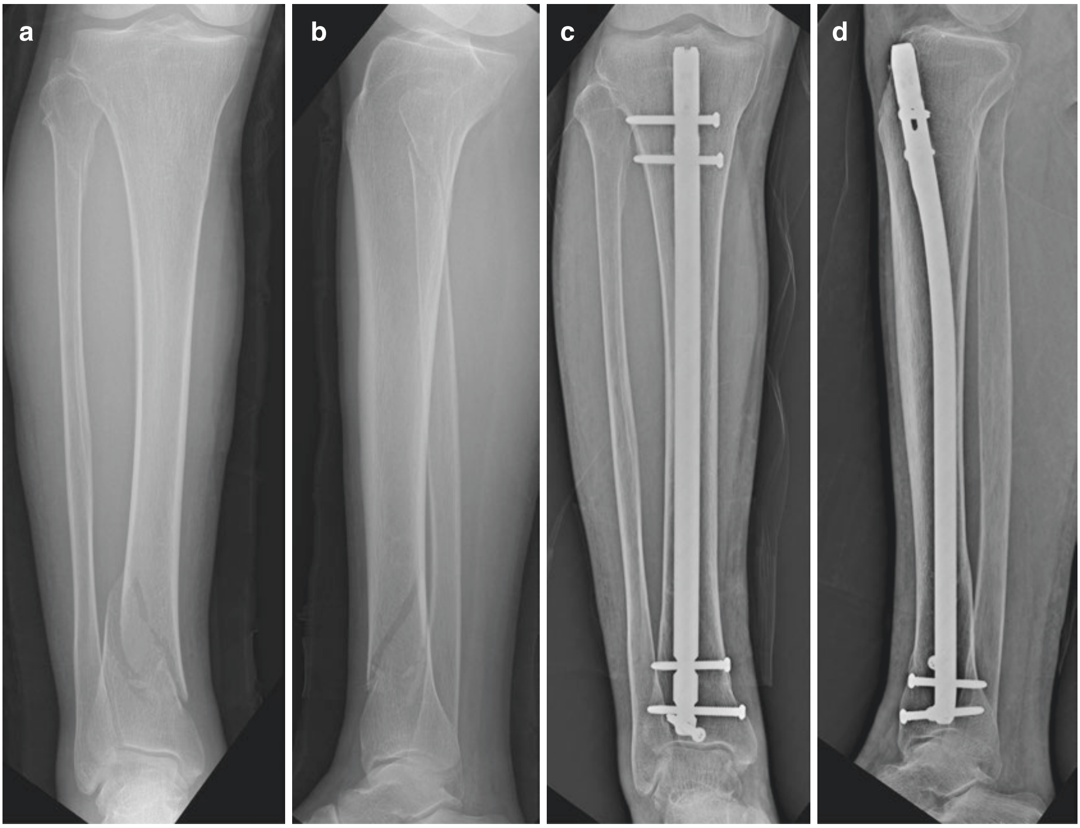

Fig. 18.5 Preoperative radiographs of right tibia with fracture on distal metaphyseal area (**a**, **b**). Postoperative 1-year radiographs show healing of fracture with good alignment (**c**, **d**)

modified this technique. Stedtfeld et al. suggested that the screw should be placed on the concave side of the deformity in the short fragment, close to the fracture [24], and a second poller screw placed on the convex side of the deformity in the short fragment, close to the tip of the nail (Fig. 18.6) to help center the nail in the short segment. More recently, Hannah et al. recommended placing a screw in the acute angles created by the long axis of the metaphyseal fragment and the fracture plane in oblique fractures [25] (Fig. 18.7).

Alternatively, a Steinmann pin or Kirschner wire can be inserted during IM fixation to facilitate reduction and then removed after fixation, which is called the *palisade method* [26]. However, if the fixation is removed, it takes away the continued biomechanical stability and increases the risk of failure during the healing process since the deforming forces persist.

Plate Augmentation Combining unicortical plating with an IMN can be helpful in obtaining and maintaining a reduction in a fracture within the metaphysis and can provide additional stability. In cases of using a nail in an intra-articular fracture, articular fragments can also be fixed with this supplemental plating. Dunbar et al. originally published this technique in open proximal tibia fractures using a plate placed through the open fracture wound and fixed with unicortical screws [27]. Others have published this technique in closed fractures including proximal third tibial fracture with extension into the tibial plateau [28], and distal femur and tibial fractures [29] (Fig. 18.8). Multiple studies report high fracture union and low complication rates [27–30].

Supplemental plating prevents angulation and loss of reduction during the nailing procedure. As

Fig. 18.6 Preoperative radiographs showing left distal femoral fracture (**a**, **b**). Good alignment was restored with IMN fixation and poller screws (**c**, **d**). Postoperative 36-month radiographs show healing of fracture with good alignment (**e**, **f**)

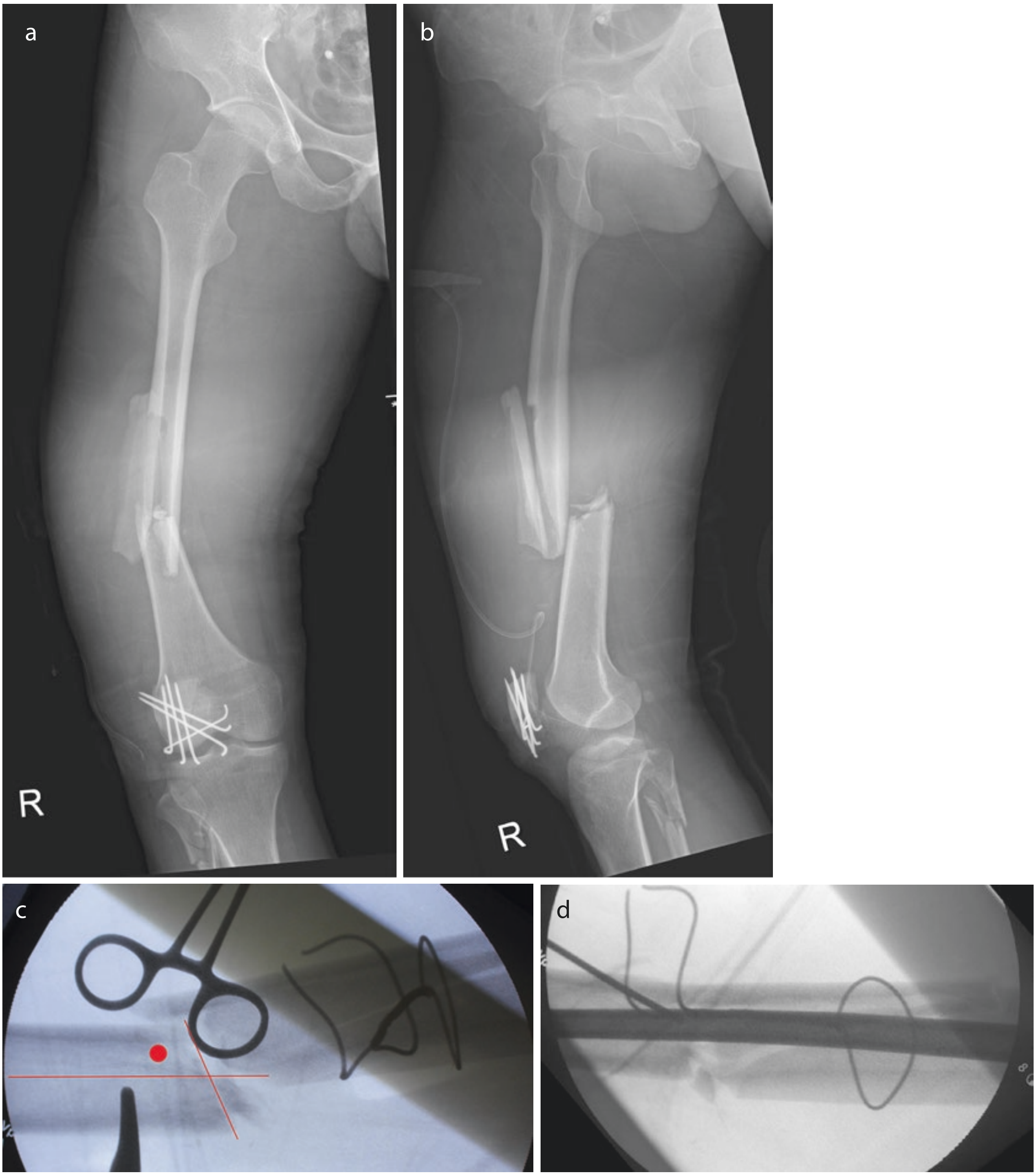

Fig. 18.7 Preoperative radiographs showing right femoral fracture with large butterfly fragment and extension to distal metaphyseal area (**a**, **b**). Fluoroscopic image shows sagittally displaced fracture. Red dot indicates the insertion point of poller screw (on acute angle of short fragment) (**c**). The fracture was well aligned after temporary fixation of 2.4-mm Steinmann pin (**d**). Postoperative radiographs showing restored alignment using IMN, percutaneous wiring, and poller screw (red arrow) (**e**, **f**). After 4 months from operation, fracture was nicely healed with callus formation (**g**, **h**)

Fig. 18.7 (continued)

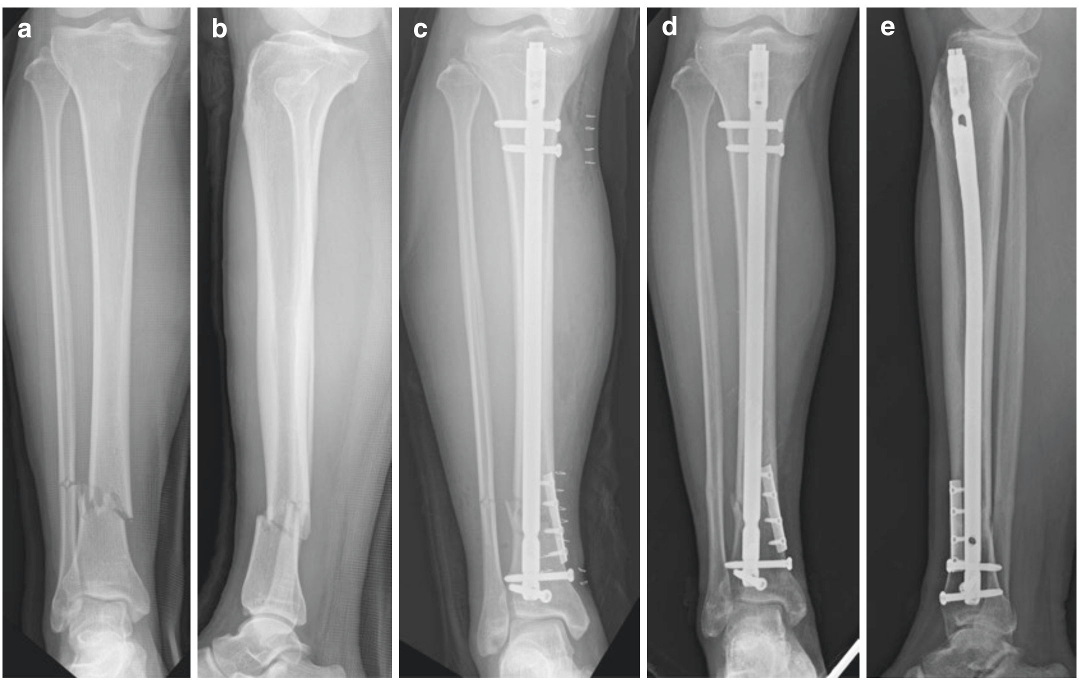

Fig. 18.8 Preoperative radiographs of right tibia with fracture on distal metaphyseal area (**a**, **b**). Good alignment was achieved with provisional plating and IMN (**c**). Postoperative 1-year radiographs show healing of fracture with good alignment (**d**, **e**)

with blocking screws, leaving the plates after IM nail fixation minimizes risk of failure during fracture healing. Additional fixation with a plate is an easy way to prevent malalignment, but it has the disadvantage of the additional surgical dissection required. Periosteal stripping can be minimized by utilizing biologically friendly soft tissue dissection.

Cases

Case 1

A 65-year-old female polytrauma patient presented with a left distal femur fracture after a motor vehicle accident (see Fig. 18.1a, b). The patient's femur was fixed using antegrade intramedullary nailing. On radiographs taken immediately after operation, a slight valgus alignment was noted (see Fig. 18.1c, d). The valgus deformity was worsened to about 14° (see Fig. 18.1e–g).

Why This Did Not Work

The wide canal of the distal femoral fragment contributed to malalignment. The stability of the bone-implant construct was lacking contact between nail and cortex. Fixation was limited to distal interlocking bolts purchase in thin cortices. Therefore, the fracture settled when the lateral cortex of distal fragment made contact with the nail, and this led to broken distal interlocking screws and increased valgus malalignment. Poller screws can be an option to achieve better alignment and increase stability.

Case 2

A 58-year-old male presented with open fracture of right distal tibia after a motor vehicle accident (see Fig. 18.2a, b). IMN was performed with residual valgus malalignment with a fracture gap at the medial aspect of tibia that was noted on postoperative radiographs (see Fig. 18.2c, d). Fracture was not healed after 11 months (see Fig. 18.2e, f).

Why This Did Not Work

The wide canal in the short fragment in distal tibia contributed to malalignment. IMN can be placed eccentrically in metaphyseal fractures. Despite having three interlocking bolts, fracture union failed. We believe this was due to fracture comminution leading to instability, valgus malalignment, and the poor biology caused by open fracture. This case is an example of where a supplemental plate and/or fibular ORIF would have added stability and controlled the fracture alignment.

Case 3

A 63-year-old male presented with fracture of right distal femur after motor vehicle accident (see Fig. 18.3a, b). Retrograde IMN was performed. Reduction including alignment and fracture gap was satisfactory postoperatively (see Fig. 18.3c, d). However, radiographs taken 5 months showed nonunion with implant failure (see Fig. 18.3e, f).

Why This Did Not Work

Since there is no bone-implant contact in short distal fragment in the wide metaphysis, three-point fixation cannot be achieved. Therefore, resistance of interlocking bolts against deforming stress is much more important than in diaphyseal fractures. However, it is difficult to gain sufficient stability because of the thin cortical bone in metaphyseal area. With only two interlocking bolts, sufficient fixation for bone healing could not be achieved. To obtain sufficient stability, an alternative option of interlocking for distal femur such as a spiral blade as well as poller screws can be used. Also short working length of the proximal fragment, due to overall length of the nail, and only one proximal interlocking screw can be considered as a risk factor for poor healing potential.

Case 4

A 72-year-old female presented with fracture of right distal femur with extension to the intra-articular area after ground level fall (see Fig. 18.4a, b). Intramedullary nailing was done with cannulated screw fixation for intra-articular fracture and percutaneous wiring for metaphyseal fracture (see Fig. 18.4c, d). Postoperative 1-year radiographs show healing of fracture with good alignment (see Fig. 18.4e, f).

Why This Worked

The fracture was a long spiral-type meta-diaphyseal fracture with extension to the knee joint. Initially, the non-displaced intra-articular fracture was fixed with screws to prevent later displacement of the intra-articular fracture during nailing. The minimally displaced metaphyseal fracture was then reduced, and percutaneous wiring performed to maintain reduction with minimal soft tissue disruption. After wiring, the fracture became a simple diaphyseal fracture making it more straightforward for IMN. Note that three interlocking bolts were used for the distal fragment to add stability.

Case 5

A 53-year-old male presented with fracture of right distal tibia after a ground level fall (see Fig. 18.5a, b). Intramedullary nailing was done with four distal interlocking screw fixation. Postoperative 1-year radiographs show healing of fracture with good alignment (see Fig. 18.5c, d).

Why This Worked

The fracture was reduced using reduction forceps with small stab incisions. After nail insertion, four interlocking bolts in multiple directions were used to increase fixation in the wide metaphyseal canal.

Case 6

A 17-year-old male presented with a fracture of left distal femur after motor vehicle accident (see Fig. 18.6a, b). The fracture was fixed with IMN using the poller screw technique (see Fig. 18.6c, d). Postoperative 36-month radiographs showed healing of the fracture with good alignment (see Fig. 18.6e, f).

Why This Worked

The poller screw technique aided in fracture reduction and provided additional stability. In this case, one screw was placed on the concave side of the deformity in the short fragment, close to the fracture. Another poller screw was placed on the convex side of deformity in the short fragment, closer to the tip of the nail.

Case 7

A 29-year-old male presented with a femoral shaft fracture due to motor vehicle accident (see Fig. 18.7a, b). During the operation, sagittal malalignment was noted on fluoroscopy. Therefore, a 2.4-mm Steinmann pin was inserted on the acute angle side (convex side of deformity) of the short (metaphyseal) fragment (see Fig. 18.7c). This corrected the sagittal fracture plane alignment as the nail was inserted (see Fig. 18.7d). After insertion of the nail, the Steinmann pin was replaced with a 3.5-mm cortical screw to avoid postoperative fracture displacement. Four interlocking bolts were used distally to maintain fracture stability (see Fig. 18.7e, f). Four months postoperatively, callus formation was noted with good alignment of the femur (see Fig. 18.7g, h).

Why This Worked

In this case, percutaneous wiring was performed to reduce large butterfly fragment and to convert the fracture to a simple pattern. The poller wire was inserted on acute angle (convex side of deformity) of the short fragment to restore sagittal plane alignment and later exchanged for a screw to maintain stability. Four distal interlocking screws were used to overcome the metaphysis and short articular segment.

Case 8

A 52-year-old male presented with a closed fracture of the right distal tibia after a mechanical fall (see Fig. 18.8a, b). IMN was performed with provisional plating. Note the one-third tubular plate with four unicortical screws used for plating (see Fig. 18.8c). Postoperative 1-year radiographs show healing of the fracture with good alignment (see Fig. 18.8d, e).

Why This Worked

Provisional plating with unicortical screws was used to obtain and maintain fracture reduction. Provisional plating can provide additional torsional fracture stability. A biologically friendly surgical approach is required to preserve microvasculature around the fracture site.

Recent Biomechanical Studies Comparing IM Fixation and Plating

In the distal femur, Heiney et al. reported that IMN fixation had a statistically significant increase in axial stiffness and significantly lower micromotion across the fracture site with axial compression compared to plating [30]. On the contrary, Zlowodzki et al. showed superior results with load to failure and lower rate of loss of fixation with the Less Invasive Stabilization System (LISS) (DePuy Synthes, Westchester, PA) compared to retrograde nail fixation [6]. But for torsional loading, IM nail was superior to LISS plating. They concluded that both implants have sufficient fixation for the proximal fragment, while LISS provides better distal fragment fixation (Table 18.1) [3, 5, 31–37].

For extra-articular proximal tibia fracture, biomechanical studies derived more consistent conclusions showing better biomechanical properties of IMN than plating [3, 30]. Nails tolerate higher loads to failure [3] and higher stiffness when compared to plating (see Table 18.1) [33].

For tibial plateau fractures, two studies compared IMN fixation to plating [34, 35]. Lasanianos et al. compared IM nails with compression bolts and plate fixation in the setting of lateral and dual (medial and lateral) plating [34]. In their study, IM nailing and dual plating showed no significant difference in subsidence and equivalent stiffness when compared to isolated lateral plating. The authors concluded that an IM nail with compression bolts provided fixation equivalent to dual plating and an elastic behavior for biological fixation equivalent to lateral plating. Hansen et al. reported equivalent axial load to failure for IMN and dual plating and superior load to failure versus lateral plating [38].

Multiple studies report the biomechanical advantages of IM fixation in the distal tibia [19, 35, 36, 39]. Hoenig et al. reported higher stiffness, load to failure, and energy to failure of IM nails compared with plating [20]. Hoegel et al. demonstrated that IM nails had superior stiffness under axial and torsional load [36]. Nourisa and Rouhi reported that IMN were biomechanically superior and tolerate earlier weight-bearing. However, they concluded that plating has advantages for bone healing due to differences in interfragmentary movement [39].

In the proximal humerus, however, plates are advantageous. Foruria et al. showed that locking plates have superior load to failure and torsional stiffness of the construct compared to IM nailing [37].

With increased interlocking fixation, IM nails may show superior biomechanical properties compared with plating. However, biomechanical superiority does not always translate into improved bone healing. Constructs with increased motion and low stiffness may result in nonunion or fixation failure [40]. On the contrary, excessively rigid fixation with no micromotion inhibits the secondary bone healing process. For secondary bone healing, controlled micromotion of the fragments is required. Therefore, increased stiffness is not always beneficial for bone healing, especially in the case of IMN fixation which

Table 18.1 Biomechanical test comparing intramedullary nail and plate in various periarticular and intra-articular fractures

Study	Type of loading	Type of specimen	Results
Distal femur			
Heiney et al. [31]	Axial	Synthetic femur	Stiffness (N/mm); IMN: 1106/DCS: 750/LCP: 625 Micromotion (mm); IMN: 1.96/DCS: 10.55/LCP: 17.74 Fatigue testing (cycles) IMN: 9000/DCS: not failed/LCP: 19,000 & 23,000
Zlowodzki et al. [5]	Axial/torsional	Human femur (fresh frozen)	Axial loading Load to failure (N); IMN: 913/LISS: 1028 Energy to failure (J); IMN: 1.1/LISS: 6.3 Stiffness (N/mm): IMN: 696/LISS: 111 Loss of distal fixation; IMN: 8/8, LISS: 1/16 Torsional loading Moment to failure (nm); IMN: 55/LISS: 30 Energy to failure (J); IMN: 18.2/LISS: 6.5 Stiffness (N/°): IMN: 1.6/LISS: 1.7
Proximal tibia			
Lee et al. [32]	Axial	Synthetic tibia	Load to failure (N); IMN: 22,879.6/LP: 12,249.3/DP: 14,387.3 Stiffness (N/mm); IMN: 5517.5/LP: 2308.7/DP: 4128.2
Högel et al. [33]	Axial	Human tibia (fresh frozen)	Load to failure (N); IMN: 1200/plate: 1350 Cycles to failure; IMN: 21,941/plate: 26,360 Stiffness (N/mm); IMN: 784/plate: 535
Lasanianos et al. [34]	Axial (compression)	Saw bone model tibial plateau fracture	Subsidence of medial plateau (mm); 500 N; IMN with compression bolts: 0.1/LP: 0.7/DP: 0.1 1000 N; IMN with compression bolts: 0.2/LP: 2.1/DP: 0.1 1500 N; IMN with compression bolts: 0.3/LP: 2.1/DP: 0.3 Stiffness (N/mm); IMN with compression bolts: 427.5/LP: 400.8/DP: 1295.6
Mueller et al. [3]	Axial	Human tibia (fresh frozen)	Maximal load (kN) IMN; CTN: 1.4/UTN: 0.96 Plate; Buttress plate: 0.54/LISS: 0.57 Relative movement in a varus direction (°) IMN; CTN: 0.51 Plate; Buttress plate: 4.17/LISS: 4.57
Distal tibia			
Kuhn et al. [35]	Axial and torsional	Synthetic tibias	Axial load Stiffness (N/mm); 350 N; IMN: 1037/plate: 465 600 N; IMN: 1081/plate: 881
			Interfragmentary movement (mm); IMN: 0.10/plate: 0.70
			Torsional load Stiffness (nm/°); 1.5 nm; IMN: 0.38/plate: 0.30 3.0 nm; IMN: 0.29/plate: 0.43
			Interfragmentary movement (mm); IMN: 0.83/plate: 0.34

(continued)

Table 18.1 (continued)

Study	Type of loading	Type of specimen	Results
Hoegel et al. [36]	Axial and torsional	Synthetic tibia	Axial load Stiffness (N/mm); IMN reamed: 709 IMN unreamed: 598 IMN unreamed with angle stable distal locking: 611 Plate: 466
			Interfragmentary movement (mm) IMN reamed: 0.1 IMN unreamed: 0.18 IMN unreamed with angle stable distal locking: 0.21 Plate: 1.03
			Torsional load Stiffness (nm/°); IMN reamed: 1.04 IMN unreamed: 0.7 IMN unreamed with angle stable distal locking: 0.73 Plate: 0.59
			Interfragmentary movement (°) IMN reamed: 8.2 IMN unreamed: 14.0 IMN unreamed with angle stable distal locking: 12.6 Plate: 15.0
Proximal humerus			
Foruria et al. [37]	Torsional	Human humerus	Interfragmentary motion (°); IMN: 3.5/LP: 3.2
			Energy to failure (J); IMN: 1.642/LP: 5.727
			Stiffness (N-M/°); IMN: 0.738/LP: 0.645

CTN Cannulated tibial nail, *IMN* Intramedullary nail, *LCP* Locking compression plate, *LISS* Less Invasive Stabilization System, *LP* locking plate, *UTN* Solid tibial nail

relies on callus formation. Similar to plating, the goal is to try and balance the rigidity of the construct for maintenance of reduction and facilitating bone healing.

Conclusion

Despite the known limitations of IM nailing for periarticular fractures, it can be used effectively when reduction and stability can be achieved, either with the nail alone or with adjunctive fixation. Recent IMN technology and techniques—such as poller screws, provisional plating, and increased interlocking fixation options—have been used successfully to expand indications for IMN. However, basic principles of internal fixation and fracture stability for periarticular fractures must be applied for healing of the fracture.

References

1. Kessler S, Hallfeldt K, Perren S, Schweiberer L. The effects of reaming and intramedullary nailing on fracture healing. Clin Orthop Relat Res. 1986;(212):18–25.
2. Klein B, Rahn A, Frigg R, Kessler S, Perren S. Reaming versus nonreaming in intramedullary nailing: interference with cortical circulation of the canine tibia. Arch Orthop Trauma Surg. 1990;109(6):314–6.

3. Mueller CA, Eingartner C, Schreitmueller E, Rupp S, Goldhahn J, Schuler F, et al. Primary stability of various forms of osteosynthesis in the treatment of fractures of the proximal tibia. J Bone Joint Surg Br. 2005;87(3):426–32.
4. Lang J, Cohen B, Bosse M, Kellam J. Proximal third tibial shaft fractures. Should they be nailed? Clin Orthop Relat Res. 1995;315:64–74.
5. Zlowodzki M, Williamson S, Cole PA, Zardiackas LD, Kregor PJ. Biomechanical evaluation of the less invasive stabilization system, angled blade plate, and retrograde intramedullary nail for the internal fixation of distal femur fractures. J Orthop Trauma. 2004;18(8):494–502.
6. Yoon RS, Gage MJ, Donegan DJ, Liporace FA. Intramedullary nailing and adjunct permanent plate fixation in complex tibia fractures. J Orthop Trauma. 2015;29(8):277–9.
7. Freedman E, Johnson E. Radiographic analysis of tibial fracture malalignment following intramedullary nailing. Clin Orthop Relat Res. 1995;(315):25–33.
8. Krupp R, Malkani A, Goodin R, Voor M. Optimal entry point for retrograde femoral nailing. J Orthop Trauma. 2003;17(2):100–5.
9. Buehler KC, Green J, Woll TS, Duwelius PJ. A Technique for intramedullary nailing of proximal third tibia fractures. J Orthop Trauma. 1997;11(3):218–23.
10. Ryan SP, Steen B, Tornetta PI. Semi-extended nailing of metaphyseal tibia fractures: alignment and incidence of postoperative knee pain. J Orthop Trauma. 2014;28(5):263–9.
11. Klemm K, Schellmann W. [Dynamic and static locking of the intramedullary nail]. Monatsschr Unfallheilkd Versicher Versorg Verkehrsmed. 1972;75:568–575. [Article in German].
12. Kneifel T, Buckley R. A comparison of one versus two distal locking screws in tibia1 fractures treated with unreamed tibia1 nails: a prospective randomized clinical trial. Injury. 1996;27(4):271–3.
13. Laflamme GY, Heimlich D, Stephen D, Kreder HJ, Whyne CM. Proximal tibial fracture stability with intramedullary nail fixation using oblique interlocking screws. J Orthop Trauma. 2003;17(7):496–502.
14. Attal R, Maestri V, Doshi H, Onder U, Smekal V, Blauth M, et al. The influence of distal locking on the need for fibular plating in intramedullary nailing of distal metaphyseal tibiofibular fractures. Bone Joint J. 2014;96–B:385–9.
15. Kuhn S, Hansen M, Rommens PM. Extending the indications of intramedullary nailing with the Expert Tibial Nail®. Acta Chir Orthop Traumatol Cechoslov. 2008;75(2):77–87.
16. Höntzsch D, Blauth M, Attal R. Angle-stable fixation of intramedullary nails using the Angular Stable Locking System® (ASLS). Oper Orthop Traumatol. 2011;23(5):387–96.
17. J H, B L, D H, B G, K S. Angle stable interlocking screws improve construct stability of intramedullary nailing of distal tibia fractures: a biomechanical study. Injury. 2009;40(7):767–71.
18. Wähnert D, Stolarczyk Y, Hofmann GO, Mückley T. The primary stability of angle-stable versus conventional locked intramedullary nails. Int Orthop. 2012;36(5):1059–64.
19. Augat P, Hoegel F, Stephan D, Hoffmann S, Buehren V. Biomechanical effects of angular stable locking in intramedullary nails for the fixation of distal tibia fractures. Proc Inst Mech Eng H. 2016;230(11):1016–23.
20. Hoenig M, Gao F, Kinder J, Zhang L, Collinge C, Merk BR. Extra-articular distal tibia fractures: a mechanical evaluation of 4 different treatment methods. J Orthop Trauma. 2010;24(1):30–5.
21. Höntzsch D, Schaser KD, Hofmann GO, Pohlemann T, Einar S, Rothenbach E, et al. Evaluation of the effectiveness of the angular stable locking system in patients with distal tibial fractures treated with intramedullary nailing: a multicenter randomized controlled trial. J Bone Joint Surg Am. 2014;96(22):1889–97.
22. Li Y, Liu L, Tang X, Pei F, Wang G, Fang Y, et al. Comparison of low, multidirectional locked nailing and plating in the treatment of distal tibial metadiaphyseal fractures. Int Orthop. 2012;36(7):1457–62.
23. Krettek C, Miclau T, Schandelmaier P, Stephan C, Mohlmann U, Tscherne H. The mechanical effect of blocking screws (‘Poller screws’) in stabilizing tibia fractures with short proximal or distal fragments after insertion of small-diameter intramedullary nails. J Orthop Trauma. 1999;13(8):550–3.
24. Stedtfeld HW, Mittlmeier T, Landgraf P, Ewert A. The logic and clinical applications of blocking screws. J Bone Joint Surg Am. 2004;86(suppl-2):17–25.
25. Hannah A, Aboelmagd T, Yip G, Hull P. A novel technique for accurate Poller (blocking) screw placement. Injury. 2014;45(6):1011–4.
26. Biewener A, Grass R, Holch M, Zwipp H. Intramedullary nail placement with percutaneous Kirschner wires. Illustration of method and clinical examples. Unfallchirurg. 2002;105(1):65–70.
27. Dunbar R, Nork S, Barei D, Mills W. Provisional plating of type III open tibia fractures prior to intramedullary nailing. J Orthop Trauma. 2005;19(6):412–4.
28. Kubiak E, Camuso M, Barei D, Nork S. Operative treatment of ipsi-lateral noncontiguous unicondylar tibial plateau and shaft fractures: combining plates and nails. J Orthop Trauma. 2008;22(8):560–5.
29. Yoon RS, Liporace FA. Intramedullary nail and plate combination fixation for complex distal tibia fractures: when and how? J Orthop Trauma. 2016;30(Suppl 4):S17–21.
30. Yoon RS, Bible J, Marcus MS, Donegan DJ, Bergmann KA, Siebler JC, et al. Outcomes following combined intramedullary nail and plate fixation for complex tibia fractures: a multi-centre study. Injury. 2015;46(6):1097–101.
31. Heiney JP, Battula S, Connor JAO, Ebraheim N, Schoenfeld AJ, Vrabec G. Clinical biomechanics distal femoral fixation : a biomechanical comparison of retrograde nail, retrograde intramedullary nail,

and prototype locking retrograde nail. Clin Biomech (Bristol, Avon). 2012;27(7):692–6.

32. Lee SM, Oh CW, Oh JK, Kim JW, Lee HJ, Chon CS, et al. Biomechanical analysis of operative methods in the treatment of extra-articular fracture of the proximal tibia. Clin Orthop Surg. 2014;6(3):312–7.
33. Högel F, Hoffmann S, Panzer S, Wimber J, Buhren V, Augat P. Biomechanical comparison of intramedullar versus extramedullar stabilization of intra-articular tibial plateau fractures. Arch Orthop Trauma Surg. 2012;133(1):59–64.
34. Lasanianos NG, Garnavos C, Magnisalis E, Kourkoulis S, Babis GC. A comparative biomechanical study for complex tibial plateau fractures: nailing and compression bolts versus modern and traditional plating. Injury. 2013;44(10):1333–9.
35. Kuhn S, Greenfield J, Arand C, Jarmolaew A, Appelmann P, Mehler D, et al. Treatment of distal intraarticular tibial fractures: a biomechanical evaluation of intramedullary nailing vs. angle-stable plate osteosynthesis. Injury. 2015;46:S99–S103.
36. Hoegel FW, Hoffmann S, Weninger P, Bu V, Augat P. Biomechanical comparison of locked plate osteosynthesis, reamed and unreamed nailing in conventional interlocking technique, and unreamed angle stable nailing in distal tibia fractures. J Trauma Acute Care Surg. 2012;73(4):933–8.
37. Foruria AM, Teresa M, Revilla C, Munuera L, Sanchez-Sotelo J. Clinical Biomechanics Proximal humerus fracture rotational stability after fixation using a locking plate or a fixed-angle locked nail: The role of implant stiffness. Clin Biomech (Briston, Avon). 2010;25(4):307–11.
38. Hansen M, Mehler D, Hessmann M, Blum J, Rommens P. Intramedullary stabilization of extraarticular proximal tibial fractures: a biomechanical comparison of intramedullary and extramedullary implants including a new proximal tibia nail (PTN). J Orthop Trauma. 2007;21(10):701–9.
39. Nourisa J, Rouhi G. Biomechanical evaluation of intramedullary nail and bone plate for the fixation of distal metaphyseal fractures. J Mech Behav Biomed Mater. 2016;56:34–44.
40. Kenwright J, Goodship A. Controlled mechanical stimulation in the treatment of tibial fractures. Clin Orthop Relat Res. 1989;(241):36–47.

19 Use in Nonunions and Malunions

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Introduction

Malunion and nonunion are distinct entities with distinct procedural goals. The goal of malunion surgery is restoration of axis to a functional or pre-morbid alignment. Nonunion surgery encompasses a number of different specific etiologies with the common goal of fracture union. Different approaches to nonunion surgery rely on precise failure analysis and customized solutions. While many have traditionally tried to explain nonunions as primarily biological deficits or mechanical deficits [1, 2], there is undoubtedly the common pathway of providing a stable mechanical environment for healing.

Intramedullary nails are most commonly utilized in lower extremity malunion and nonunion repair, especially for diaphyseal applications. Newer nail implant designs [3, 4] have allowed for expanded application in metaphyseal applications. These same newer generation nail designs have expanded applications in the upper extremity, including in the proximal humerus. There are several indications for the use of intramedullary nail in repairs of nonunion and malunion. First, nails outperform plates in osteoporotic applications, and a significant number of these repairs occur in clinical situations with compromised bone quality. Second, nails are load-sharing implants, and thus, most of these procedures can be performed to allow for immediate postoperative weight-bearing. Finally, the central position of the implant creates a unique stability profile with an ideal combination of stiffness and permissive micromotion that is optimal for bone healing.

Key Concepts for Intramedullary Malunion and Nonunion Repair

Manage Adjacent Joint Mobility

One of the most critical and often overlooked factors is the technical feasibility of nailing through a stiff joint. For many periarticular nonunions, the adjacent joint will have significant stiffness due to long-term pain and immobility. In this setting, the mobility of the short segment is critical to establishing a perfect insertion site and angle which are needed to restore alignment and optimize fixation paths for interlocking screws. If preoperative exam suggests that mobility is so restricted that necessary joint movement is not feasible, then nailing may not be a feasible fixation alternative. Alternatively, nailing can still be performed but will require initial joint exploration, lysis of adhesions, and mobilization. This, as in most fixation constructs, adds considerably morbidity to the procedure but is critical to not

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only allow for instrumentation but also to decrease stress on the short segment fixation.

Determine the Etiology

Analysis of the cause of the nonunion or malunion is critical for planning the proposed solution. For malunions, a myriad of causes is commonly seen, but some of the most important issues include recognition of an initially incorrect implant choice or position and simply a poorly executed or selected surgical plan. An alternative technique will typically be required for correction. For nonunions, the failure analysis is even more critical and can range from obvious to completely subtle or even indeterminate. Frequently an overaggressive surgical tactic has disrupted local biology, and the proposed solution must add minimal further biological insult. Mechanical stability frequently negatively affects the normal physiologic repair process. Fixation constructs can be too stiff [5–7] or lack adequate stability; however, the solution is rarely reversal of the initial problem. The initial healing response can provide a gauge of local biologic healing potential and can provide information about the potential for healing and the need for biological augmentation. In most situations, stability is optimized regardless of the initial stability deficit since most repairs will involve a primary bone healing pathway.

Recognize the Deformity and Restore Axis

The key to malunion repair is an adequate deformity evaluation. For periarticular deformity, a detailed analysis of anatomic and mechanical axis deviation is performed. Contemporary computer or web-based systems are helpful to precisely determine magnitudes of deformity but also have the benefit of allowing for trial reduction efforts to help select the optimal correction and implant position. For nonunions, subtle deformity must be identified and when feasible corrected as part of the nonunion repair. Mechanical alignment restoration frequently optimizes loading across a nonunion/malunion repair site and is a critical, often overlooked step in successful treatment of nonunions. Frequently, alignment correction alone changes the local mechanical environment significantly enough that healing can occur without biological augmentation, especially when malalignment was significant.

Preserve the Local Biologic Environment

By necessity, most osteotomies for malunion correction are biologically costly in their requirement for significant exposure and then sectioning of the bone. Safe technique requires some near circumferential protection of adjacent soft tissue/neurovascular structures during osteotomy and associated elevation of soft tissues from the bone. Undoubtedly this insult can lead to slow healing, and all efforts should be made to use meticulous soft tissue handling and dissection techniques. Nonunions are similarly managed and maintenance of local blood supply can often be overlooked in the typically challenging efforts to expose nonunion sites and related fixation hardware. When possible, consider a technique to optimize the blood supply to the nonunion site such as osteoperiosteal decortication [8] versus the use of a high-speed burr.

Maximize Stability in the Short Segment

Nails for metaphyseal nonunions and malunions were more unusual historically because of the challenges of maintaining stability in the epiphyseal segment. Newer approaches and technology have buoyed their use. Contemporary interlocking bolts that provide some angular stability and do not easily toggle under loading conditions [9] allow nails to be used in many traditional neutralization plating solutions. While many nails have the ability to create some degree of fracture compression, the exact magnitude of compression is not well described in the literature, and the compression is occurring through metaphyseal screw fixation (poorer bone quality than diaphyseal), so it is safe to assume that this is not equivalent to the compression that can be achieved with a plate transfixing a fracture in diaphyseal bone.

Additionally, modern use of plate nail constructs allows for periarticular osteotomies or nonunion repairs in very short segments even in poor bone quality [10]. The plate adds multiple points of fixation and thus significant torsional stability augmentation above and below a nonunion or osteotomy site, as screws can easily be targeted around the intramedullary implant.

Case 1: Tibial Exchange Nailing

This is a 28-year-old patient struck by automobile at 40 mph. He sustained a high grade open, comminuted tibial shaft fracture (Fig. 19.1a, b). He underwent uneventful tibial nailing requiring a posterior blocking screw for maintenance of sagittal plane reduction. Postoperative radiographs revealed acceptable alignment (Fig. 19.1c, d). At 5 months post-surgery, this patient was reporting progressive pain in the proximal tibia with weight-bearing, and his radiographs demonstrated resorption at the fracture line. There was minimal evidence of active healing (Fig. 19.1e, f). He underwent exchange nailing to a larger reamed nail and had a medial compression plate added along with intramedullary bone graft from the ipsilateral femur (Fig. 19.1g). His postoperative radiographs demonstrated good fracture site compression with slight varus deformity (Fig. 19.1h, i). He progressed to uneventful healing 8 months later (Fig. 19.1j, k).

Why This Works

The initial failure in this nail construct was multifactorial. There were both instability and subtle distraction or resorption at the fracture site. An exchange nail improves the mechanical environment with increases in bending ($radius^3$) and torsional stiffness ($radius^4$). However, the ability to provide compression in a nonunion situation is unreliable in anything more than a minimal gap. Additionally, simple dynamization may lead to some interfragmentary compression but at the cost of more fracture site torsional instability. Major changes in fracture site mechanics are seen more with exchange nails done for diaphyseal nonunions, and the effect of a larger nail would not be as significant in this situation where the nonunion site is in the proximal half of the tibia. Plate fixation around the nail is well described for femur nonunions [11]. In this setting, it provided optimal compression across the nonunion site and improved the torsional stability across the nonunion, creating an optimized fracture healing environment.

Case 2: Closed Femoral Shortening

This is a 45-year-old male who had nonsurgical management of a right femur fracture as a teenager. He healed uneventfully but had an approximately 1.5-cm limb length deficit from shortening of the femur during healing. He presented to our clinic complaining of worsening contralateral hip pain, lateral foot pain, and bothersome limb length discrepancy to his left lower extremity. His symptoms were progressive and began impacting his ability to work. Standing full length radiograph demonstrates femoral length asymmetry and pelvic obliquity (Fig. 19.2a). We performed closed femoral shortening procedure below the lesser trochanter using a large antegrade reamed nail which we also aggressively backslapped to maximize contact (Fig. 19.2b–g). Follow-up standing radiography demonstrates excellent correction of length asymmetry and pelvic tilting and good fracture healing (Fig. 19.2h–j).

Why This Works

Any diaphyseal osteotomy can be slow healing due to local procedural dissection and bone sawing and/or drilling. Use of the intramedullary saw maintains soft tissue attachments to the

Fig. 19.1 (**a**, **b**) Injury anteroposterior and lateral radiographs of the tibia demonstrate a displaced, comminuted fracture of the tibia. (**c**, **d**) AP and lateral radiographs following surgery demonstrating good restoration of alignment. (**e**, **f**) Five months after original surgery, patient describes increasing pain and radiograph shows no obvious healing and some resorption at the primary fracture line. (**g**) Intraoperative radiograph showing use of articulated tensioning device to optimize compression (**h**, **i**) medially creating subtle varus deformity. Eight months later, he progresses to uneventful union with return to function (**j**, **k**)

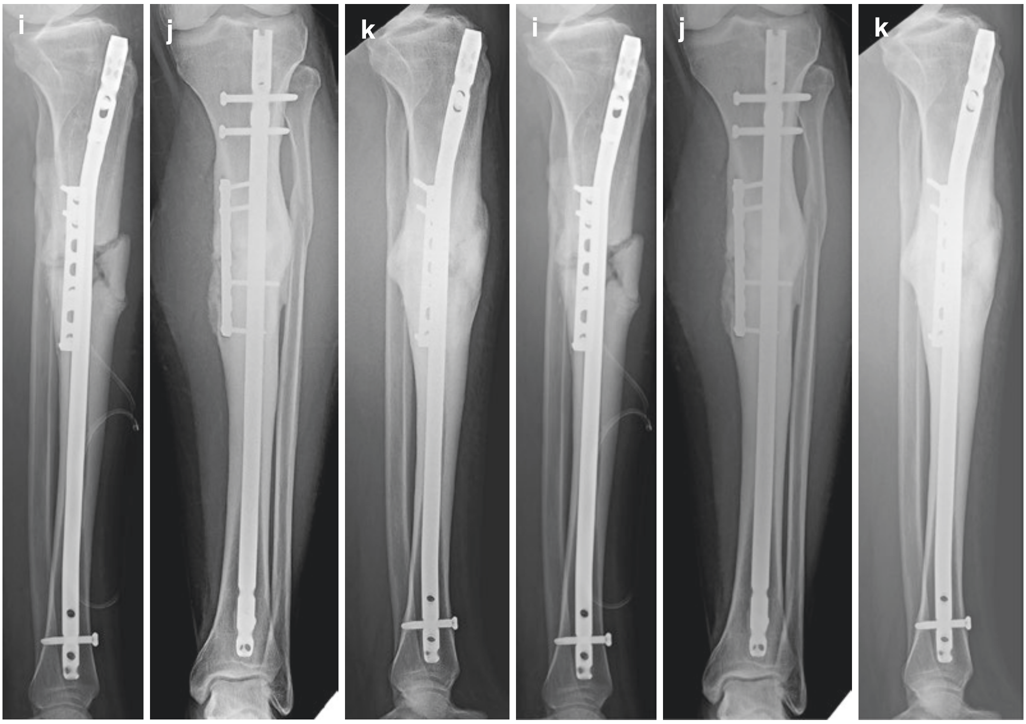

Fig. 19.1 (continued)

osteotomized bone segments. From a purely mechanical standpoint, circumferential diaphyseal contact after osteotomy optimizes axial stability especially after the nail is backslapped or compression is achieved. Any asymmetry in the osteotomy creates point loading and limits contact. The intramedullary saw allows for perfect circumferential contact. The ideal level is not clearly defined by the literature for intramedullary osteotomy, but the proximal diaphysis is typically selected and simplifies use of the rigid, straight saw. A large caliber, canal fitting nail is used to optimize all parameters of stability, and healing is typically rapid and predictable.

Case 3: Tibial Diaphyseal Metaphyseal Clamshell Osteotomy

This is a 75-year-old woman with a long-standing left tibia malunion. She had a fracture of her tibia from low-energy fall. This was treated with casting and bracing for protracted period with subsequent refracture. Surgery was not offered by her local surgeons and she proceeded to union. She was previously a community ambulator, but she was unable to after her treatment due to instability and ankle pain. Radiographs of her tibia demonstrate a well-fixed total knee arthroplasty and a tibial malunion with a 35-degree magnitude valgus deformity and a fibular malunion (Fig. 19.3a, b). A clamshell osteotomy [12, 13] was performed with standard antegrade nailing (Fig. 19.3c–h). To augment stability, the nail was positioned against the anterior cortex in the distal fragment with the use of a blocking screw [14]. Near complete restoration of alignment was achieved (Fig. 19.3i–k). Patient healed slowly but returned to unrestricted ambulation over the subsequent 3 months (Fig. 19.3l, m).

Why This Works

This technique was successful in highly compromised bone due to the added stability benefit of

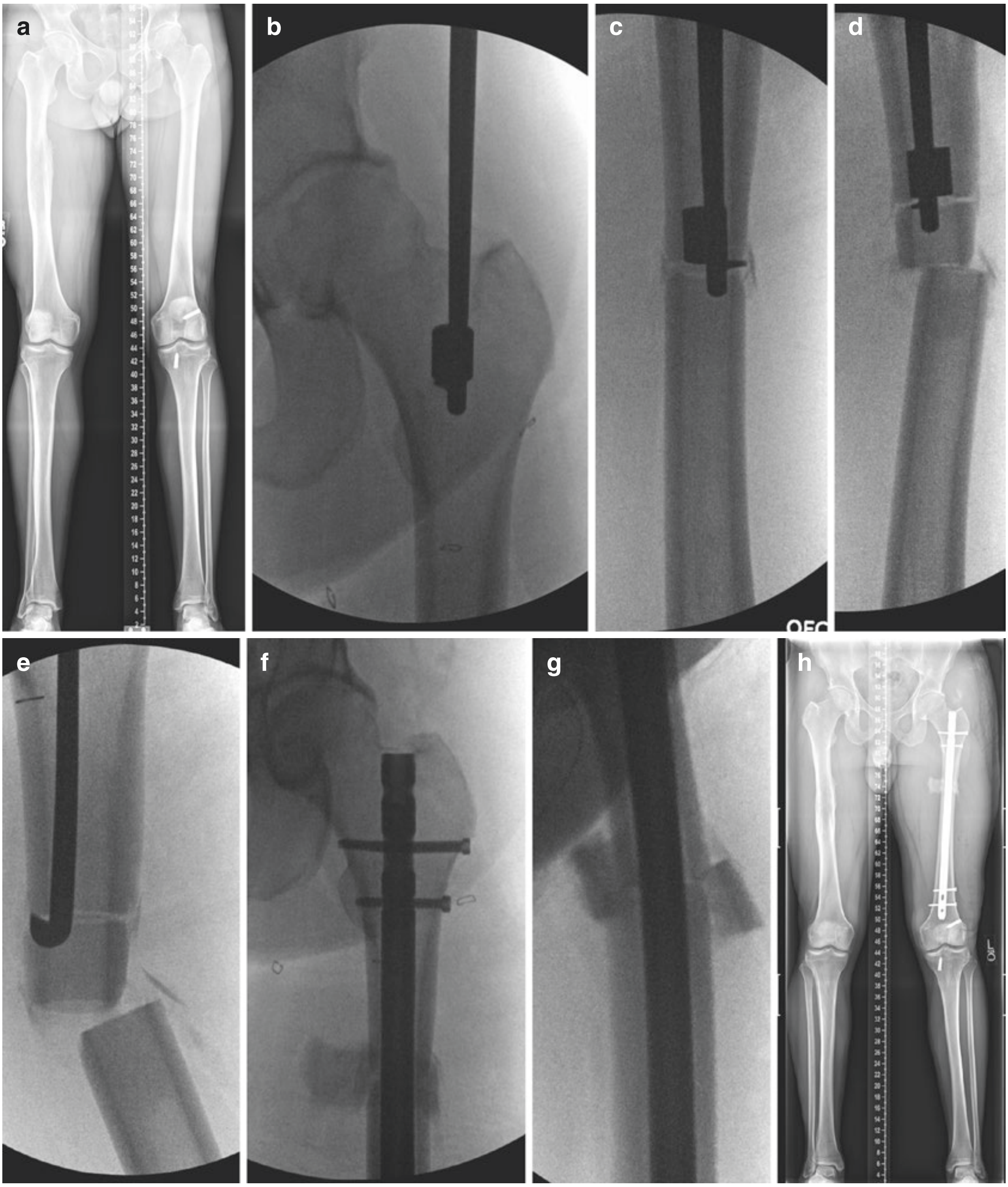

Fig. 19.2 (**a**) Standing long cassette radiography for limb length and alignment demonstrates limb length discrepancy and marked pelvic obliquity. (**b**) Insertion of intramedullary saw through piriformis fossa. (**c**) Completion of distal cut. (**d**) Completion of proximal cut and limb rotation to ensure mobility of osteotomy segments. (**e**) Intramedullary ring cutting device to divide intervening shortened segment. (**f**, **g**) AP and lateral views of large diameter interlocked nail placement. Aggressive backslapping performed to optimize contact. (**h**) Standing alignment film following surgery demonstrates excellent correction of pelvic obliquity. (**i**, **j**) Good healing of the osteotomy is shown in perpendicular views with early resorption and partial incorporation of cortical segments

Fig. 19.2 (continued)

an intramedullary implant and new interlocking technology that limits screw toggling and loosening under physiological loading conditions. The nail design provides for multiplanar interlocking to improve short segment stability and frontal plane. The use of the intramedullary nail for fixation instead of plate fixation places the implant along the anatomic axis and minimizes the bending loads with weight-bearing. Distal fixation was maintained even with canal-implant mismatch with the use of a blocking screw that guided the nail anteriorly and optimized endosteal contact. This osteotomy technique has long union times, especially in poorer hosts, so specialized interlocks are also critical for maintaining stability during healing.

Case 4: Recalcitrant Distal Femur Plate Nonunion Exchanged to Nail

A 64-year-old female smoker sustained a distal femur fracture above a TKA (Fig. 19.4a, b). She was initially plated with standard lateral locking plate (Fig. 19.4c, d) but went on to symptomatic atrophic nonunion 6 months later (Fig. 19.4e, f). She was revised with an open approach, medial fibular allograft, lateral plate exchange and application of bone morphogenetic protein-2 (BMP-2) and freeze-dried corticocancellous chips (Fig. 19.4g, h). Six months following this revision operation, her pain was increasing with signs of hardware instability (Fig. 19.4i, j). Hardware removal and placement of retrograde intramedullary nail with reaming were performed through her TKA without supplemental plating or bone grafting (Fig. 19.4k, l). She proceeded to uneventful healing 4 months later (Fig. 19.4m, n).

Why This Works

The ideal stiffness for periarticular fractures of the distal femur remains elusive [15–17]. While there has been concern about too much stiffness in distal femur fixation constructs, fractures are usually stabilized with long plates and well-distributed screws. When healing fails, traditional approaches are utilized which typically involve hardware exchange, a strong stable fixation construct, and local bone grafting. Revision failures are unusual, but both biological and mechanical etiologies are frequently investigated. In this case, the biological environment seemed favorable, despite a smoking history, and since a potent biological implant was previously utilized, mechanical stability seemed to be a more likely problem. The stability achieved from an intramedullary implant is unique from plate fixation constructs because it depends less on bone quality and more on medullary canal fill and interlock stability. There is a unique type of stability achieved with intramedullary fixation that is not clearly defined. However, what is known is

Fig. 19.3 (**a**, **b**) AP and lateral views of valgus tibia malunion with well-fixed TKA. (**c**) Intraoperative view of fibular osteotomy through small lateral approach. (**d**) Standard antegrade start site available in front of tibial tray for TKA. (**e**) Mobilization of proximal osteotomy with osteotome. Clamshell drill holes can be seen in intervening segment. (**f**) A medial femoral distractor is used to fine-tune the reduction and assist in centralization of the guide wire. (**g**) A posterior blocking screw is added to optimize stability in the small distal segment. (**h**) The nail is centralized and passed deep into the distal fragment. (**i**–**k**) Postoperative radiographs show excellent correction of alignment. (**l**, **m**) Radiographs demonstrating late fracture healing with mature callus formation

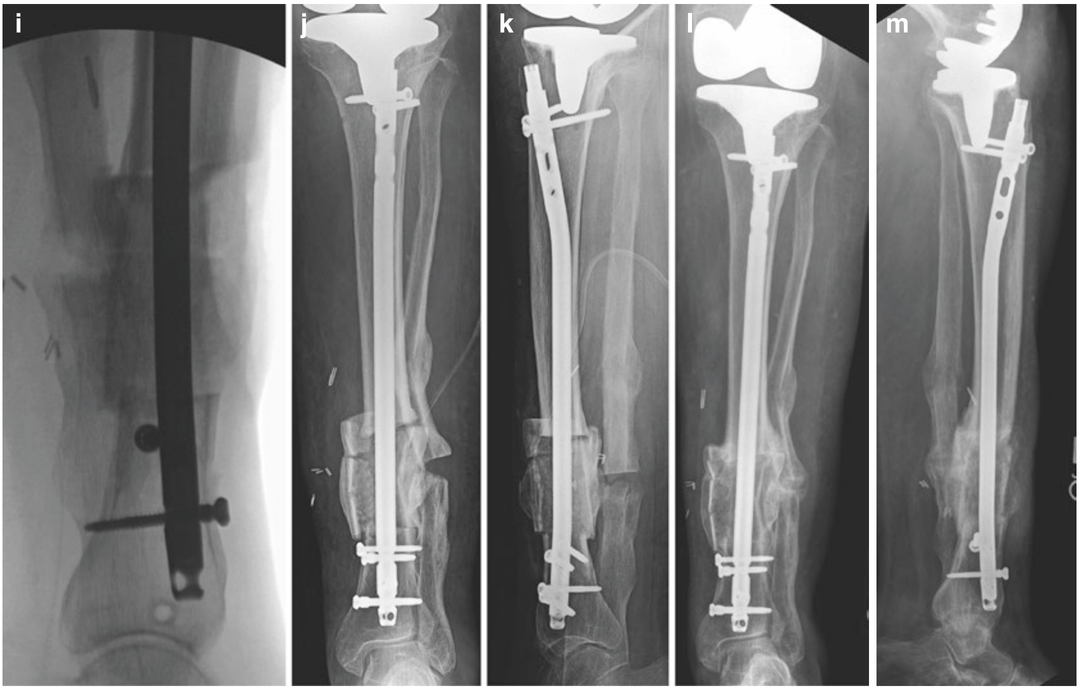

Fig. 19.3 (continued)

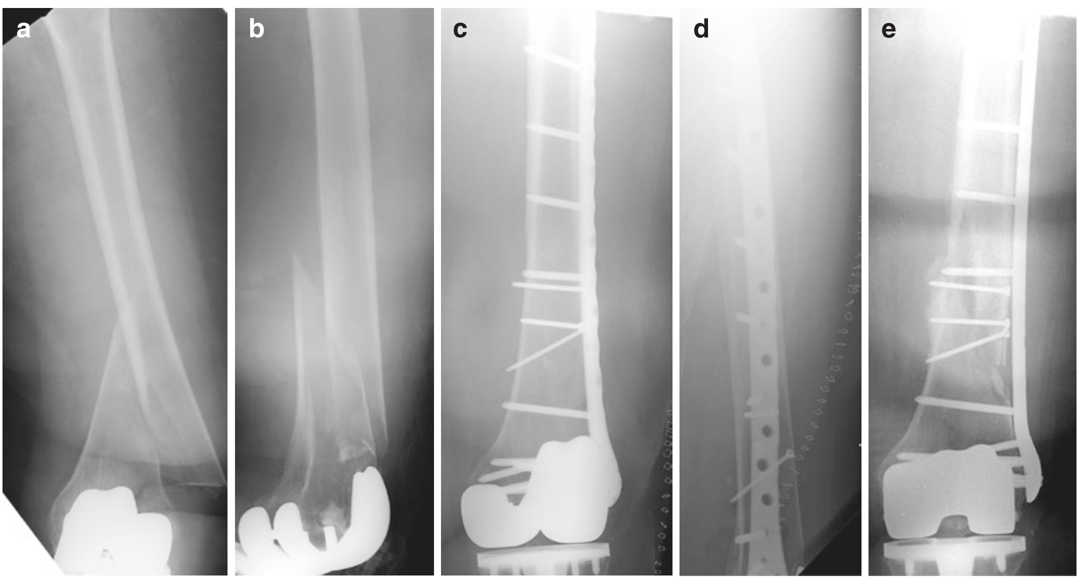

Fig. 19.4 (**a**, **b**) Injury radiographs demonstrate a simple, displaced, oblique distal femur fracture above a TKA. (**c**, **d**) Postoperative radiographs after initial fixation attempt using lag screws and the plate for neutralization. (**e**, **f**) Development of symptomatic atrophic nonunion without hardware failure. (**g**, **h**) Revision for nonunion with medial structural fibular allograft, corticocancellous allograft, and BMP-2 to the fracture site. (**i**, **j**) Development into recalcitrant nonunion with resorption at fracture line and severe pain. (**k**, **l**) Nonunion treated again with hardware removal and placement of retrograde reamed nail without grafting. (**m**, **n**) Solid healing and return to full function after several months

Fig. 19.4 (continued)

that the working lengths are not as easily modulated as they are with plate constructs, and the nail provides a seemingly ideal combination of durable fixation with permissive flexibility for secondary healing, even in subacute healing situations like nonunions. We placed the largest possible retrograde nail and performed static locking to optimize axial and bending stiffness. We added distal fixation by using a blade device fixed to the nail to improve metaphyseal stability. We also allowed immediate weight-bearing to allow for physiological construct deformation favorable for healing.

Conclusion

Intramedullary nailing is a newer and powerful technique for complex osteotomies and nonunion care. Nailing provides the benefit of a surgical approach that can optimize maintenance of biological attachments to healing bone fragments and also has the added benefit of allowing for weight-bearing during the healing period. Principles of osteotomy and nonunion treatment and management do not change significantly with the use of nails. Axis measurement and correction planning are still critical. However, the ability to achieve stable compression with a single implant is not equivalent to the stability achieved with a plate. On the other hand, compromised bone quality is commonly present in many nonunions and malunions, and the added stability of the interlocked intramedullary implant is superior. The previous limitations of using intramedullary nails for many nonunions and osteotomies outside of the diaphysis have been addressed with new interlocking technology that allows for longer-term stability of screws and combining adjunctive plating with nails. However, the long-term functional results of these newer approaches require further study. Nonetheless, intramedullary nailing using newest implant technology and techniques should now be considered as a viable option for many complex nonunions and malunions.

References

1. McKibbin B. The biology of fracture healing in long bones. J Bone Joint Surg Br. 1978;60-B(2):150–62.
2. Sen MK, Miclau T. Autologous iliac crest bone graft: should it still be the gold standard for treating nonunions? Injury. 2007;38(Suppl 1):S75–80.
3. Pape HC, Archdeacon MT. Introduction: biology and new technical trends in intramedullary nailing of the femur. J Orthop Trauma. 2009;2(5 Suppl):S1.
4. Kaspar K, Schell H, Seebeck P, Thompson MS, Schütz M, Haas NP, Duda GN. Angle stable locking reduces interfragmentary movements and promotes healing after unreamed nailing. Study of a displaced osteotomy model in sheep tibiae. J Bone Joint Surg Am. 2005;87(9):2028–37.
5. Henderson CE, Lujan TJ, Kuhl LL, Bottlang M, Fitzpatrick DC, Marsh JL. 2010 mid-America Orthopaedic Association Physician in Training Award: healing complications are common after locked plating for distal femur fractures. Clin Orthop Relat Res. 2011;469(6):1757–65.
6. Parks C, McAndrew CM, Spraggs-Hughes A, Ricci WM, Silva MJ, Gardner MJ. In-vivo stiffness assessment of distal femur fracture locked plating constructs. Clin Biomech (Bristol, Avon). 2018;56:46–51.
7. Alexander J, Morris RP, Kaimrajh D, Milne E, Latta L, Flink A, Lindsey RW. Biomechanical evaluation of periprosthetic refractures following distal femur locking plate fixation. Injury. 2015;46(12):2368–73.
8. Ramoutar DN, Rodrigues J, Quah C, Boulton C, Moran CG. Judet decortication and compression plate fixation of long bone non-union: is bone graft necessary? Injury. 2011;42(12):1430–4.
9. Wähnert D, Stolarczyk Y, Hoffmeier KL, Raschke MJ, Hofmann GO, Mückley T. Long-term stability of angle-stable versus conventional locked intramedullary nails in distal tibia fractures. BMC Musculoskel Disord. 2013;14:66. https://doi.org/10.1186/1471-2474-14-66.
10. Liporace FA, Yoon RS. Nail plate combination technique for native and periprosthetic distal femur fractures. J Orthop Trauma. 2019;33(2):e64–8.
11. Hakeos WM, Richards JE, Obremskey WT. Plate fixation of femoral nonunions over an intramedullary nail with autogenous bone grafting. J Orthop Trauma. 2011;25(2):84–9.
12. Russell GV, Graves ML, Archdeacon MT, Barei DP, Brien GA Jr, Porter SE. The clamshell osteotomy: a new technique to correct complex diaphyseal malunions. J Bone Joint Surg Am. 2009;91(2):314–24.
13. Russell GV, Graves ML, Archdeacon MT, Barei DP, Brien GA Jr, Porter SE. The clamshell osteotomy: a new technique to correct complex diaphyseal malunion: surgical technique. J Bone Joint Surg Am. 2019;92(Suppl 1 Pt 2):158–75.

14. Krettek C, Miclau T, Schandelmaier P, Stephan C, Möhlmann U, Tscherne H. The mechanical effect of blocking screws ("Poller screws") in stabilizing tibia fractures with short proximal or distal fragments after insertion of small-diameter intramedullary nails. J Orthop Trauma. 1999;13(8):550–3.
15. Henderson CE, Kuhl LL, Fitzpatrick DC, Marsh JL. Locking plates for distal femur fractures: is there a problem with fracture healing? J Orthop Trauma. 2011;25(Suppl 1):S8–14.
16. Hora N, Markel DC, Haynes A, Grimm MJ. Biomechanical analysis of supracondylar femoral fractures fixed with modern retrograde intramedullary nails. J Orthop Trauma. 1999;13(8):539–44.
17. Kwon GJ, Jo ML, Oh JK, Lee SJ. Effects of screw configuration on biomechanical stability during extra-articular complex fracture fixation of the distal femur treated with locking compression plate. J Biomed Eng Res. 2010;31(3):199–209.

Use in Arthrodesis

20

Kyle M. Schweser and Brett D. Crist

Introduction

Arthrodesis is usually a salvage procedure, with many patients having a previous injury and/or surgery. A patient undergoing arthrodesis may have several issues that can complicate a successful fusion, including necrotic bone (i.e., talar body avascular necrosis [AVN]), prior infection, poor soft tissue quality, and poor vascularity. Obtaining a successful fusion is predicated on maximizing the biomechanical properties necessary for fusion, while limiting potential complications. Extramedullary arthrodesis (plates, circular external fixators) and intramedullary arthrodesis (intramedullary nails [IMN]) are viable options. However, an issue with extramedullary fusion—especially plate fixation around the knee and ankle—is the limited or damaged soft tissue envelope. Hardware around the ankle can be prominent, and if wound breakdown occurs, then the plate can be exposed, leading to compromised fixation, infection, soft tissue coverage procedures, or even eventual amputation. Intramedullary arthrodesis with a nail can help negate some of those issues by limiting dissection and the amount of extramedullary hardware. Previous concerns about intramedullary arthrodesis included torsional control, and obtaining and maintaining compression. The technical skill required to reduce a joint successfully utilizing a nail can be more demanding than with plate fixation. This chapter will focus on the biomechanics of joint arthrodesis and why IMNs work.

Intramedullary arthrodesis is typically utilized in the knee and hindfoot due to the accommodating nature of the anatomy and the familiarity of utilizing intramedullary implants in both the femur and tibia. The literature and cases we will present show that the majority of intramedullary arthrodesis occurs in the hindfoot. There are several indications for the use of intramedullary knee arthrodesis, with the most common indication being a failed/infected total knee arthroplasty. Other indications for arthrodesis typically center on an inability to receive a total knee arthroplasty (increased risk of infection, arthrofibrosis, and poor soft tissue envelope). Indications for hindfoot fusion nails include two-joint arthritis/pathology, severe hindfoot trauma, osteonecrosis of the talus, severe malalignment deformities of the hindfoot, and Charcot arthropathy [1].

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Key Concepts for Intramedullary Arthrodesis

1. *Prepare the joint.*
   In order to achieve an arthrodesis, a critical step is to prepare the joint to create a compressible healing surface. Preparation of the joint is not limited to denuding the cartilage. The subchondral surface must be violated to increase the blood flow to the arthrodesis surface. The subchondral plate must be perforated, fish-scaled, or scored [2], but complete removal of the subchondral plate is avoided because it decreases the load to failure [3]. The ultimate goal is to create a compressible healing environment similar to a simple fracture in order to stimulate primary bone healing. Removing the subchondral bone exposes cancellous bone that cannot withstand compression as well as subchondral bone.
2. *Maximize the surface area for fusion.*
   The larger the arthrodesis surface contact area, the greater the distribution of load on the bone at the level of the fusion surface and the decreased risk of fatigue failure of the IMN locking mechanism. When there is bony contact, the IMN is a load-sharing device, rather than a load-bearing device, and any increase in stability at the arthrodesis site translates directly to decreased stress seen on the implant. The decreased stress at the screw/bone interface is important for maintaining compression, which will be discussed later, as increased stress can lead to screw loosening or breakage. Equal distribution of contact is also important. Any arthrodesis surface that is asymmetric will lead to the elevated surfaces experiencing the most force during compression and can lead to bony resorption or angular displacement [2, 4]. Maintaining the natural shape of the joint, if possible, has also been shown to be more biomechanically stable than completely flat surfaces [2].
3. *Compression—obtain and maintain.*
   The benefit of utilizng a plate for arthrodesis is the ability to compress the arthrodesis with multiple techniques and increase stability by adding more screws. As stated above, you are creating an environment similar to a simple fracture and attempting to achieve primary bone healing–the optimal choice for fusion. For fracture fixation, IMNs are usually thought to provide relative stability leading to secondary bone healing because the ability to generate compression is limited to cortical contact, the available locking mechanisms, and implant. This becomes especially difficult with a longer working length (knee arthrodesis) or limited bony contact (hindfoot and knee arthrodesis). The overall stiffness and fatigue resistance of the construct is not typically the issue and compares well to both plates and external fixators [5, 6].

   Compression, on the other hand, can be difficult. It is generally obtained via internal compression with a screw, or through external means (back slapping, application of external compression devices, etc.). The ability to create compression may be a contributing factor to the 11–40% nonunion rate in knee and hindfoot arthrodesis [5, 7–9]. The advent of fixed angle screws in IMNs has improved their ability to both obtain and maintain compression [10, 11]. Generating compression across an arthrodesis site is crucial to the initial biomechanical stability, as well as to the likelihood of eventual arthrodesis [11, 12]. Maintaining compression throughout the healing process is critical for stability and healing. Although there has been concern about the ability of certain hindfoot nails to maintain their compression compared to other fixation constructs [7, 13], the eventual fusion rates are acceptable [14]. A postoperative means to generate more compression is nail dynamization, either by using a single locking bolt in the diaphysis of the long bone of choice in the elliptical locking hole away from the arthrodesis surface, or not using any locking bolts in the diaphysis. When the patient weight bears, the bone moves around the nail and compresses the arthrodesis surface. However, this decreases the overall construct rigidity in torsion and axial stability, particularly if no locking bolts are used in the diaphysis, which may lead to early construct loosening and nonunion.
4. *Maintain mechanical alignment when possible.*
   Restoring the mechanical alignment of the joint in question allows the normal mechani-

cal forces to act on the fusion site, thus limiting stress on the implant. This will also increase the effectiveness of nail dynamization at the fusion site if utilized. IMNs are load-sharing devices along the axis of rotation when bony contact exists, so forces transmitted through the fusion site are shared with the nail and distributed evenly throughout the areas of bony contact. This is opposed to a load-bearing device, like a plate, which would asymmetrically distribute the load and experience more stress at the fusion site, especially during the bone resorption phase of healing.

5. *Promote primary bone healing.*
   The goal of arthrodesis is a solid fusion, with that fusion occurring through primary bone healing. A contributing factor is the use of a rigid construct that allows enough micromotion to keep the strain below 2% [15]. Hindfoot IMNs are superior to crossed screws and equivalent to other methods of fixation in terms of initial construct stability [6, 16–18]. In the knee, external fixation and IMNs are more common than plating and tend to be the methods of choice, especially in the setting of infection. Limited data is available to compare arthrodesis IMNs biomechanically to other constructs [19]. However, IMNs are typically known for promotion of secondary healing and promote increased fusion masses in other areas of the body [20]. The greatest risk for loss of stability and compression is during the resorptive phase of bone healing. While nails are inferior at maintaining compression during this phase, they are well suited to withstand the stresses during this phase and are associated with less overall stress shielding [7, 11].
6. *Associated bone loss and secondary healing.*
   In certain cases, direct bony apposition of the arhrodesis surfaces cannot be obtained without augmentation or limb shortening. While some limb shortening is tolerated and may be desired for limb clearance during gait, a significant shortening of the limb is not tolerated. Therefore, every effort should be made to maintain the limb length within 1 cm of the uneffected limb. If there is bone loss, augmentation can be considered. Not all cases that require augmentation are candidates for bulk allograft or metal augments, especially in the setting of infection. While not optimal, in these situations, intramedullary fusion nails are well suited for secondary bony fusion through callus formation and can still be successful. The working length of the nail can be increased by increasing the nail length (for example, spanning from hip to ankle if necessary for a knee fusion) to promote secondary healing while still maintaining stability. IMNs can bridge arthrodesis gaps and promote callus formation, while resisting the detrimental forces placed on the extremity [21, 22]. In cases like these, the joint surface should be prepared to promote a bone healing environment—debriding back to viable/bleeding bone, using bone graft with or without metal augments, while maintaining the strain at the arthrodesis site between 2% and 10% [15].

The ability to obtain—and especially maintain—compression at the fusion site has been proven difficult in bone gap models [11]. Gap healing differs from primary bone healing in that union and Haversian remodeling do not occur simultaneously [15]. Usually, the gap must be less than 1 mm and can account for the delay in appearance of bony union due to increased strain when there is a fixed gap. If the gap is not fixed, and the surfaces due appose with stress, there is still a chance for union. Secondary bone healing occurs when the gap is larger and flexible so the fusion site can maintain the appropriate strain. The loss of compression at the fusion site, if occurring early enough, would likely necessitate secondary healing to be successful.

When dealing with large gaps that are not amenable to secondary healing with auto- or allograft, metal augments or mesh spine cages are useful adjuncts around the nail. The cannulation of the augments allow the nail to pass through the center. Several augments exist, including mesh spine cages and porous metal augments utilized in revision total knee arthroplasty surgery. Mesh spine cages can be filled with osteoinductive or osteoconductive material, and their mesh design allows the inflow of host nutrients. They also work in tandem with nails in terms of their load-sharing properties and ability to transmit loads between both the proximal and distal segments if cortical contact can be obtained [23, 24]. Porous metal augments used in revision total knee arthroplasty have a biomechanical profile similar to bone. This means that they are strong enough to undergo compression and promote bony ingrowth while limiting the amount of stress shielding and stress concentration [25]. They can also be combined to fill larger defects [29]. Trabecular metal also carries a coefficient of friction higher than that of allograft or autograft, which imparts added stability to the initial construct [26] when there is bony contact. These are still combined with osteoinductive material, like autograft, to promote bony healing. An alternative to using augments or limb shortening may be using limb lengthening IMNs to maintain limb length and achieve arthrodesis. Please see the "Intramedullary Lengthening and Compression Nails" chapter for details.

Case 1

A 50-year-old female involved in a high-speed motor vehicle crash years prior presented with foot drop and failed previous nonoperative management. She had several tendon transfer procedures performed, but her foot drop returned quickly after surgery and she required an ankle-foot orthosis (AFO). Her ankle remained mobile since she was compliant with her home exercise program (Fig. 20.1a, b). After discussing surgical options, she elected to proceed with hindfoot fusion in order to ambulate without the use of an AFO (Fig. 20.1c, d).

Why This Works

The patient's recalcitrant foot drop after several tendon transfer procedures was multifactorial, and a repeat tendon transfer would have likely been unsuccessful. The options for her were to continue nonoperative management with an AFO and other braces, or proceed with surgical intervention. A hindfoot fusion was selected in this case to provide a stable platform for ambulation. The patient did not have any significant arthritis, so subchondral sclerosis was limited. The patient's cartilage was denuded and scored to increase the healing surface area, while resisting compressive loads. Her anatomic alignment and normal bony anatomy allowed for a large healing surface to increase the ability to achieve compression and stability at the arthrodesis site, leading to successful primary bony union. Using the fibula as a bone plate (fixed with two screws) increased stability and resisted lateral translation of the hindfoot. The procedure decreased the stress seen by the implant and increased the likelihood of maintaining compression.

Case 2

A 24-year-old male sustained an open talar neck fracture-dislocation that required multiple surgeries, including open reduction and internal fixation (ORIF) and soft tissue coverage. Over the following year, he experienced difficulty with weight-bearing secondary to pain. Conservative treatment with medication and bracing failed to provide enough pain relief to allow for activities of daily living. Radiographs and advanced imaging showed avascular necrosis of the talus with collapse and fracture (Fig. 20.2a, b). The patient was eventually treated with a tibiotalocalcaneal fusion utilizing a hindfoot fusion nail, complete talectomy, femoral head allograft, bone marrow aspirate concentrate (BMAC), and fibular autograft. He went on to eventual fusion and pain-free ambulation with shoe modification (Fig. 20.2c, d).

Why This Works

In a young patient, avascular necrosis of the talus is a devastating problem. Some patients can be managed with conservative therapy (rigid AFOs, anti-inflammatories, etc.); however, management after failed conservative therapy is limited almost exclusively to arthrodesis. Both the tibiotalar and subtalar joints are effected, making a hindfoot fusion nail an excellent choice. In this particular case, the patient also had a large soft tissue injury affecting the anterior, medial, and posterior aspect of his ankle, so limiting dissection through those areas was critical to success. The ability to create a stable support bed for the arthrodesis was also critical.

Fig. 20.1 Preoperative radiographs of an ankle with normal bony anatomy and neutral alignment (**a**, **b**). Short-term follow-up after a hindfoot fusion, demonstrating early signs of tibiotalar and subtalar fusion (**c**, **d**)

This patient's talus did not provide the proper environment for fusion because its structure has been significantly altered by the avascular necrosis and the bone was dead. The talar body had to be removed and replaced with a structural allograft or autograft to increase construct stability. With a defect this large, obtaining enough structural autograft is difficult. An excellent option is the use of femoral head allograft. The femoral head is strong enough to support compression, maintain limb length, and large enough to accommodate an intramedullary implant without fracturing during preparation and implantation [27]. It is usually perforated several times and can be supplemented with viable osteogenic cells and signal (i.e., bone marrow aspirate concentrate (BMAC)) to stimulate bony ingrowth and support bone healing. One of the challenges with this particular method is achieving proper alignment of the limb. Achieving normal mechanical alignment of the hindfoot is crucial for proper weight-bearing through the effected joint, leading not only to improved function but also increased chances of successful fusion by limiting the load seen by the IMN.

Fig. 20.2 Radiographs one year after ORIF of his open talar neck fracture-dislocation demonstrating a sclerotic talar body with subchondral collapse (**a**, **b**). One-year follow-up radiographs demonstrating a successful hindfoot fusion after complete talectomy and bulk allograft insertion (**c**, **d**)

Case 3

A 58-year-old female had sustained multiple injuries, including a right open pilon and fibula fracture with significant central plafond bone loss, a contralateral pilon and calcaneus fracture, and a pelvic ring injury (Fig. 20.3a, b). A significant defect over the fibula was present that would have required flap coverage. After an extensive discussion with the patient about her central bone loss and significant soft tissue trauma, a primary fusion of her hindfoot was selected. The tibiotalar joint was approached through a transfibular approach. This was done to allow for the fibula to be used as bone graft in the tibiotalar joint and as a lateral buttress/support and to decompress the lateral ankle, allowing for primary wound closure. A vascularized fibular onlay graft was utilized, with the medial portion of the fibula used to graft the central defect (Fig. 20.3c, d). She went on to heal her lateral wound, successfully fuse her tibiotalar joint, and have an asymptomatic subtalar nonunion (Fig. 20.3e, f).

Why This Works

Acute fusion for comminuted pilon fractures has been shown to be an effective treatment method [28]. Something to consider in this case is the central tibial bone loss and the limitation in terms of joint compression. Other biomechanical principles must be closely adhered to in order to obtain a successful result. To overcome the lack of compression, autograft was added to the central portion of the joint, but compression remained limited and bulk allograft was deemed inappropriate secondary to the open injury and risk of infection. The joint was meticulously prepared, ensuring preservation of as much healthy subchondral bone as possible. The subchondral bone was scored and perforated to increase the surface area and promote stimulation by growth factors. As can be seen in the postoperative images (see Fig. 20.3e, f), compression led to some proximal migration of the talus with minimal subtalar compression. For the tibiotalar joint, the normal architecture of the bone was maintained, and the talus was situated in the fracture site to increase the surface area available for fusion. The overall alignment of the lower extremity was maintained to limit stresses on the implant, and allow for anatomical mechanical forces to act on the fusion site. A fibular onlay graft was utilized to increase the fusion surface and to increase the stability of the deficient tibiotalar joint.

Fig. 20.3 Injury radiograph and computed tomography (CT) scan demonstrating the initial injury with significant anterolateral and central plafond bone loss (**a**, **b**). Intraoperative fluoroscopy demonstrating reduction of the talus into the central defect with maximum coaptation of the talus and plafond (**c**, **d**). Long-term follow-up radiographs demonstrating complete fusion of the tibiotalar joint with a subtalar nonunion that was asymptomatic (**e**, **f**)

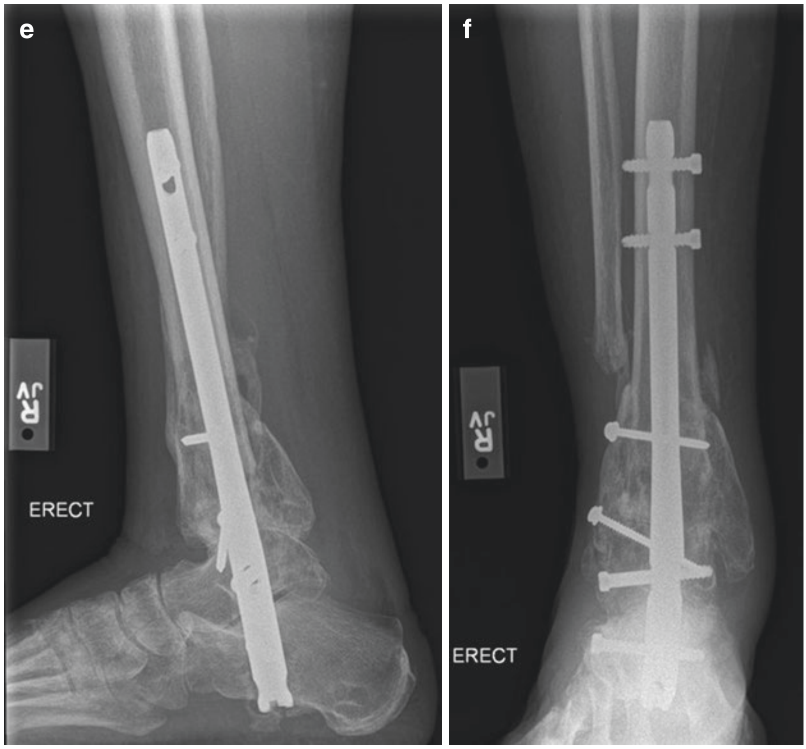

Fig. 20.3 (continued)

Case 4

A 66-year-old female presented to clinic with complaints of ankle pain, swelling, and instability. She had previously undergone a total ankle replacement after a failed ORIF several years prior. Radiographs revealed a failed total ankle with subsidence and distal tibiofibular synostosis (Fig. 20.4a, b). After a negative infection workup, a hindfoot fusion was performed utilizing femoral head allograft for bone defect management (Fig. 20.4c, d). She subsequently went on to a successful fusion with pain-free ambulation (Fig. 20.4e, f).

Why This Works

A failed total ankle arthroplasty can be a difficult problem to manage, and infection should always be ruled out before any intervention. One of the biggest issues is the substantial loss of tibiotalar bone stock. When this occurs, you can sacrifice limb length and attempt to fuse the calcaneus to the remaining tibia, or you can place a bulk allograft or metal augment. Both of these options attempt to increase the surface area and provide some compressive strength to achieve fusion. In this case, a bulk allograft was used to maintain limb length due to the synostosis of the distal tibiofibular joint, which could have caused impingement issues with a shortened limb. Similar to the previous bulk allograft case, the femoral head must be prepared like a joint surface, with care taken to preserve the underlying bony architecture to allow for compression. This includes perforating or scoring the surface and augmenting with growth factors, autograft, etc., in order to promote bony ingrowth. Attempts should be made to get as congruous a fit as possible to increase stability. In this case, acetabular reamers were used to create a surface that would accommodate the femoral head allograft, creating apposition between the remaining bone and allograft, further increasing the stability and lowering the strain placed on the implant.

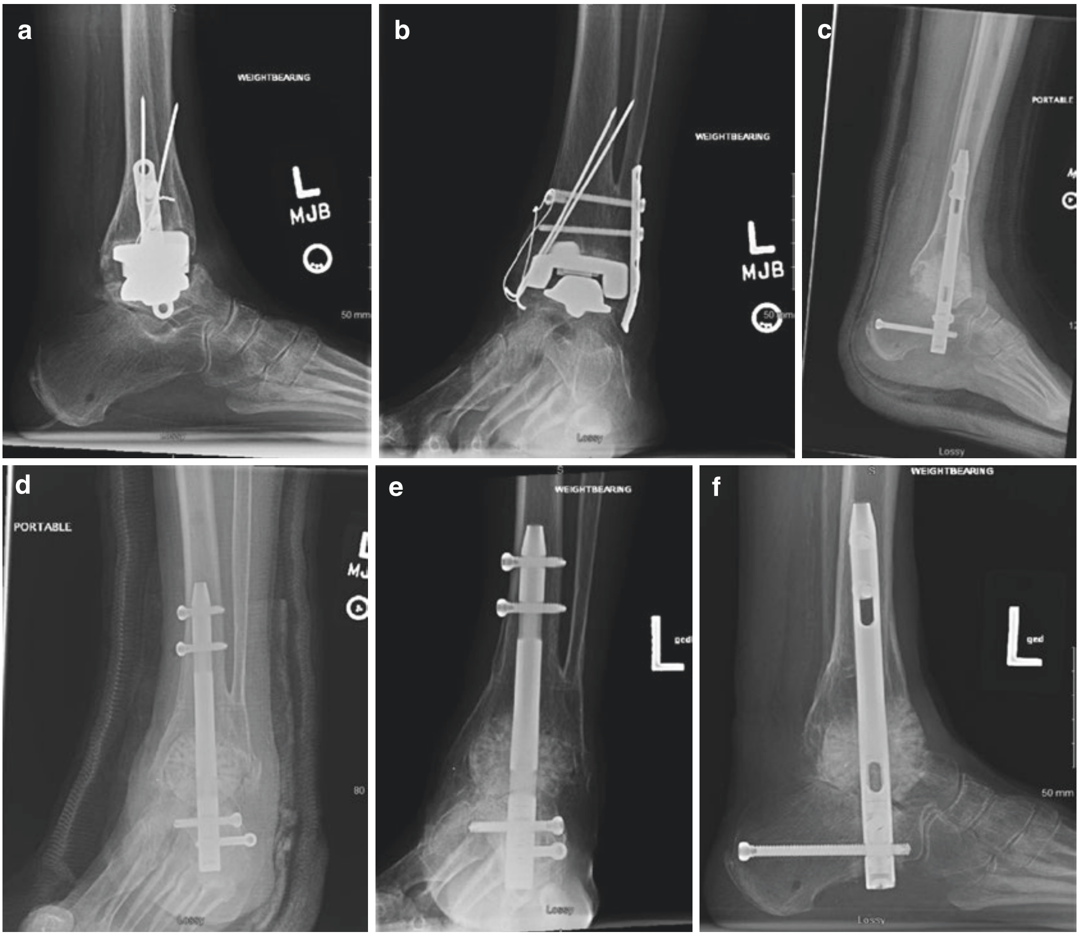

Fig. 20.4 Preoperative radiographs of a failed total ankle arthroplasty (**a**, **b**). Immediate postoperative radiographs after a hindfoot fusion nail with utilization of bulk allograft (**c**, **d**). Long-term follow-up demonstrating incorporation of the bulk allograft and complete fusion (**e**, **f**)

Case 5

A patient is a 56-year-old male with a recalcitrant infected total knee arthroplasty that was referred after explant with antibiotic spacer placement (Fig. 20.5a, b). After a discussion with the patient, he elected to undergo a knee fusion as opposed to reimplantation. He had minimal bone loss, but primary bony apposition was not possible without shortening due to the previous total knee components. A large allograft or metal augment was avoided due to the history of recalcitrant infections. Taking these factors into account, he underwent a fusion utilizing a long IMN with autograft supplementation (Fig. 20.5c–e). Short-term follow-up for the patient demonstrated graft incorporation and callus formation (Fig. 20.5f).

Why This Works

In this particular case, bony contact was not achieved for several reasons; thus, other biomechanical principles had to be followed. Bony contact was not necessary because the gap that needed to be bridged was relatively small and easily achieved via callus formation, and the principles utilized for long bone nailing could effectively be applied. A bone healing environment was obtained by joint preparation and incorporation of autograft

Fig. 20.5 Preoperative radiographs of the knee demonstrating the antibiotic spacer placement (**a**, **b**). Immediate postoperative radiographs demonstrating the long fusion nail with Reamer/Irrigator/Aspirator (Depuy Synthes, West Chester, PA) obtained autograft (**c–e**). Short-term follow-up radiographs demonstrating early graft consolidation and fusion (**f**)

utilizing the Reamer/Irrigator/Aspirator (RIA) (DePuy Synthes, West Chester, PA). Due to the lack of bony contact, the goal of this surgery was to obtain secondary bone healing through callus, meaning that a completely rigid construct would be disadvantageous. In order to promote callus formation and the correct amount of strain at the fusion site, the working length was maximized, and alignment was maintained. By increasing the working length, while still providing a stable construct, we were able to appropriately decrease the strain at the fusion site while minimizing the stress on the implant. Mechanical alignment was important to transfer uniform loads to the nail and equal distribution of implant stress.

Case 6

A 21-year-old male involved in a high-speed motor vehicle collision sustained an open intra-articular distal femur and patella fracture with loss of his lateral femoral condyle at the accident. An initial injury radiograph was performed in the trauma bay (Fig. 20.6a). He was initially treated with a knee-spanning external fixator, patellectomy, and antibiotic spacer. Figure 20.6b, c shows the postoperative radiographs after irrigation and debridement, external fixation, and placement of an antibiotic spacer into the lateral defect. After several operative debridements and antibiotic spacer exchanges, the patient's options were discussed with him, and he elected to undergo a knee fusion. He had significant bone loss laterally, but good bone stock medially. His medial femoral condyle was repaired, and an arthrodesis nail was placed with medullary autograft obtained with the RIA system, and several bone morphogenetic protein 2 (BMP-2)-soaked sponges (Infuse, Medtronic, Minneappolis, MN) used to fill the lateral bone defect. Figure 20.6d–f shows the postoperative radiographs after medial condylar fixation, bone grafting of the lateral defect, and placement of an intramedullary fusion nail. The patient went on to a successful fusion and returned to work. Figure 20.6g, h reveals long-term follow-up demonstrating successful fusion.

Why This Works

Fusions are typically salvage procedures, with every case unique, and one must utilize what is given to maximize the ability to achieve a fusion. In this particular case, the patient had a significant lateral femoral condyle bone defect after a severe open injury. His medial femoral condyle, however, was intact and could help maintain limb length and serve as a compressible surface if made stable. This case utilizes several principles to achieve fusion in a unique way. First, the joint surfaces that remained were stabilized (via ORIF of the medial femoral condyle) and then prepared by removing the cartilage and penetrating the subchondral plate. This provided bony contact and a compressible surface to impart stability to the fusion site, thus attempting to achieve the goal of primary bony fusion. However, the loss of bone laterally was addressed by impacting bone graft and BMP-2 sponges into the site, thus limiting the compressibility. In this particular case, providing a rigid enough construct for primary bony fusion medially would have been detrimental to the secondary bony fusion required laterally secondary to the strain mismatch. In order to achieve fusion, the entire fusion site would have to undergo secondary bone healing. This was achieved by maximizing the working length of the nail while still maintaining stability. The preserved medial condyle imparted stability to the fusion site, and relieved strain felt at the nail/fusion interface, to increase the bony surface area for fusion, and to provide support for the impacted bone graft. The fusion site was optimized in terms of creating a fracture environment with the use of subchondral penetration, BMP-2, and medullary bone graft obtained with the RIA system.

Case 7

A 65-year-old female presented to the emergency room after a ground level fall with an open distal third tibia fracture with no intra-articular involvement on computed tomography (CT) scan. She

Fig. 20.6 Initial trauma bay radiograph demonstrating the open distal femoral injury with bone loss (**a**). Eventually, the external fixator and antibiotic spacer (**b**, **c**) were converted to an open reduction internal fixation with intramedullary fusion nail placement and bone grafting (**d**–**f**). Follow-up radiographs demonstrate successful fusion (**g**, **h**)

Fig. 20.7 Anteroposterior view of the tibia demonstrating failure of the nail (**a**) and axial computed tomography (CT) scan of the joint (**b**). The patient underwent revision ORIF (**c**) with subsequent infection (**d**) and spacer placement (**e**). Eventually, she underwent a hindfoot fusion with a metal augment for bone defect management (**f**, **g**)

underwent an uneventful intramedullary nailing. At her 6-week follow-up appointment, it was found that she had sustained a fracture around the implant with nail penetration into the tibiotalar joint (Fig. 20.7a, b). She underwent a subsequent revision ORIF of her tibial shaft fracture, as well as ORIF of her new pilon fracture (Fig. 20.7c). She subsequently developed an infection with wound dehiscence and tendon exposure, eventually requiring explant of the hardware and antibiotic spacer placement (Fig. 20.7d, e). Eventually she underwent a staged hindfoot fusion with a trabecular metal augment to address the bony defect (Fig. 20.7f, g).

Why This Works

Infections around the ankle can be difficult to manage and often result in bone loss. As discussed above, the bone loss can be managed with bulk allograft; however, an alternative is the utilization of metal augments. In this case, fusion was obtained by observing several principles for primary bone healing. The trabecular metal augment allows direct compression of the unopposed bone edges while also maintaining normal anatomic length and alignment. It is also cannulated to accept the IMN. This construct is very rigid, especially with the increased friction imparted by the trabecular metal design. Normally, an overly rigid construct would result in stress shielding and potentially delayed union. However, in this case, the metal augment has a modulus of elasticity similar to bone. Due to this similarity, the fusion area should respond in the same fashion as a standard nail with bony coaptation. Preparation of the fusion site is done with autograft supplement, like RIA or BMAC, as discussed above. The metal augment also allows dispersion of the forces at the fusion site minimizing concentration of stress at a single point, thus limiting the amount of stress seen at both the implant and the bone/augment interface.

Conclusion

Intramedullary nailing is a viable and effective option for achieving arthrodesis when certain principles are maintained. The point of intramedullary nailing is to preserve soft tissue while providing a stable environment for bone healing. The same applies when used for fusion. Both primary and secondary bony fusion can be achieved with IMNs, and certain biomechanical principles should be observed based on the type of bone healing desired. When primary bone healing is the goal, a stable and rigid construct with bony contact and compression should be obtained. Compression is paramount to achieving primary fusion and can be obtained through internal or external means. Bony contact and coaptation will be the primary method of achieving stability of any construct to decrease the stress on the implant. Maintaining the mechanical axis and allowing compression through a relatively normal anatomic alignment will also alleviate detrimental forces on the implant, thus limiting hardware failure and abnormal strain at the arthrodesis site. If good coaptation and adequate bony contact cannot be obtained or maintained, then secondary bony fusion should be the goal. Secondary fusion is where intramedullary nailing has an advantage over plates, because the intramedullary device is along the axis of rotation within the long bone, which decreases the moment arm and subsequently the bending forces seen during axial load. This leads to improved fatigue resistance and allows the stress of fracture gaps to be withstood, while still maintaining the appropriate strain environment to promote bone healing. Increasing the working length of the nail and maximizing the local arthrodesis environment are important principles in achieving fusion via secondary means.

References

1. Thomas RL, Sathe V, Habib SI. The use of intramedullary nails in tibiotalocalcaneal arthrodesis. J Am Acad Orthop Surg. 2012;20(1):1–7.
2. Parker L, Singh D. (i) The principles of foot and ankle arthrodesis (Mini-symposium foot and ankle). Orthop Trauma. 2009;23(6):385–94.
3. Ray RG, Ching RP, Christensen JC, Hansen ST Jr. Biomechanical analysis of the first metatarsocuneiform arthrodesis. J Foot Ankle Surg. 1998;37(5):376–85.
4. Kowalski RJ, Ferrara LA, Benzel EC. Biomechanics of bone fusion. Neurosurg Focus. 2001;10(4):E2.
5. Miller SD. Compression forces of internal and external ankle fixation devices with simulated bone resorption. Foot Ankle Int. 2010;31(5):469–70; author reply 70–1.
6. Fragomen AT, Meyers KN, Davis N, Shu H, Wright T, Rozbruch SR. A biomechanical comparison of micromotion after ankle fusion using 2 fixation techniques: intramedullary arthrodesis nail or Ilizarov external fixator. Foot Ankle Int. 2008;29(3):334–41.
7. McCormick JJ, Li X, Weiss DR, Billiar KL, Wixted JJ. Biomechanical investigation of a novel ratcheting arthrodesis nail. J Orthop Surg Res. 2010;5:74.
8. Berson L, McGarvey WC, Clanton TO. Evaluation of compression in intramedullary hindfoot arthrodesis. Foot Ankle Int. 2002;23(11):992–5.
9. Mückley T, Hoffmeier K, Klos K, Petrovitch A, von Oldenburg G, Hofmann GO. Angle-stable and compressed angle-stable locking for tibiotalocalcaneal arthrodesis with retrograde intramedullary nails. Biomechanical evaluation. J Bone Joint Surg Am. 2008;90(3):620–7.
10. Kaspar K, Schell H, Seebeck P, Thompson MS, Schutz M, Haas NP, et al. Angle stable locking reduces interfragmentary movements and promotes healing after unreamed nailing. Study of a displaced osteotomy model in sheep tibiae. J Bone Joint Surg Am. 2005;87(9):2028–37.
11. Woods JB, Burns PR. Advances in intramedullary nail fixation in foot and ankle surgery. Clin Podiatr Med Surg. 2011;28(4):633–48.
12. Mückley T, Eichorn S, Hoffmeier K, von Oldenburg G, Speitling A, Hoffmann GO, Bühren V. Biomechanical evaluation of primary stiffness of tibiotalocalcaneal

fusion with intramedullary nails. Foot Ankle Int. 2007;28(2):224–31.
13. Yakacki CM, Khalil HF, Dixon SA, Gall K, Pacaccio DJ. Compression forces of internal and external ankle fixation devices with simulated bone resorption. Foot Ankle Int. 2010;31(1):76–85.
14. Griffin MJ, Coughlin MJ. Evaluation of midterm results of the Panta nail: an active compression tibiotalocalcaneal arthrodesis device. J Footo Ankle Surg. 2018;57(1):74–80.
15. Marsell R, Einhorn TA. The biology of fracture healing. Injury. 2011;42(6):551–5.
16. Berend ME, Glisson RR, Nunley JA. A biomechanical comparison of intramedullary nail and crossed lag screw fixation for tibiotalocalcaneal arthrodesis. Foot Ankle Int. 1997;18(10):639–43.
17. Alfahd U, Roth SE, Stephen D, Whyne CM. Biomechanical comparison of intramedullary nail and blade plate fixation for tibiotalocalcaneal arthrodesis. J Orthop Trauma. 2005;19(10):703–8.
18. Froelich J, Idusuyi OB, Clark D, Kogler GF, Paliwal M, Dyrstad B, et al. Torsional stiffness of an intramedullary nail versus blade plate fixation for tibiotalocalcaneal arthrodesis: a biomechanical study. J Surg Orthop Adv. 2010;19(2):109–13.
19. Kim K, Snir N, Schwarzkopf R. Modern techniques in knee arthrodesis. Int J Orthop. 2016;3(1):487–96.
20. Peterson JM, Chlebek C, Clough AM, Wells AK, Batzinger KE, Houston JM, et al. Stiffness matters: Part II – The effects of plate stiffness on load-sharing and the progression of fusion following ACDF in vivo. Spine (Phila Pa 1976). 2018;43(18):E1069–76. https://doi.org/10.1097/BRS.0000000000002644.
21. Bong MR, Kummer FJ, Koval KJ, Egol KA. Intramedullary nailing of the lower extremity: biomechanics and biology. J Am Acad Orthop Surg. 2007;15(2):97–106.
22. Schneider E, Michel MC, Genge M, Zuber K, Ganz R, Perren SM. Loads acting in an intramedullary nail during fracture healing in the human femur. J Biomech. 2001;34(7):849–57.
23. Lindsey RW, Gugala Z, Milne E, Sun M, Gannon FH, Latta LL. The efficacy of cylindrical titanium mesh cage for the reconstruction of a critical-size canine segmental femoral diaphyseal defect. J Orthop Res. 2006;24(7):1438–53.
24. Lindsey RW, Gugala Z. Cylindrical titanium mesh cage for the reconstruction of long bone defects. Osteo Trauma Care. 2004;12(3):108–15.
25. Frigg A, Dougall H, Boyd S, Nigg B. Can porous tantalum be used to achieve ankle and subtalar arthrodesis? A pilot study. Clin Orthop Relat Res. 2010;468(1):209–16.
26. Henricson A, Rydholm U. Use of a trabecular metal implant in ankle arthrodesis after failed total ankle replacement: a short-term follow-up of 13 patients. Acta Orthop. 2010;81(6):745–7.
27. Bussewitz B, DeVries JG, Dujela M, McAlister JE, Hyer CF, Berlet GC. Retrograde intramedullary nail with femoral head allograft for large deficit tibiotalocalcaneal arthrodesis. Foot Ankle Int. 2014;35(7):706–11.
28. Mauffrey C, Zagrocki L, Jordan RW, Seligson D. Retrograde tibiotalocalcaneal nails: an option for complex open pilon fractures. Current Orthop Pract. 2012;23(5):507–11.
29. Blake Peterson, Sonny Bal, Ajay Aggarwal, Brett Crist, (2016) Novel Technique: Knee Arthrodesis Using Trabecular Metal Cones with Intramedullary Nailing and Intramedullary Autograft. The Journal of Knee Surgery. 29(06):510–15.

21 Intramedullary Lengthening and Compression Nails

Austin T. Fragomen, Mitchell Bernstein, and S. Robert Rozbruch

Introduction

The recipe for limb lengthening through distraction osteogenesis was founded on a marriage between a biologically optimized osteotomy technique and a mechanically sound circular external fixation system that attached to the skeleton with fine, tensioned wires [1]. While this technique was exhilarating and solved a myriad of orthopedic enigmas, patients undergoing treatment experienced pain, pin infections, and the awkwardness of life wearing a cumbersome external fixator. Surgeons looked toward a future with internal lengthening nails, but early designs required more faith than skill [2] and were inferior to external-fixator-assisted methods [3]. The emergence of the FITBONE® (Wittenstein, Igersheim, Germany) lengthening nail fundamentally changed limb lengthening surgery by guaranteeing total rate control, but this implant was only available in select centers. The PRECICE® magnetic intramedullary lengthening nail (NuVasive, San Diego, CA, USA) soon followed. With its ease of insertion and ability to move in both directions, the PRECICE® implant was well received among the international limb lengthening community. While studies demonstrate that both the FITBONE® and PRECICE® implants are associated with rapid healing and superior control over lengthening, they both experience similar complications [4]. This chapter will focus exclusively on the PRECICE® implant, from which these authors can draw on their combined experience.

The P1 (first generation) PRECICE® prototype was a modular system with titanium outside, stainless-steel internal components, and rare earth metals used for the magnet. This design yielded excellent worldwide results, but encountered mechanical limitations including fracturing of the actuator pin (threaded rod in the center of the nail) and catastrophic failure of the nail at the welding seam (Table 21.1) [5–18]. The engineers responded quickly, releasing the P2 (second gen-

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Table 21.1 PRECICE® clinical performance

Author	Nail model	Bone	Goals achieved with initial surgery?	BHI (days/cm)	Mechanical complication (no.)
Fragomen [5]	P1 and P2	F-40	35/40 (88%)	30.5	PMC (2); varus regenerate and P2 crown failure (1); varus regenerate (1); P1 nail Fx (1); OL corrected with compression (1)
Iobst [6]	P1 and P2	F-27	93% LL 81% MA	42	PMC (1), distal femur flexion def >10 deg (1) Better alignment achieved with the use of 6-mm pins and > 2 blocking screws
Hammouda [7]	P2	H-6	6/6 (100%)	36	None
Furmetz [8]	P2	H-1	0/1 (0%)	–	P2 nail crown failure (1)
Hammouda [9]	P1 and P2	F-17	16/17 (94%) LL	32	PMC (2)
Hammouda [10]	P1	F-13	10/13 (77%)	–	Late FTD during lengthening (1), PMC (1)
Wagner [11]	P1	F-24 T-8	92%	36	None
Weibking [12]	P1	F-5 T-4	8/9 (89%)	33	P1 nail retracted (1), P1 nail retracted and Fx (1)
Karakoyun [13]	P1 and P2	F-21 T-6	26/27 (96%)	34	Nail Fx (1), OL corrected with compression (7)
Tiefenboeck [14]	–	F-5 T-5	6/10 (60%)	43	P2 nail crown failure with retraction (1), P2 nail crown failure with retraction and nail Fx (1)
Laubscher [15]	–	F-20	20/20 (100%)	31	Backing out of locking bolts (2)
Fragomen [16]	P2	F-9T-5	13/14 (93%)	–	None
Schiedel [17]	P1 only	F-20 T-6	22/26 (85%)	–	FTD (2), PMC (1), P1 nail Fx (2), less length than expected = low precision (10)
Kirane [18]	P1 and P2	F-17 T-8	86%	–	FTD (1), PMC (1)

P PRECICE®, *F* Femur, *T* Tibia, *H* Humerus, *BHI* Bone healing index, *PMC* Premature consolidation, *Fx* Fracture, *OL* Over-lengthening, *LL* Limb length, *MA* Mechanical axis, *FTD* Failure to distract

eration) model with a thicker actuator pin and a change from a two-piece construct to a continuous outer casting of titanium. Elimination of the weld in the P2 design produced a nail four times stronger in bending and three times stronger in axial loading [19]. The problem of nail fracture has been virtually eliminated, but fracture of the anti-rotation crown at the junction of the large barrel and the telescopic sections of the nail was reported [8, 20]. The P2.1 version strengthened the crown improving rotational control and preventing varus deformity at this junction [19], resulting in far fewer crown failures. The magnetic lengthening nail continues to evolve with the recent release of the Stryde nail (NuVasive, San Diego, CA, USA), made of a stainless-steel alloy with a stronger magnet and reinforced crown. Stryde promises to tolerate far more weight-bearing loads. For example, a PRECICE® 10.7-mm implant allows for 50 lbs weight-bearing, while the Stryde 10-mm implant allows for 150 lbs weight-bearing.

The PRECICE® implant has significantly improved the lengthening experience for patients [21]. It has been our observation that during lengthening with this device, narcotic use has decreased, and antibiotic use has dropped from 70% to 0% of cases when compared to external fixation. Patients treated at different times with both internal and external fixation reported reduced pain, improved physical therapy sessions, a better cosmetic result, and overall improved satisfaction after lengthening with the PRECICE® implant [21]. The bone healing index (BHI) defines the rate at which the lengthening bone is fully healed. It is typically reported as month or day per centimeter. The BHI for femoral lengthening with the PRECICE® is rapid with an average of 34 days/cm (see Table 21.1). However, limb lengthening with an intramedullary implant has some limitations: metaphyseal osteotomies will be difficult to control and are best performed as two separate osteotomies [22] (one in the metaphysis for deformity correction and a second in the diaphysis for lengthening), deformity correction needs to be acute and is without postoperative adjustability [6], and blocking screws need to be inserted with high accuracy [23].

Preoperative Assessment

All components of limb deformity must be determined, even if your patient is seeking treatment for what appears to be a simple limb length discrepancy (LLD). These include frontal, sagittal, and axial malalignment. In addition, range-of-motion and stability testing should be performed and compared to the contralateral side. This will aid in determining if any soft-tissue releases are necessary adjuncts to the surgical procedure (Fig. 21.1).

Radiographs should include orthogonal, calibrated images of the whole bone in question. In addition, a standing hip-to-ankle x-ray should be obtained (Fig. 21.2). Surgeons need to be aware of common deformity language and normal values of both anatomic and mechanical axes of the bone (Fig. 21.3).

Once a complete problem list is generated, the surgeon must have a comprehensive operative strategy. It is based on the following considerations, and each will be discussed in detail in the case presentations:

1. The bone to be lengthened (femur vs. tibia)
2. Anterograde versus retrograde: deformity location and thigh circumference
3. Starting point and trajectory
4. Size of the implant (note that current available diameters for PRECICE® femoral nails are 8.5, 10.7, and 12.5 mm)

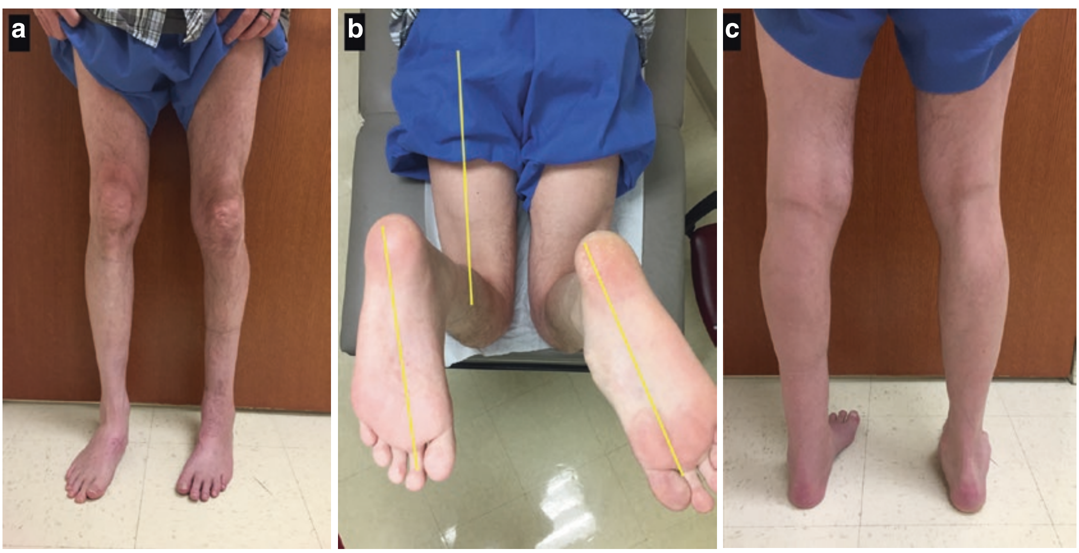


Fig. 21.1 (**a**) This is a patient who presented with a posttraumatic LLD from a remote injury. He sustained femur and tibia fractures that were treated with intramedullary rods. On clinical exam, in addition to his LLD, it was noted that he had an external rotation deformity of his femur and internal rotation deformity of his tibia. The patient is standing in shorts in order to examine the entire limb. The patella on the left side was slightly externally rotated as the patient ambulated. Note the position of the foot (internally rotated) (**b**). Thigh-foot-axis measurement demonstrated a tibia that was internally rotated. (**c**) Hindfoot varus on the left side was evident when examining the patient from the back view

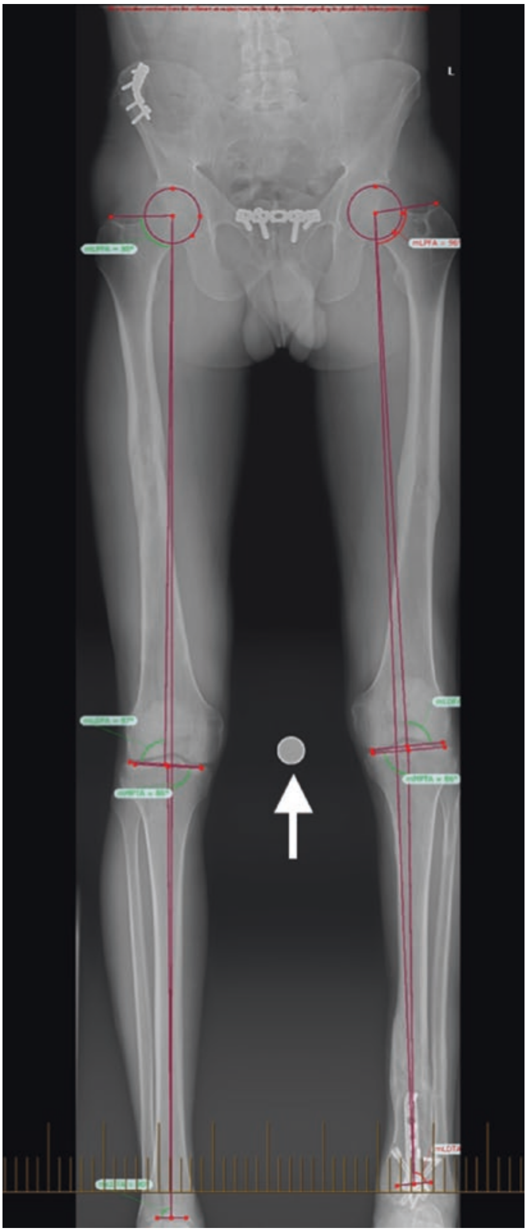


Fig. 21.2 Standing hip-to-ankle radiograph of a patient with 33-mm posttraumatic LLD originating from the femur and tibia. Note the calibration marker (*white arrow*) which is used to ensure accurate measurements. This is also useful when importing images into a computer-aided design (CAD) program to generate segment lengths and joint orientation angles

5. Nail length
6. The need for simultaneous rotational, sagittal, or coronal corrections
   (a) Blocking screw placement
   (b) Intraoperative external fixator-assisted nailing
7. Adjunct procedures
   (a) Iliotibial band release
   (b) Gastrocsoleus recession
   (c) Common peroneal or other nerve releases

Intraoperative Execution

Iobst et al. [6] showed that using half pins with intraoperative fixator-assisted nailing improved the accuracy of the nail placement and deformity correction [6]. Likewise using two or more blocking screws improved accuracy. The tibia can be approached using the infrapatellar or suprapatellar entry point [6]. The semi-extended position may improve accuracy of the nail insertion. Reamers may be rigid or flexible. The osteotomy technique that has been so successful in multiple series is the percutaneous, multiple drill hole, osteotome method. The drill holes are made prior to reaming and act as vent holes through which the reamings are deposited yielding excellent healing rates [4].

Postoperative Protocol

The latency period and rate of lengthening vary among surgeons and the particular long bone being treated. In general, the latency for femur lengthening is shorter than that for tibia lengthening surgery. The bones are distracted at 1 mm/day and adjusted based on postoperative radiographs. Weight-bearing is strictly prescribed by NuVasive and is based on the diameter of the nail used

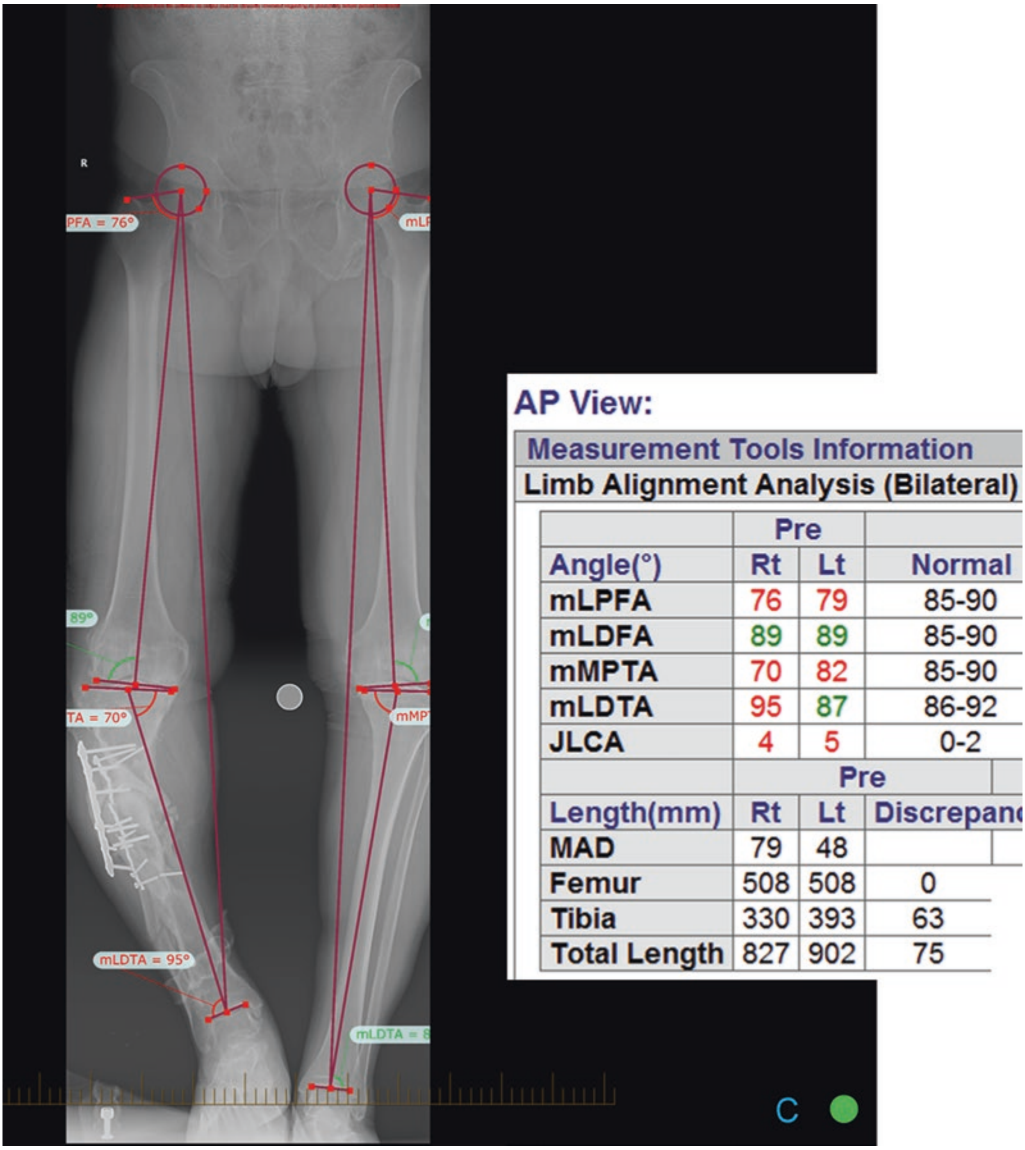


Fig. 21.3 Standing hip-to-ankle radiograph of a patient who has repeated attempts at achieving union after an open tibia fracture 12 years prior. Joint orientation angles and limb segment lengths are calculated as part of the pre-operative assessment

(Table 21.2). Excessive weight-bearing on an unconsolidated regenerate site has led to failures. For longer femur lengthening projects, the position of the magnet in the nail will move relative to the overlying skin as the bone distracts. This means that the skin marking made at the time of nail implantation will no longer be accurately positioned as lengthening proceeds and will need to be adjusted.

Table 21.2 Weight-bearing recommendations for PRECICE®

Nail diameter (mm)	In distraction Max weight-bearing (lbs)	In compression Max weight-bearing
8.5	30	–
10.7	50	WBAT[a]
12.5	70	WBAT[a]

[a]Weight-bearing as tolerated (WBAT) ambulation has not been advocated by NuVasive but has been safe in our practice

Case 1

A 47-year-old male was involved in a motor-vehicle accident. He sustained a closed right femur fracture that had been treated in traction. He presented with knee pain. Imaging demonstrated varus alignment, medial compartment knee arthritis, limb shortening, and femoral malunion. The treatment strategy consisted of using a piriformis entry internal limb lengthening rod to correct femoral varus, procurvatum, and shortening. This was followed by a staged proximal tibial osteotomy to correct residual tibial varus. Piriformis entry was chosen since it allowed for intuitive anatomic axis planning in the femur (Figs. 21.4, 21.5, 21.6, 21.7, 21.8, 21.9, and 21.10) [24].

Why This Worked

This worked well due to concurrent limb realignment and limb lengthening. In this case, using a piriformis entry IM rod allowed for access into the anatomic axis of the femur. Anatomic axis realignment in this case allowed for moving the mechanical axis line of the limb into a normal position.

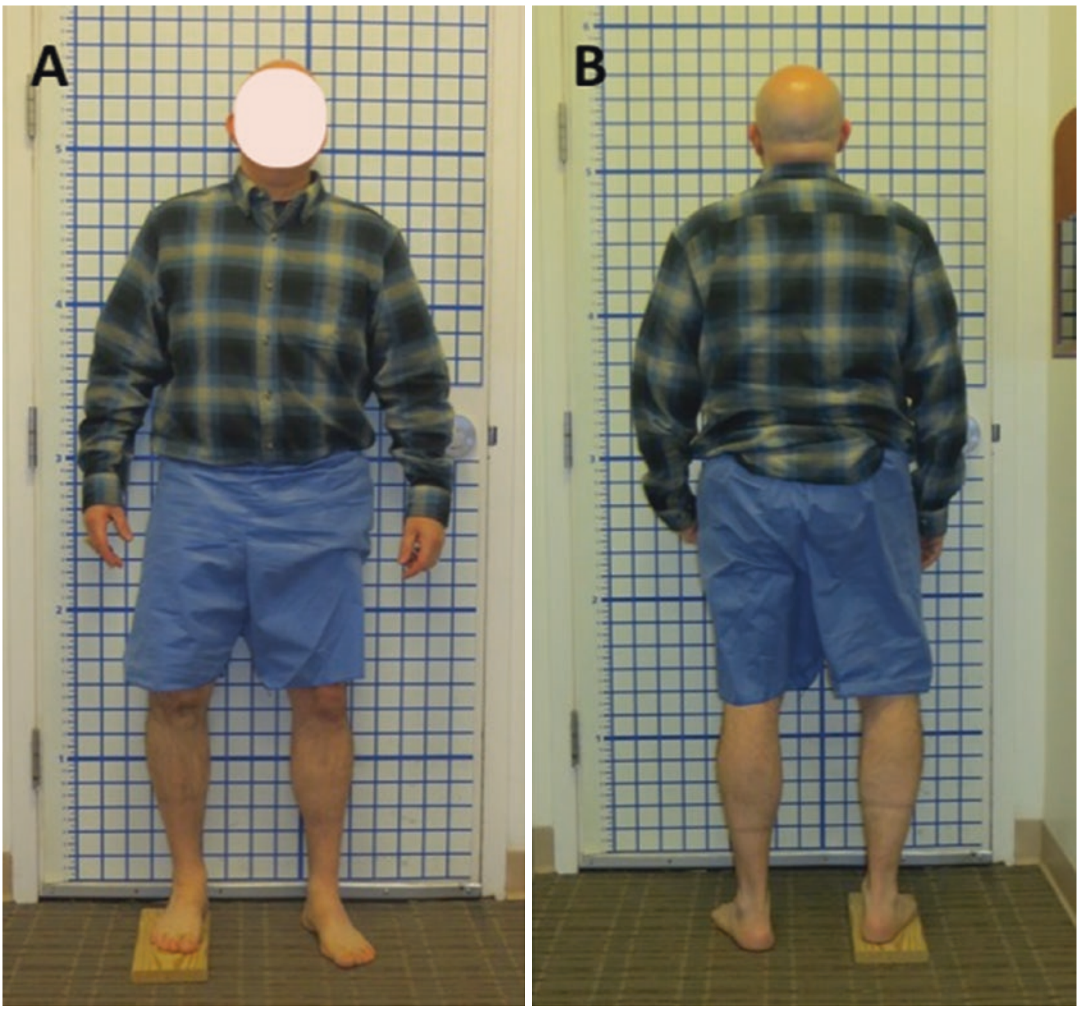


Fig. 21.4 Preoperative clinical photos of the patient from the front (**a**) and back (**b**). Varus alignment and shortening are noted. The patient felt comfortable with a 25-mm (1 inch) block

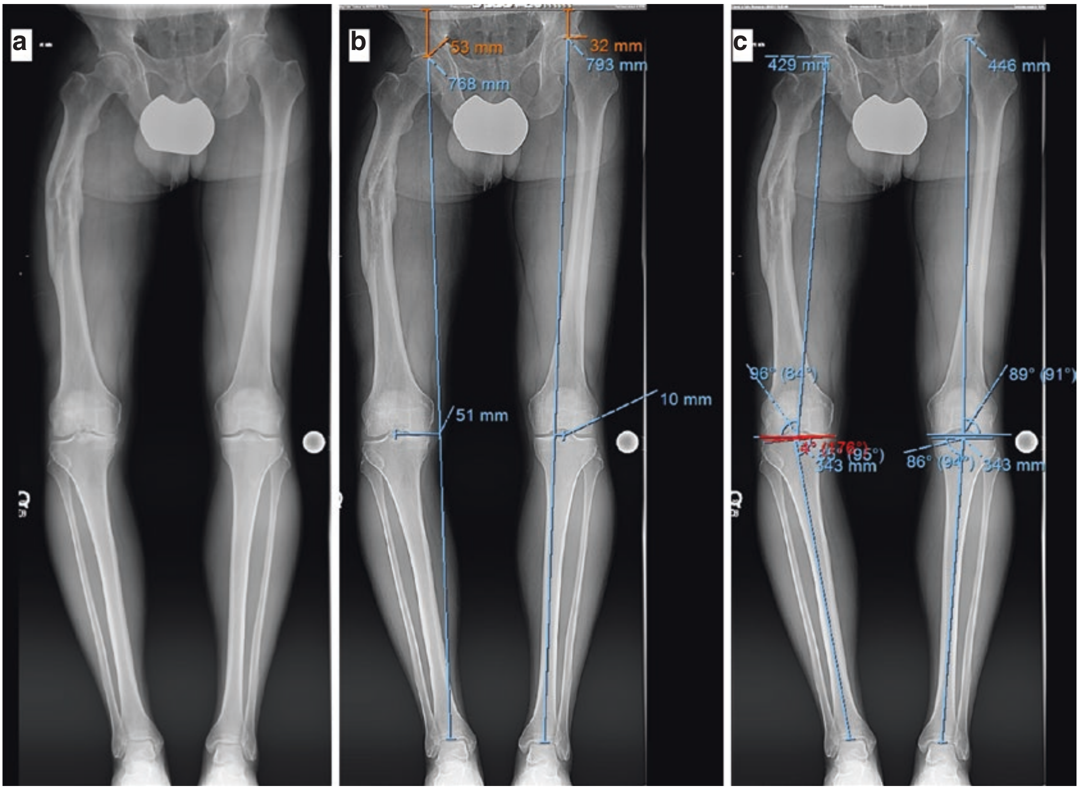


Fig. 21.5 (**a**) Preoperative standing hip-to-ankle radiograph with a calibration marker. (**b**) Limb length and mechanical axis assessments are performed. The distance from the center of the knee to the mechanical axis line (mechanical axis deviation, MAD) quantifies the amount of coronal malalignment. (**c**) Joint orientation angles [24] are measured to determine which bone(s) the varus malalignment originates from. In this case, the mLDFA (mechanical lateral distal femoral angle) and MPTA (medial proximal tibial angle) were both abnormal, measuring 96° and 85°, respectively

Case 2

A 42-year-old female presented with a valgus deformity and shortening of her left lower limb from a previous growth arrest (Figs. 21.11, 21.12, 21.13, 21.14, 21.15, and 21.16). Deformity analysis indicated the valgus alignment of the limb originated in the femur as the mLDFA was 82 degrees. Concurrent limb lengthening and coronal realignment were therefore planned. Planning of the distal femur osteotomy demonstrated that a trajectory of 82 degrees relative to the joint orientation line of the distal femur would yield the appropriate realignment. Blocking screws on the concavity of the deformity, distal and proximal to osteotomy, are necessary since the osteotomy is located in the metaphysis (non-isthmic location).

Why This Worked

It is imperative to understand which bone contributes to the coronal deformity. Since it was located in the femur, and an internal lengthening nail was planned (for limb equalization), the surgeon uses “anatomic axis planning” and blocking screws to ensure the correct entry point and trajectory in the distal femur.

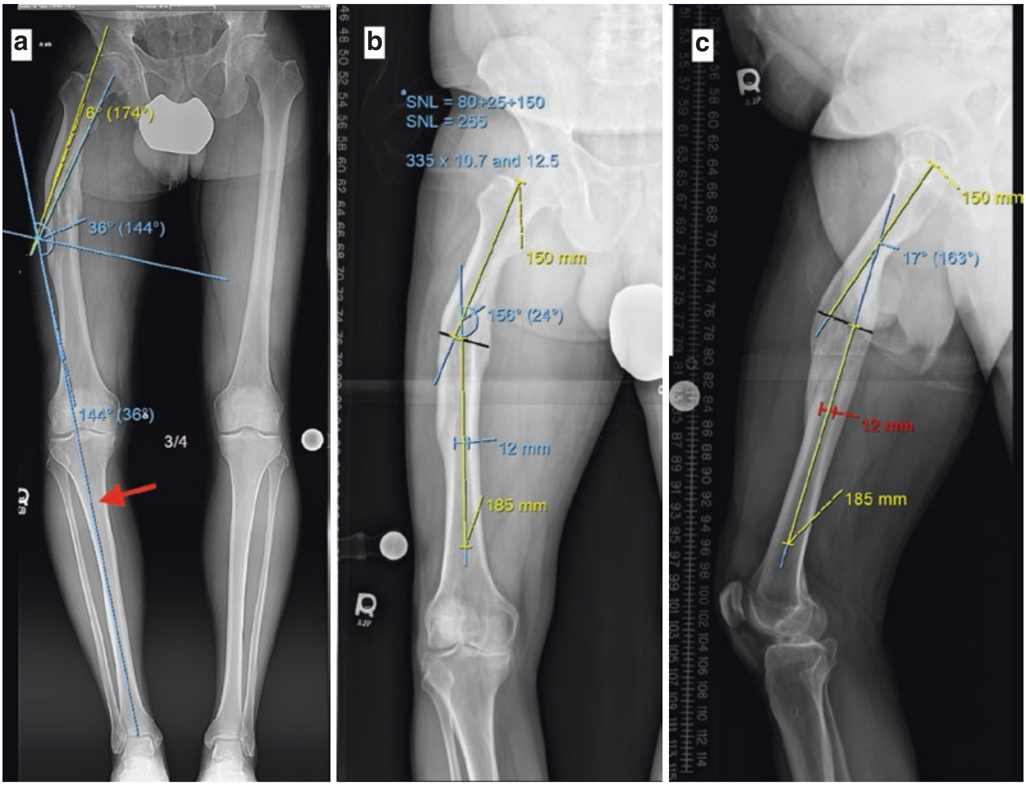


Fig. 21.6 Preoperative planning. (**a**) Mechanical axis planning. The *red arrow* indicates the desired final mechanical axis. The surgeon chose to have this pass through the lateral tibial spine since there was preexisting medial compartment knee arthritis. The proximal mechanical axis is generated using an anatomical-mechanical angle (AMA) of 6°. This generates an apex of the deformity and a magnitude of 36°. (**b**) Anteroposterior radiograph drawing the anatomic axes of the proximal and distal segments highlights the apex of the deformity and the magnitude (24°). Osteotomy measures 150 mm from the tip of the greater trochanter, which is then applied to the lateral radiograph. (**c**) The SNL, or shortest length nail calculation, is used to determine the shortest intramedullary nail that would be able to stabilize the femur at the end of lengthening. As the nail elongates, the smaller diameter portion at the tip of the nail (which protrudes 30 mm) extends. This segment does not afford canal stability. We recommend 50 mm of the thicker portion of the rod to be maintained past the distraction zone. Therefore, in this case, based upon 25 mm of lengthening, the SNL = 25 + 150 + 30 + 50 = 255 mm. The surgeon chose a 335-mm rod based on the inventory from NuVasive and the anatomy of the patient's femur

Fig. 21.7 Intraoperative clinical photograph of the patient undergoing osteotomy. A biplanar external fixator was used to maintain the alignment during canal preparation in order to ensure non-eccentric reaming. This is important since the IM rod is straight, and the realignment of the proximal and distal anatomic axes dictates the accurate coronal and sagittal deformity correction

Case 3

A 22-year-old male sustained a right open Gustilo and Anderson type IIIB tibia fracture from a motor-vehicle accident. He had a previous bone transport with a healed docking site. He presented with a residual limb length discrepancy of 30 mm and valgus malalignment. The strategy to correct his limb malalignment was to insert a PRECICE® nail to obtain equalized limb length and coronal angular correction with a laterally placed blocking screw (Figs. 21.17, 21.18, 21.19, 21.20, 21.21, and 21.22).

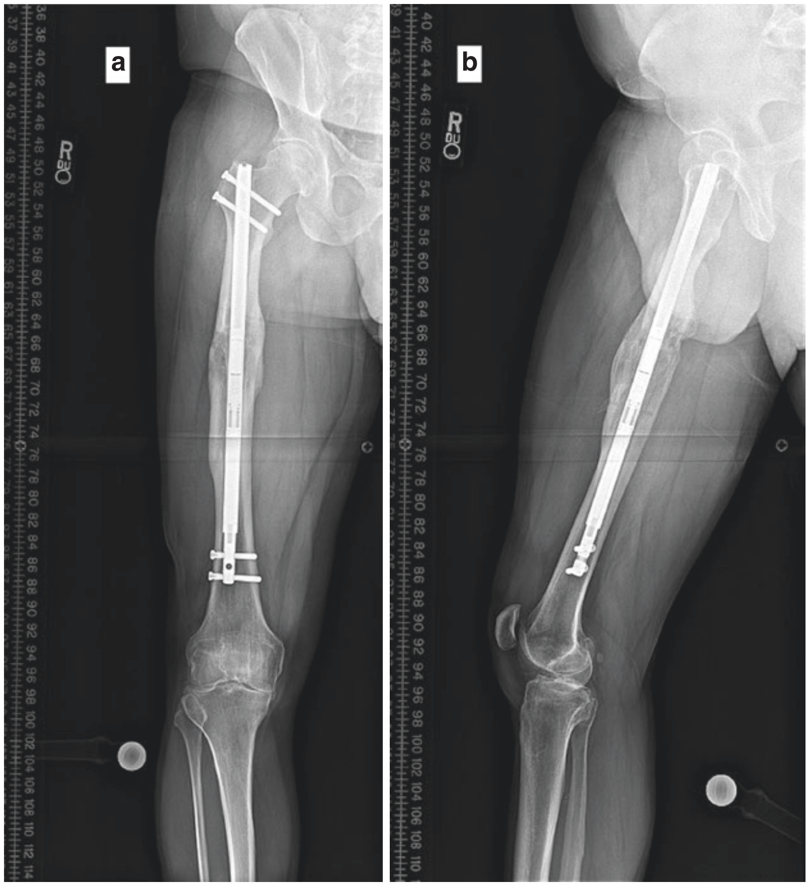


Fig. 21.8 (**a**) Postoperative anteroposterior and (**b**) lateral radiographs demonstrate healed regenerate on all four cortices

Why This Worked

This worked well since the deformity in the coronal plane was located in the diaphysis of the tibia. Therefore, realignment with anatomic axes (using an intramedullary rod) would subsequently realign the limb. Since limb equalization was also sought, an internal lengthening nail with blocking screws in the concavity of the deformity would achieve limb equalization and coronal realignment.

Complications

Failure to Distract

Despite the fact that every PRECICE® implant is tested prior to shipping out of the factory, failure of a freshly implanted nail to distract has been documented [17, 18] and can be prevented by testing the distraction mechanism at the end of the surgical procedure. The external remote controller (ERC) is wrapped in a sterile bag and positioned over the limb to distract the nail 0.5 to 1.0 mm. The distraction space can be seen on the fluoroscopy screen. At the surgeon's discretion, the distraction can be reversed (compressed) back to neutral length. A "dead" nail will be detected immediately and can be replaced during the same surgery. Recently, a magnet has been made available that can be attached to a drill so intraoperative distraction or compression can be done outside of the patient to ensure that the nail is functioning and/or adjusted to the appropriate length needed for compression or distraction.

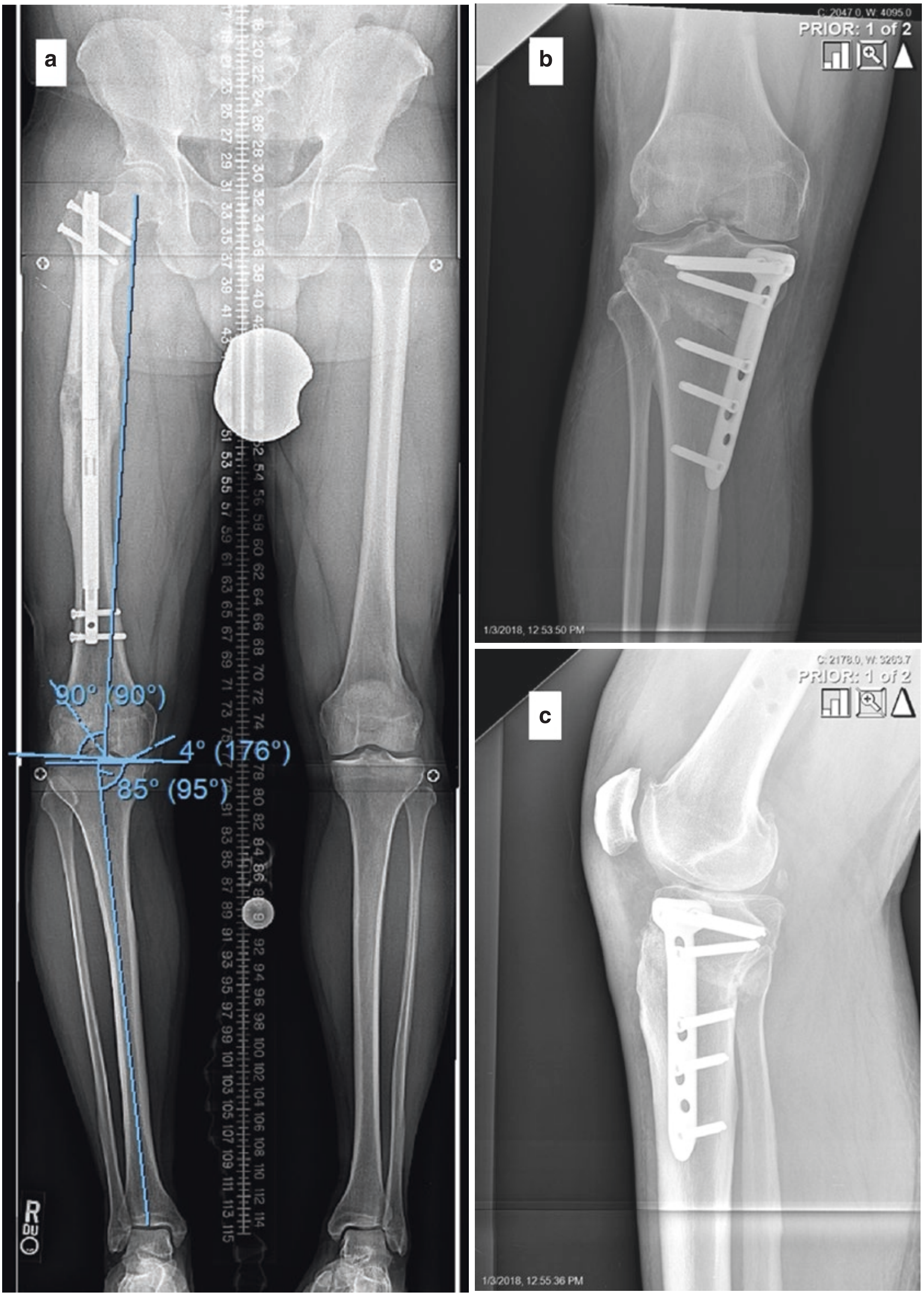


Fig. 21.9 Standing hip-to-ankle radiograph after limb lengthening, (**a**) Patient still has residual varus originating from the tibia, which was known since femoral malunion correction was predicted to correct only part of the entire coronal limb deformity. (**b**, **c**) Anteroposterior and lateral radiographs after proximal tibial opening wedge osteotomy to bring the limb alignment through the lateral tibial spine

Fig. 21.10 Final clinical photograph of the patient standing from the front (**a**) and the back. (**b**) Patient had his limbs equalized and mechanical axis corrected

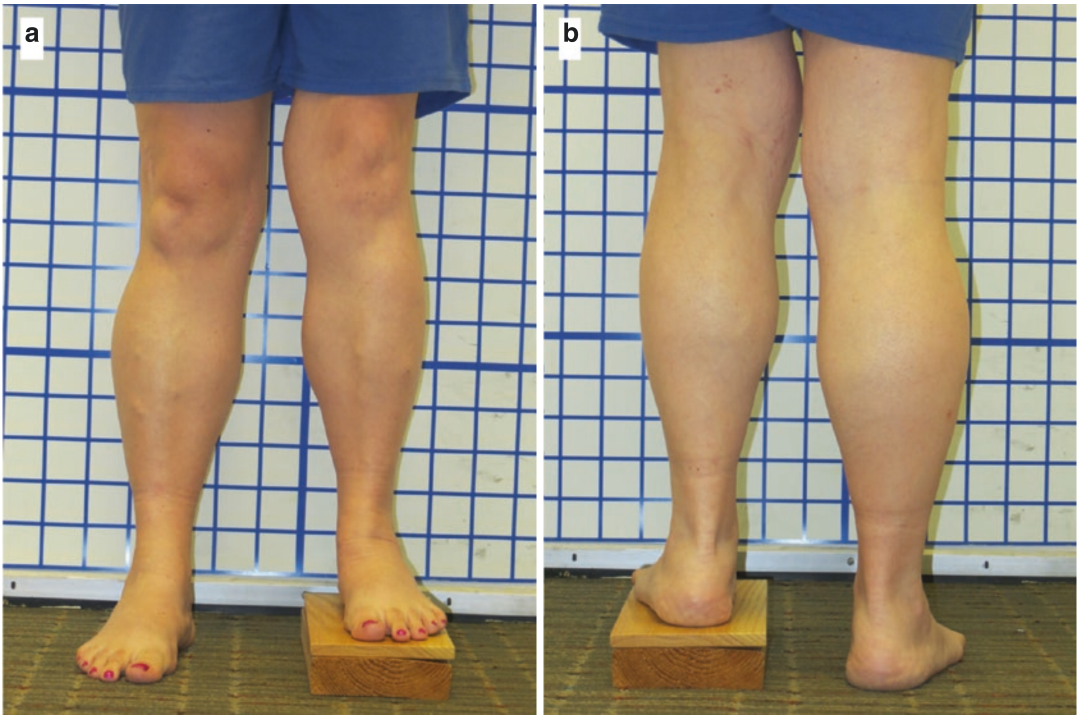


Fig. 21.11 Standing clinical preoperative assessment of the patient from the front (**a**) and the back. (**b**) Patient should be evaluated using blocks with different heights to equalize the pelvis. Patient felt most comfortable with a 1.25-inch block under her left foot

Crown Failure

The P2 design addressed the concern for rotational stability issue of the nail by improving the telescopic section's resistance to torsion at the crown of the nail. Despite this effort, crown failures occur [8, 20] and produce rotational instability. Patients will recognize the problem immediately. They may report hearing a popping sound during a rotational maneuver followed by a sensation of loss of control of the knee position. This phenomenon has been seen in P2 and P2.1 models and in all diameters: 8.5, 10.7, and 12.5 mm. It occurs almost exclusively in cases of bilateral femoral lengthening and is thought to be related to excessive weight-bearing (Fig. 21.23). The nail may still be able to distract despite a crown failure or it may retract (shorten) [14].

Premature Consolidation

Although premature consolidation is not typically a mechanical issue, certain etiologies for this complication are mechanical. In most of the retrospective series of PRECICE® lengthenings, the causes of this complication were not included. If the ERC is not communicating with the nail's magnet, the distraction will slow or stop and lead to premature consolidation. This can be due to a large thigh where too much tissue lies between the magnets. It can also occur during longer lengthenings when the nail's magnet moves relative to the initial skin marking. This "migration" of the magnet can be tracked by placing a metal BB over the skin marking during radiographs. The skin marking can then be adjusted at each visit to ensure it lies over the internal magnet

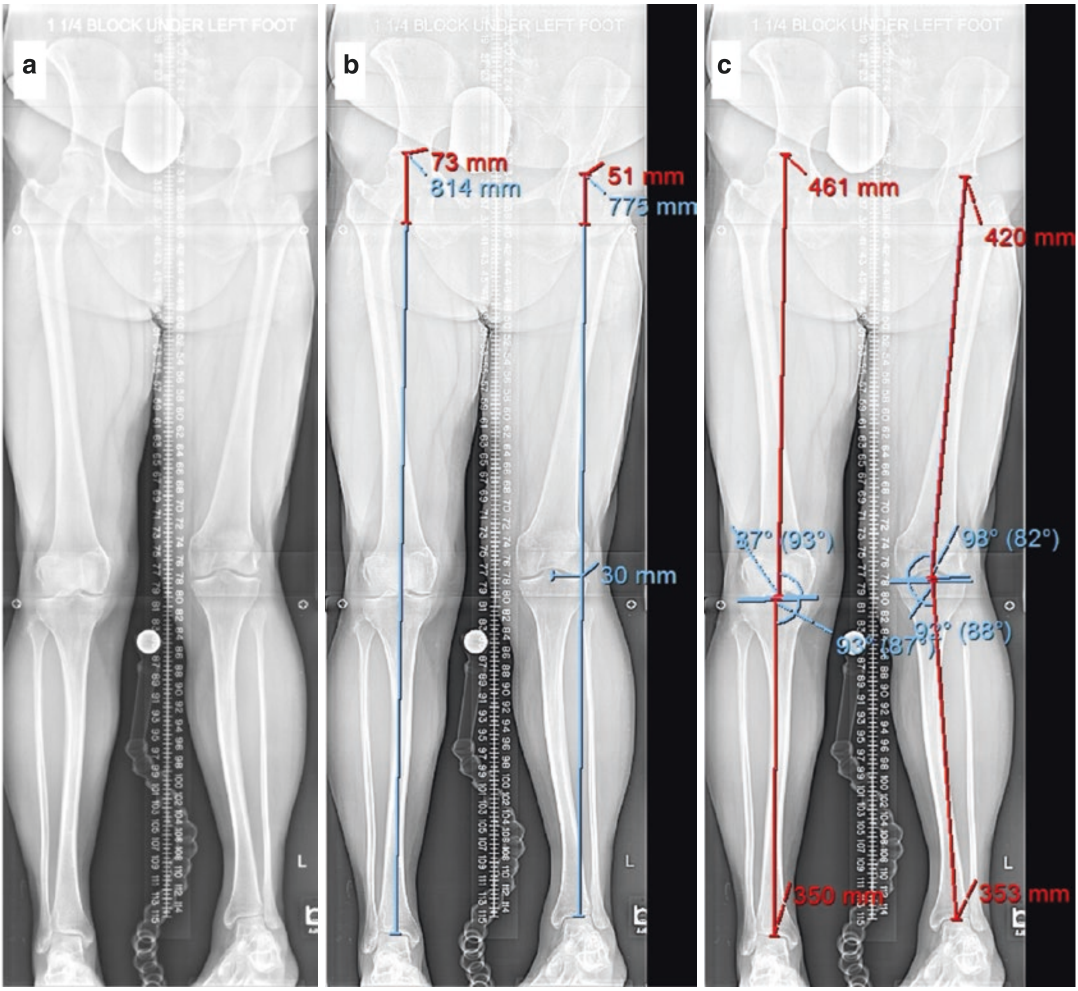


Fig. 21.12 (**a**) Standing hip-to-ankle radiograph with the same 1.25-inch block under the left foot to level the pelvis. (**b**) Limb length calculations are performed and mechanical axis deviation is assessed. Note that the left side is 39 mm short (814–775), and the patient has valgus alignment (MAD, 30 mm lateral to the center of the knee). (**c**) Joint orientation angles (JOA) are calculated to determine the origin of the valgus. In this example, the distal femur is in valgus with a mechanical lateral distal femoral angle (mLDFA) of 82°. In addition, the femur and tibial limb segments are measured to determine where the LLD is coming. In this case, the femur is short by 41 mm (461–420)

(Fig. 21.24). A dead nail (fails to distract) can be the cause of a premature consolidation and requires a fresh nail available in the OR during the revision osteotomy.

Regenerate Insufficiency

Insufficiency of the regenerate bone is a biological failure that can lead to mechanical failure of the implant. The P1 nail's welding seam was particularly weak [12–14, 20, 27]. While a redesign seems to have solved outright fracture in the P2 generation, crown failure continues to be a problem. A delayed union needs to be addressed to avoid the chance of mechanical breakdown. Typically, this is done with bone grafting and nail retention or exchange nailing with reaming and insertion of a larger diameter trauma intramedullary nail (Fig. 21.25).

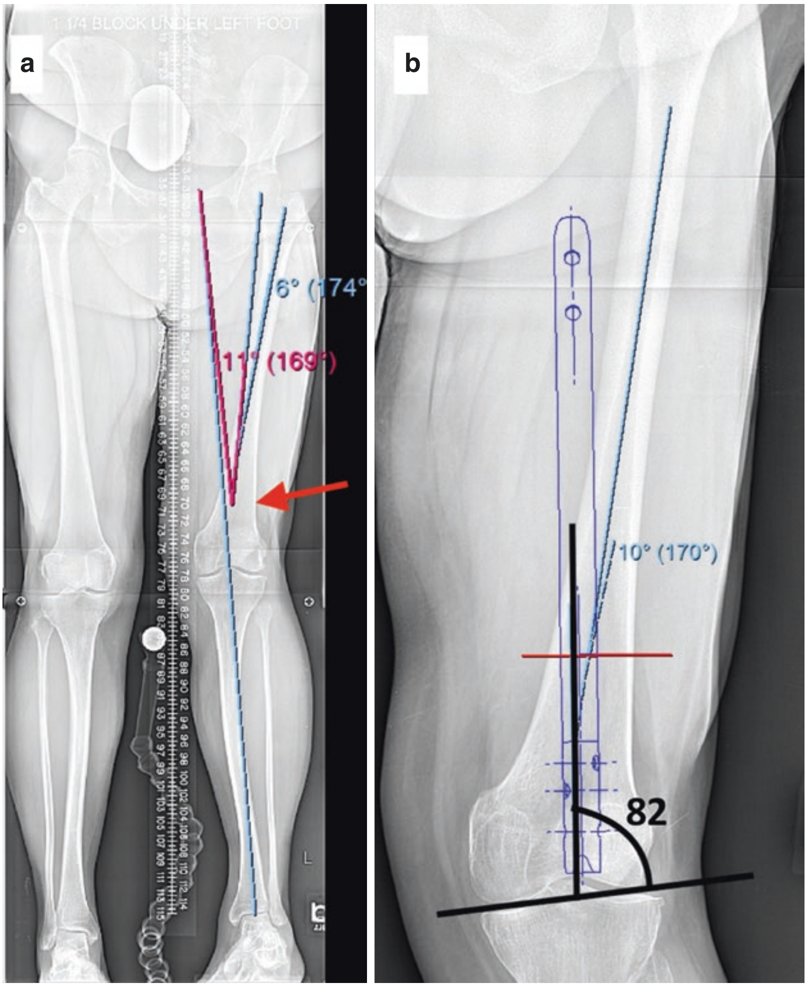


Fig. 21.13 (**a**) Mechanical axis planning to determine the magnitude of the deformity, which measured 11°. The *red arrow* indicates the level of the osteotomy, and an anatomic-mechanical angle (AMA) of 6° is used. (**b**) Preoperative anatomic axis planning, for a 10° coronal angulation correction. The red line represents the level of the osteotomy. Since the osteotomy is located in the metaphyseal portion of the femur, blocking screws will be necessary to obtain and maintain the alignment. The intersection of the black lines (joint orientation line and anatomic axis line) represents the desired entry point and trajectory in the distal segment of the femur in order to generate a 10° correction as the IM nail engages the diaphyseal segment of the femur. In this case, the desired anatomic lateral distal femoral angle (aLDFA) is 82°

MRI Incompatibility

The current manufacturer of the PRECICE® nail, NuVasive Specialized Orthopedics™, recommends removal of the implant prior to obtaining an MRI due to incompatibility. Theories of potential maladies that would arise from MRI scanning include heating of the nail and extremity, and uncontrolled distraction or compression. Gomez et al. [25] studied the effect of 3T MRI on

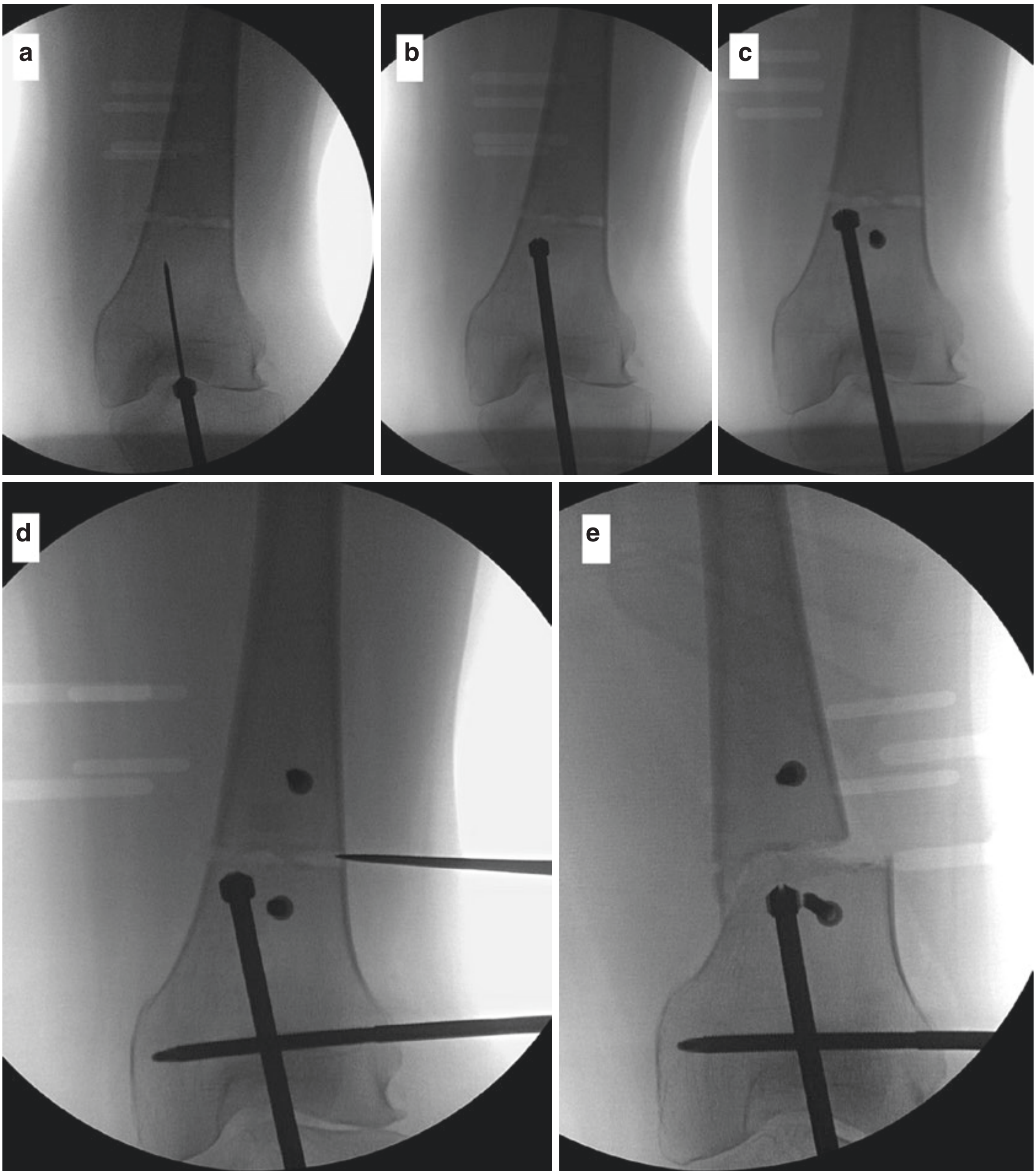


Fig. 21.14 These fluoroscopic images demonstrate the intraoperative execution of the preoperative planning. (**a**) A starting guide wire is inserted with the same entry point and trajectory as preoperatively planned. (**b**) A rigid reamer is used to ensure that the path in the distal segment is maintained. Small adjustments to the trajectory can be performed. With the reamer left in place, blocking screws are placed on the concavity of the deformity in both the proximal (**c**) and distal (**d**) segments. The osteotomy is completed, and reaming is performed in the proximal segment (**e**)

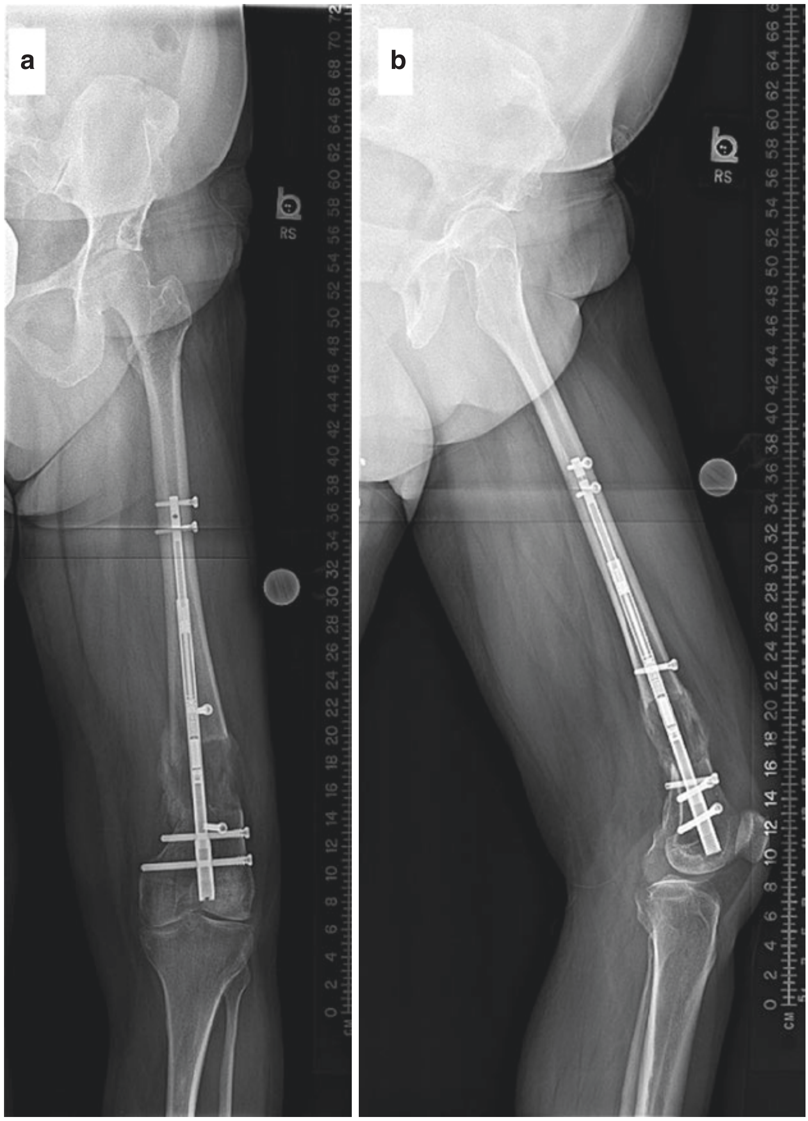


Fig. 21.15 (**a**) Postoperative anteroposterior and (**b**) lateral images after lengthening was performed. Note the blocking screws increase the stability of the construct and maintain the coronal alignment during lengthening

PRECICE® nails and found there was no involuntary nail distraction and minimal implant temperature increased from 3.3 to 3.6 degrees Celsius. However, there was a profound effect on the nail's ability to generate distraction force. The 3T scanning damaged the nail's internal mechanism reducing the distraction force of the femur nails by 62% and the tibia nails by 90% [26]. In summary, MRI will "kill" the nail but not injure the patient.

Corrosion and Late Failure

PRECICE® nails are typically removed at the completion of consolidation; therefore, most surgeons have not considered the long-term effects of this implant. The MAGEC nail (NuVasive), which relies on a similar design, remains in vivo for several years and has taught us much. Concerning reports document high systemic serum titanium and vanadium levels

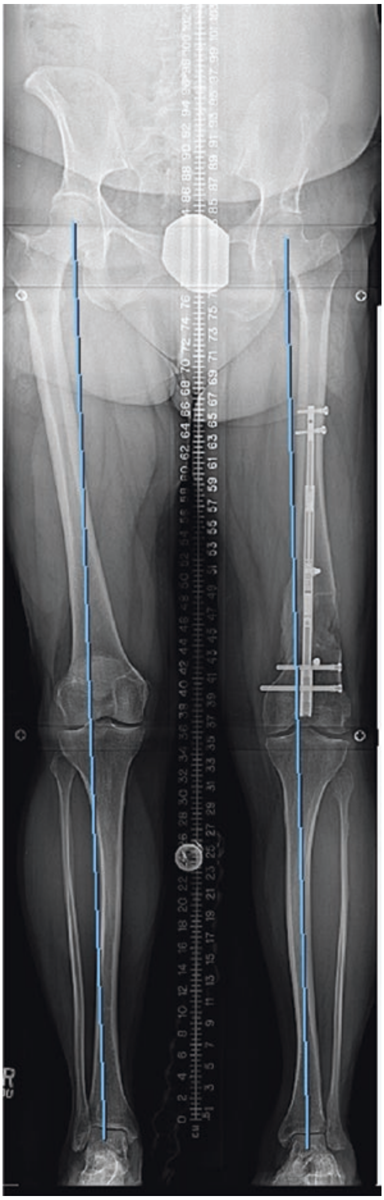


Fig. 21.16 Final standing hip-to-ankle radiograph of the patient. Note mechanical axis and limb lengths have been corrected

and corrosion of the internal mechanism through the entrance of titanium wear particles and biologic material into cracks in the previously sealed gear box [26]. The mixing of titanium and stainless-steel molecules led to corrosion and failure of the actuator pin in many MAGEC rods [27]. Sectioning of previously used P2 nails showed no signs of corrosion but did reveal the entrance of biological debris into the sealed area creating an environment for corrosion over time and making an argument for continuing the practice of early removal of the nails [28]. Foong et al. further studied material wear of the PRECICE® nails by analyzing eleven nails retrieved from patients [29]. The investigators found that the P2.1 model sustained significantly less wear than its predecessors, due to an improved internal anti-rotation mechanism. The nail diameter and amount of lengthening were not related to the severity of wear. The authors concluded that PRECICE® wear was minimal compared to the MAGEC system due to the absence of actuator pin fractures.

Compression Nail and Staged Lengthening

The PRECICE® intramedullary nail can also be used in a compression mode to heal difficult fractures [30] or to deliver sustained compression to a long bone nonunion [16, 20]. This technique requires pre-distracting the telescopic portion of the nail 10–13 mm creating room for compression. Once a nonunion has consolidated, an osteotomy can be performed around the dormant (“sleeper”) nail and the nail lengthened. Personal experience has demonstrated a poorer quality regenerate than we have become accustomed to seeing with PRECICE® femoral lengthenings highlighting the importance of osteotomy technique. Osteotomy around an existing nail with the lengthening over a nail (LON) method was described for correcting post-traumatic limb shortening. This technique involves carefully performing the corticotomy around an existing intramedullary nail. Authors found a sluggish BHI of 52 days/cm reinforcing the need to distract slowly [31] (Fig. 21.26).

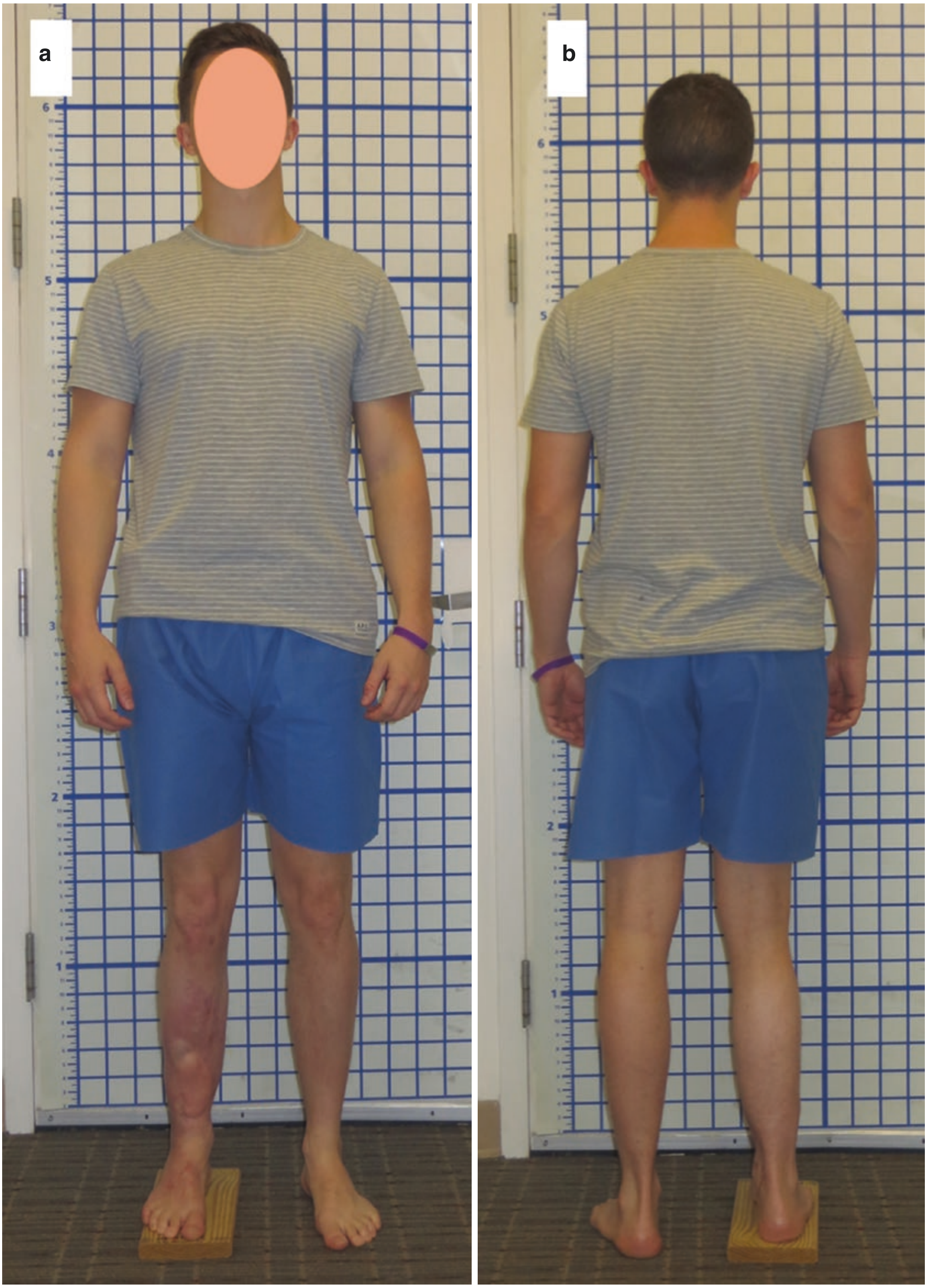


Fig. 21.17 Preoperative clinical photos of the patient from the front (**a**) and back. (**b**) Patient felt comfortable with a 30-mm block under his right foot

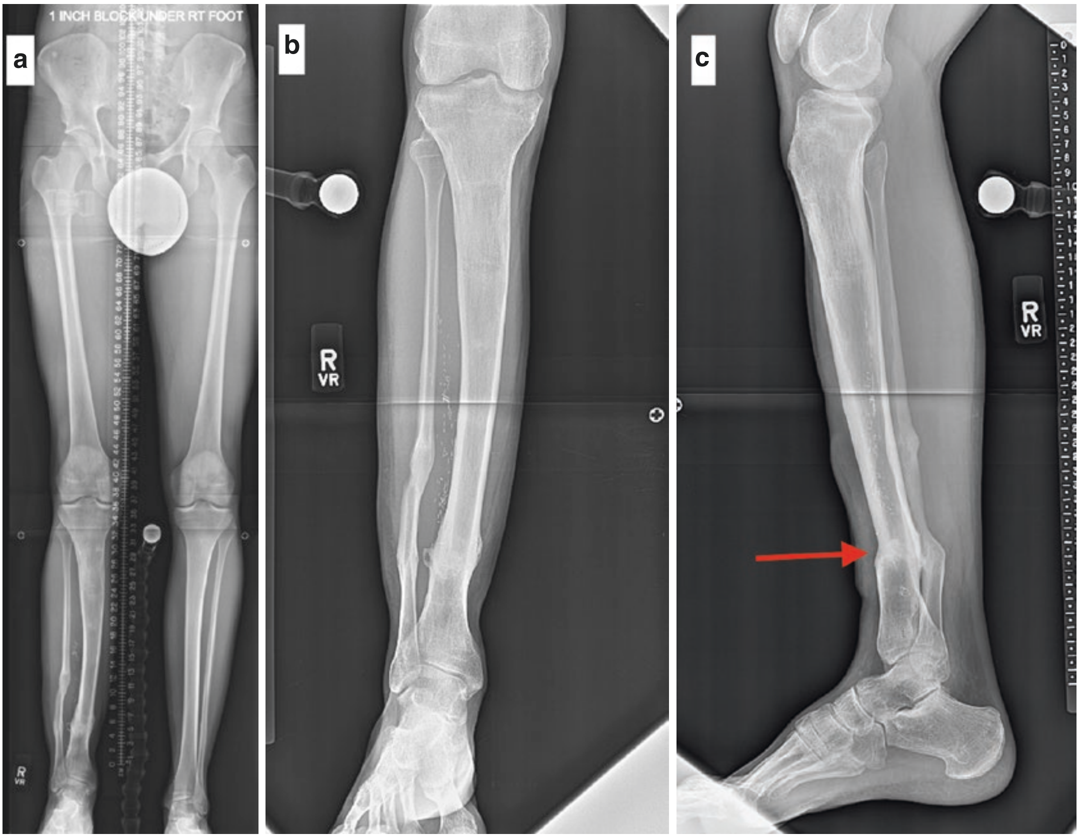


Fig. 21.18 As part of the preoperative radiographic examination, a standing hip-to-ankle radiograph is ordered with a block under the short limb to equalize the pelvis (**a**). Dedicated views of the affected bone are ordered (**b**, **c**). It is important to instruct the radiograph technicians to obtain the entire bone on one cassette. This helps appreciate the deformity, in this case, valgus. In addition, the docking site was translated (*red arrow*). Thus, planning for the IM nail would need to stop short of this distal segment

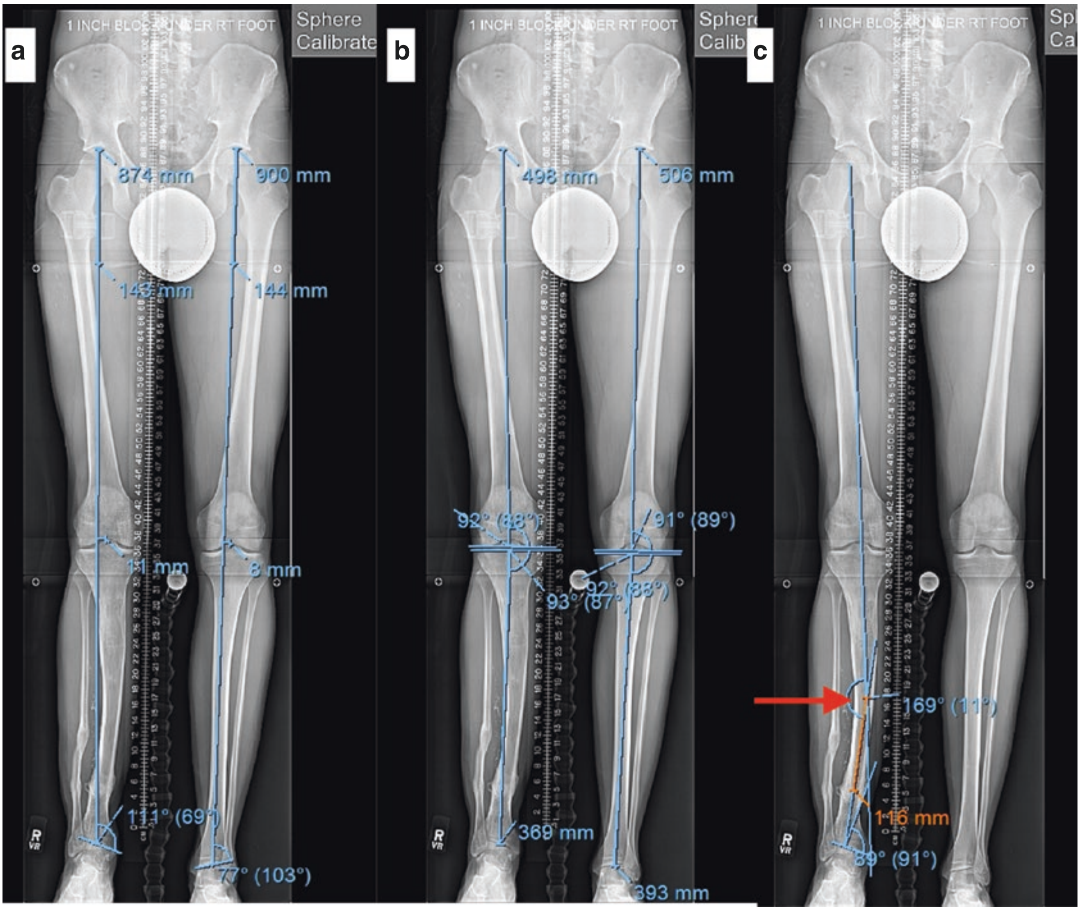


Fig. 21.19 Coronal preoperative planning performed on the standing hip-to-ankle radiograph. (**a**) Limb lengths and mechanical axis deviation are assessed. The right limb is in valgus (MAD 11 mm lateral) and short by 26 mm (900–874). (**b**) Joint orientation angles and limb segment lengths are measured and indicated that the right tibia is short and in valgus. (**c**) Mechanical axis planning aiming for a final MAD that passes through the center of the knee. Apex of the deformity (*red arrow*) and magnitude (11°) are noted

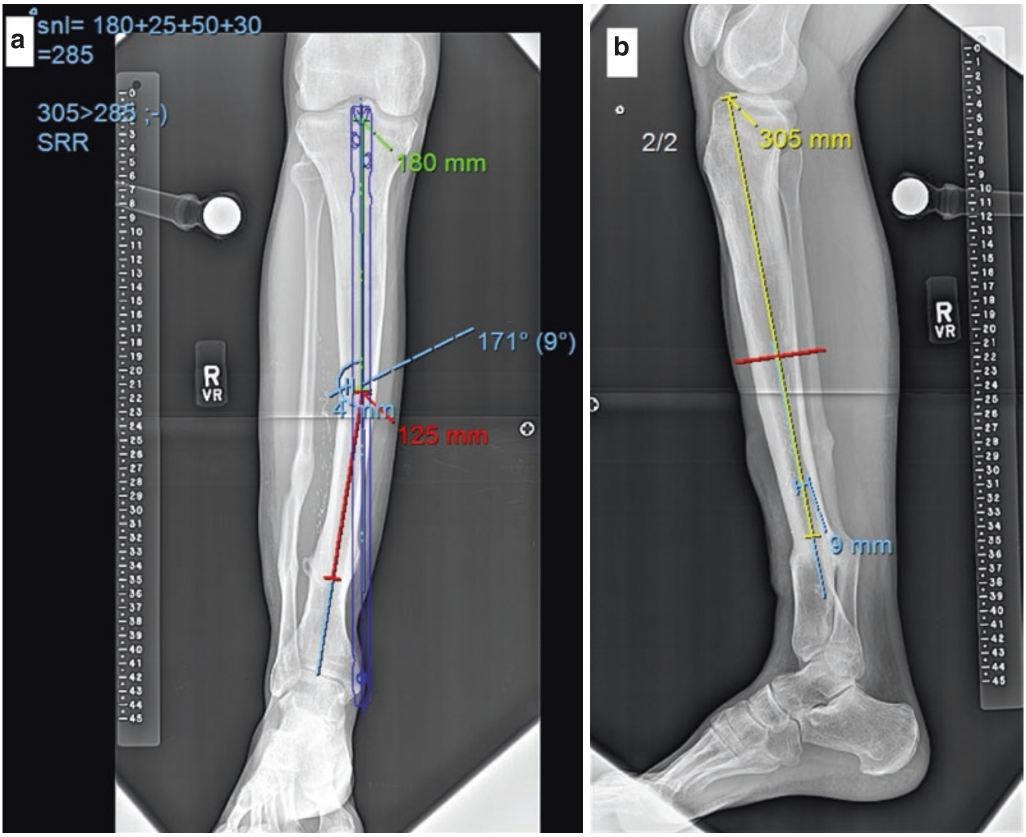


Fig. 21.20 (**a**) Anteroposterior whole view of the tibia for preoperative planning. Note the calibration ball for accurate measurements. Anatomic axis planning in the proximal and distal segments obtains the correction of 9°. The shortest nail length (SNL) calculation is performed for a 25-mm lengthening. Osteotomy is calculated at 180 mm from the joint line. The shortest tibial IM rod is referenced with the NuVasive inventory and is estimated to be 305-mm nail. The lateral radiograph is then referenced (**b**) to ensure that the IM rod stops just short of the docking site malunion

Fig. 21.21 Sequential anteroposterior (AP) radiographs of the right tibia after insertion of a tibial IM rod. Note that the blocking screw in the lateral tibial diaphysis obtains the initial valgus to varus correction and maintains this correction throughout the distraction and consolidation phases

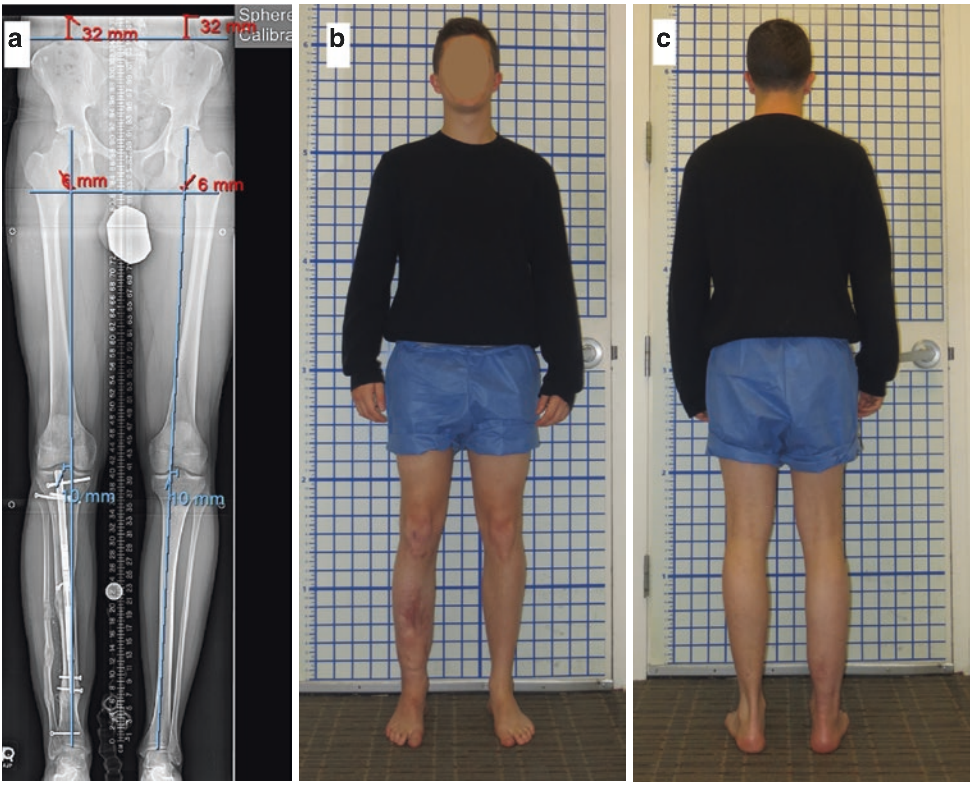


Fig. 21.22 (**a**) Final standing hip-to-ankle radiograph. Limb lengths and MAD have been equalized and matched to the contralateral limb. (**b**, **c**) Final standing clinical photographs of the patient

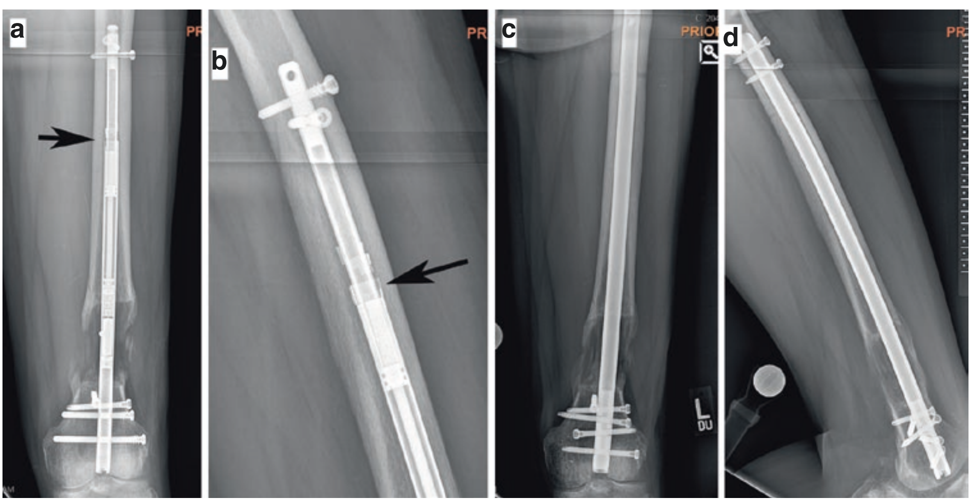


Fig. 21.23 (**a**) This patient underwent a 6-cm lengthening of the left femur with correction of varus using the P2.1. The crown of the nail fractured when he crossed his legs which was accompanied by a sensation of rotational instability. Radiographs showed fragmentation of the crown (*black arrow*) antirotational mechanism in the anteroposterior. (**b**) A lateral projection of the same. This nail was no longer able to lengthen and was exchanged for a trauma nail without loss of length. (**c**, **d**) Rapid healing ensued, as seen on the anteroposterior and lateral radiograph

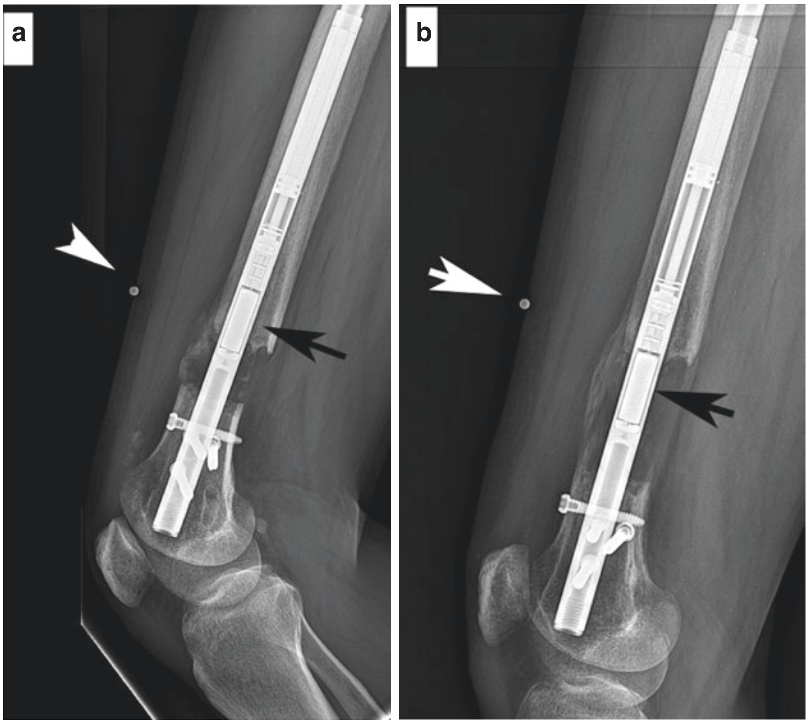


Fig. 21.24 (**a**) The external remote controller (ERC) needs to be positioned over the internal magnet during distraction; however, simply marking the internal magnet position on the skin during surgery does not guarantee proper location after lengthening has begun. For this radiograph, the patient has placed a metal BB over the area where the ERC is being used. The BB (*white arrow*) is directly over the internal magnet (*black arrow*). (**b**) After 6 cm of lengthening, the BB (*white arrow*) is positioned in the same location, but now it is no longer aligned with the internal magnet (*black arrow*) which may reduce the effectiveness of the ERC

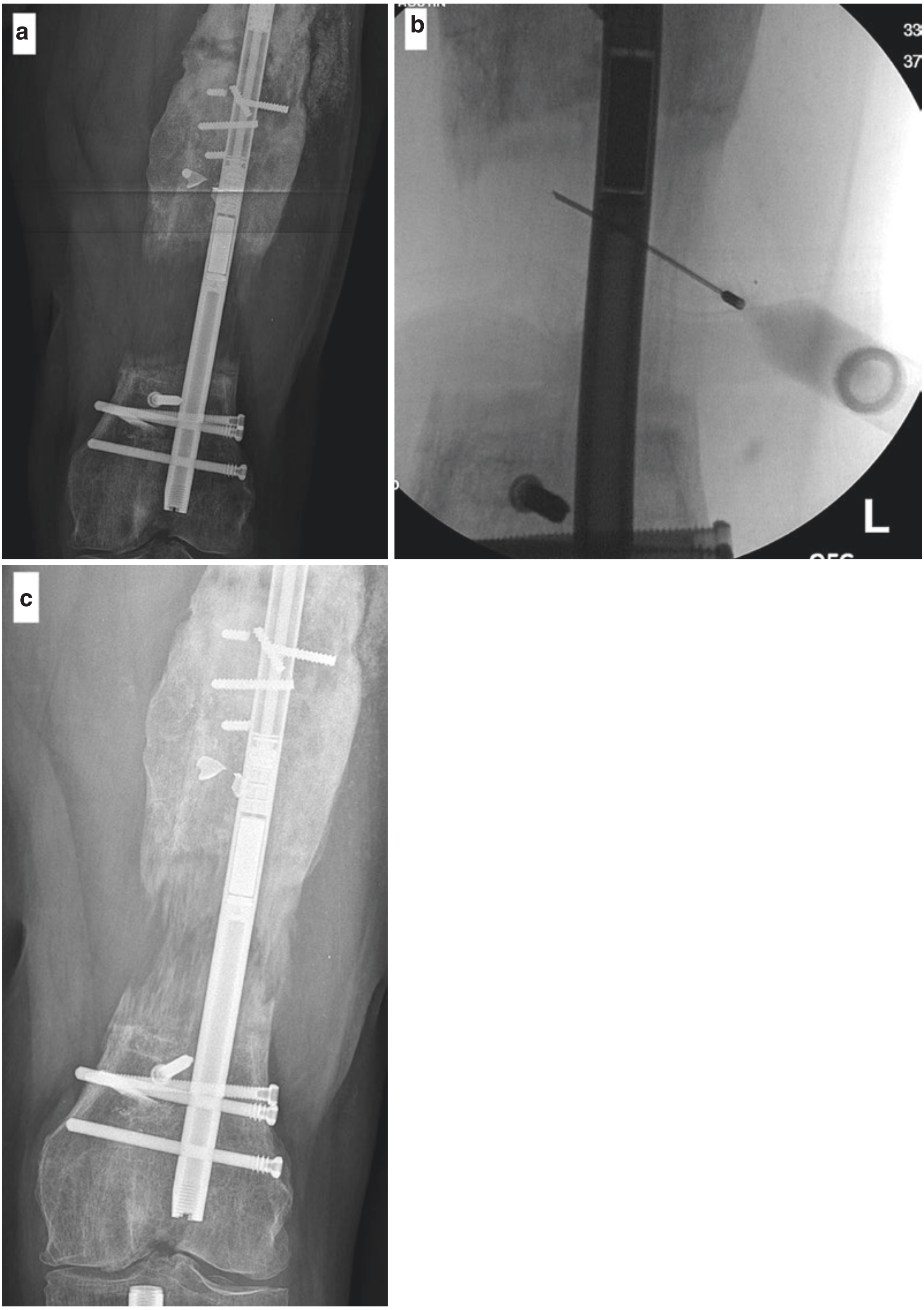


Fig. 21.25 (**a**) This patient was unable to produce a robust regenerate despite a very slow distraction rate and frequent pauses during lengthening. The patient was treated with stimulation of the regenerate by drilling with a K-wire and injection of bone marrow aspirate concentrate (BMAC) obtained from the iliac crest. (**b**) The needle can be seen injecting the BMAC. (**c**) This minor surgical intervention resulted in a vigorous regenerate within a few months' time

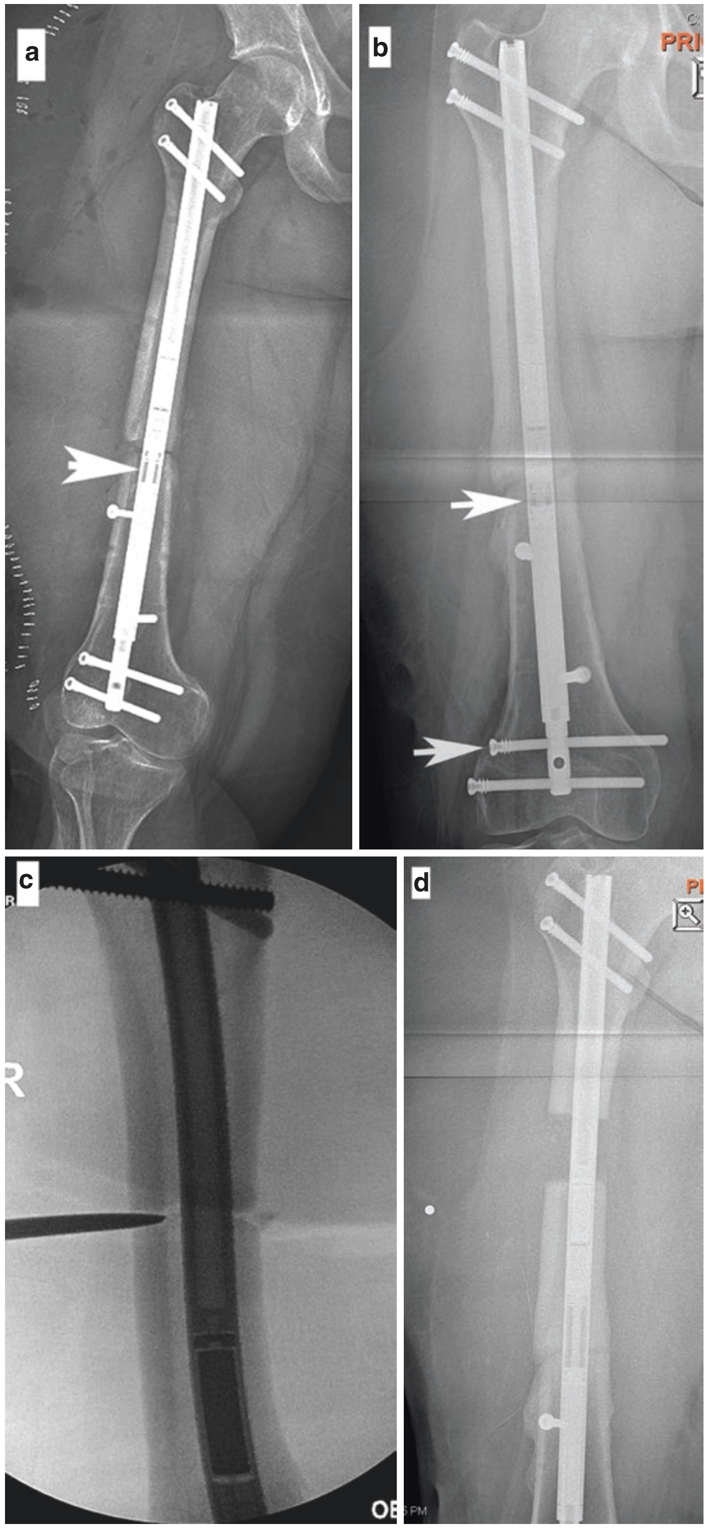
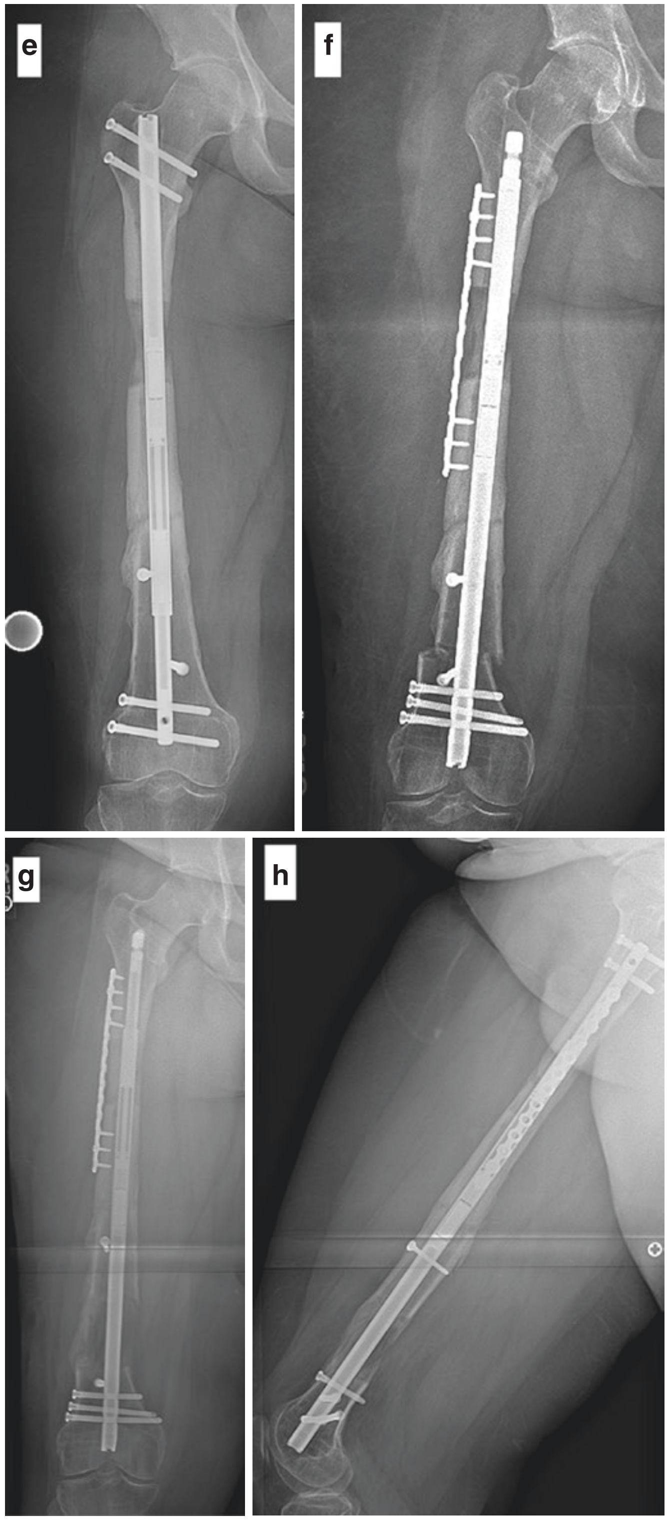


Fig. 21.26 This 28-year-old male suffered from a nonunion of the femur with 10 cm of limb shortening. (**a**) Before insertion, the nail was predistracted 15 mm (*arrow*) to allow for gradual compression at the nonunion site. (**b**) The nail was compressed 10 mm with notable shortening (*white arrow*) and bending of the distal locking bolt (*grey arrow*) indicating strong bone contact. (**c**) Once bony union was complete, an osteotomy for lengthening (osteoplasty) was created around the "sleeper nail" for distraction osteogenesis. (**d**) Lengthening proceeded at 1 mm/day with excellent functioning of the previously compressed nail but with poor regenerate formation. (**e**) The lengthening rate was slowed, and a total of 5 cm of length was achieved at the osteoplasty site with delayed healing. (**f**) A distal femoral osteoplasty with reaming and nail exchange with a retrograde nail was performed to lengthen the missing 5 cm and to correct some congenital distal femoral valgus. The previous lengthening site needed to be secured with a unicortical plate to prevent further separation at this compromised area. (**g**) The distal osteoplasty site produced a typical robust regenerate with distraction at 1 mm/day reestablishing normal limb length. (**h**) This lateral radiograph shows early healing at both osteoplasty sites

Fig. 21.26 (continued)



Conclusion

The Ilizarov method using internal lengthening nails is a reliable method to achieve limb equalization. Limb length discrepancy is one component of alignment, and the surgeon must be familiar with preoperative assessment (clinical and radiographic) of deformity in order to propose an appropriate strategy. In addition, the treating surgeon must be familiar with the multitude of complications associated with limb lengthening, some of which are unique to the PRECICE® intramedullary rod. However, the ability to use medullary nails for lengthening has improved the patient's experience and the surgeon's ability to perform these surgeries by decreasing complexity.

References

1. Ilizarov GA. The tension-stress effect on the genesis and growth of tissues. Part I. The influence of stability of fixation and soft-tissue preservation. Clin Orthop Relat Res. 1989;238:249–81.
2. Kenawey M, Krettek C, Liodakis E, Wiebking U, Hankemeier S. Leg lengthening using intramedullary skeletal kinetic distractor: results of 57 consecutive applications. Injury. 2011;42(2):150–5.
3. Mahboubian S, Seah M, Fragomen AT, Rozbruch SR. Femoral lengthening with lengthening over a nail has fewer complications than intramedullary skeletal kinetic distraction. Clin Orthop Relat Res. 2012;470(4):1221–31. 4.
4. Fragomen AT, Rozbruch SR. Retrograde magnetic internal lengthening nail for acute femoral deformity correction and limb lengthening. Expert Rev Med Devices. 2017;14(10):811–20.
5. Fragomen AT, Kurtz AM, Barclay JR, Nguyen J, Rozbruch SR. A comparison of femoral lengthening methods favors the magnetic internal lengthening nail when compared with lengthening over a nail. HSS J. 2018;14(2):166–76.
6. Iobst CA, Rozbruch SR, Nelson S, Fragomen A. Simultaneous acute femoral deformity correction and gradual limb lengthening using a retrograde femoral nail: technique and clinical results. J Am Acad Orthop Surg. 2018;26(7):241–50.
7. Hammouda AI, Standard SC, Robert Rozbruch S, Herzenberg JE. Humeral lengthening with the PRECICE magnetic lengthening nail. HSS J. 2017;13(3):217–23.
8. Furmetz J, Kold S, Schuster N, Wolf F, Thaller PH. Lengthening of the humerus with intramedullary lengthening nails-preliminary report. Strategies Trauma Limb Reconstr. 2017;12(2):99–106.
9. Hammouda AI, Jauregui JJ, Gesheff MG, Standard SC, Conway JD, Herzenberg JE. Treatment of post-traumatic femoral discrepancy with PRECICE magnetic-powered intramedullary lengthening nails. J Orthop Trauma. 2017;31(7):369–74.
10. Hammouda AI, Jauregui JJ, Gesheff MG, Standard SC, Herzenberg JE. Trochanteric entry for femoral lengthening nails in children: is it safe? J Pediatr Orthop. 2017;37(4):258–64.
11. Wagner P, Burghardt RD, Green SA, Specht SC, Standard SC, Herzenberg JE. PRECICE® magnetically-driven, telescopic, intramedullary lengthening nail: pre-clinical testing and first 30 patients. SICOT J. 2017;3:19.
12. Wiebking U, Liodakis E, Kenawey M, Krettek C. Limb lengthening using the PRECICE(TM) nail system: complications and results. Arch Trauma Res. 2016;5(4):e36273.
13. Karakoyun O, Sokucu S, Erol MF, Kucukkaya M, Kabukcuoglu YS. Use of a magnetic bone nail for lengthening of the femur and tibia. J Orthop Surg (Hong Kong). 2016;24(3):374–8.
14. Tiefenboeck TM, Zak L, Bukaty A, Wozasek GE. Pitfalls in automatic limb lengthening - First results with an intramedullary lengthening device. Orthop Traumatol Surg Res. 2016;102(7):851–5.
15. Laubscher M, Mitchell C, Timms A, Goodier D, Calder P. Outcomes following femoral lengthening: an initial comparison of the Precice intramedullary lengthening nail and the LRS external fixator monorail system. Bone Joint J. 2016;98-B(10):1382–8.
16. Fragomen AT, Wellman D, Rozbruch SR. The PRECICE magnetic IM compression nail for long bone nonunions: a preliminary report. Arch Orthop Trauma Surg. 2019;139(11):1551–60.
17. Schiedel FM, Vogt B, Tretow HL, Schuhknecht B, Gosheger G, Horter MJ, et al. How precise is the PRECICE compared to the ISKD in intramedullary limb lengthening? Reliability and safety in 26 procedures. Acta Orthop. 2014;85(3):293–8.
18. Kirane YM, Fragomen AT, Rozbruch SR. Precision of the PRECICE internal bone lengthening nail. Clin Orthop Relat Res. 2014;472(12):3869–78.
19. Paley D. PRECICE intramedullary limb lengthening system. Expert Rev Med Devices. 2015;12(3):231–49.
20. Fragomen AT. Transitioning to an intramedullary lengthening and compression nail. J Orthop Trauma. 2017;31(Suppl 2):S7–S13.
21. Landge V, Shabtai L, Gesheff M, Specht SC, Herzenberg JE. Patient satisfaction after limb lengthening with internal and external devices. J Surg Orthop Adv. 2015;24(3):174–9.
22. Fragomen AT, Fragomen FR. Distal femoral flexion deformity from growth disturbance treated with a two-level osteotomy and internal lengthening nail. Strategies Trauma Limb Reconstr. 2017;12(3):159–67.
23. Muthusamy S, Rozbruch SR, Fragomen AT. The use of blocking screws with internal lengthen-

ing nail and reverse rule of thumb for blocking screws in limb lengthening and deformity correction surgery. Strategies Trauma Limb Reconstr. 2016;11(3):199–205.

24. Paley D, Tetsworth K. Mechanical axis deviation of the lower limbs. Preoperative planning of uniapical angular deformities of the tibia or femur. Clin Orthop Relat Res. 1992;(280):48–64.
25. Gomez C, Nelson S, Speirs J, Barnes S. Magnetic intramedullary lengthening nails and MRI compatibility. J Pediatr Orthop. 2018;38(10):e584–e7.
26. Yilgor C, Efendiyev A, Akbiyik F, Demirkiran G, Senkoylu A, Alanay A, et al. Metal ion release during growth-friendly instrumentation for early-onset scoliosis: a preliminary study. Spine Deform. 2018;6(1):48–53.
27. Panagiotopoulou VC, Tucker SK, Whittaker RK, Hothi HS, Henckel J, Leong JJH, et al. Analysing a mechanism of failure in retrieved magnetically controlled spinal rods. Eur Spine J. 2017;26(6):1699–710.
28. Panagiotopoulou VC, Davda K, Hothi HS, Henckel J, Cerquiglini A, Goodier WD, et al. A retrieval analysis of the Precice intramedullary limb lengthening system. Bone Joint Res. 2018;7(7):476–84.
29. Foong B, Panagiotopoulou VC, Hothi HS, Henckel J, Calder PR, Goodier DW, et al. Assessment of material loss of retrieved magnetically controlled implants for limb lengthening. Proc Inst Mech Eng H. 2018;232(11):1129–36.
30. Watson JT, Sanders RW. Controlled compression nailing for at risk humeral shaft fractures. J Orthop Trauma. 2017;31(Suppl 6):S25–S8.
31. Kim HJ, Fragomen AT, Reinhardt K, Hutson JJ Jr, Rozbruch SR. Lengthening of the femur over an existing intramedullary nail. J Orthop Trauma. 2011;25(11):681–4.

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