Chapter

Biomechanical Basis of Bone Fracture and Fracture Osteosynthesis in Small Animals

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Abstract

The mastery of concepts related to biomechanics in bone fracture is crucial for the surgical success of osteosynthesis. The understanding of the basics of bone fracture is a skill fundamental to the choice of the correct method of osteosynthesis. Deep knowledge of implants, namely, their mechanic characteristics, and the correct technical use following the recommended guidelines for each type are crucial factors to decrease surgical failure and complications rate. This chapter reviews the biomechanical parameters of fracture repair that influence construct stiffness and strength. The authors also provided practical examples of the biomechanics concepts applied in clinical practice during this chapter. Metal alloys used in orthopedic implants are also fundamentally reviewed in their physical properties during this chapter. Fracture patterns vary hugely among patients and contributed to the difficult understanding of forces acting in fracture lines. However, fracture biomechanics basic knowledge and how osteosynthesis methods counteract the forces acting on fractures are key to surgical success.

Keywords: biomechanics, bone tissue, forces, fracture, dog, cat, osteosynthesis

1. Introduction

Biomechanics is a sub-branch of mechanics that studies the concepts of mechanics applied to the musculoskeletal system and the biomaterials used for treating orthopedic diseases. The structure, function, and motion of musculoskeletal tissues and their changes in orthopedic diseases are the main research topics of this science. The basic knowledge of the physical, chemical, and mechanical properties of biomaterials used for producing implants and prostheses is key to the orthopedic surgeon's understanding of why certain materials are used instead of others. The functional (mechanical) performance of implants and prostheses is strictly related to their material composition and design, being therefore the basic knowledge that the orthopedic surgeon should master and potentially influence his surgical planning or clinical decision.

Biomechanics encompasses the traditional branches of mechanics: kinematics, statics, and dynamics.

Kinematics is the study of motion without considering the forces that cause it and includes concepts such as trajectory, velocity, and acceleration. Motion can be a combination of translations and rotations, with translations involving the same displacement vector for all points in the body, while rotations involve different displacement vectors for different points.

Statics characterizes the forces acting on an object at rest or moving at constant velocity with zero acceleration. These forces can be direct forces or moments, which are equal and opposite forces acting on a body separated by a distance. The application of forces and moments to a body changes its state of rest. Equilibrium is a key principle in statics, and a body is in equilibrium when the sum of all applied loads is zero. In joints, applied forces include external loads such as body weight and internal loads such as muscle forces generated to maintain the joint in equilibrium. The equilibrium principle is used to analyze joint loading in a static context, where the joint of interest is studied in isolation from the rest of the body, and all forces and moments acting on it are identified. The resulting joint reaction force is then determined using the equilibrium condition.

Dynamics, a branch of mechanics, is concerned with the effects of forces on an object and the changes they produce in the object's motion. It encompasses the principles of both statics and kinematics by examining the actions of forces and the resulting motion and acceleration of the object. In orthopedic biomechanics, dynamic analysis is frequently utilized for activities such as gait studies. This involves determining the acceleration of body parts at any given time and the forces necessary to create these accelerations. The resulting forces are then determined using static analysis methods to obtain the resulting forces over the desired range of motion.

The interaction of biomaterials with tissues and cells is the ability of a biomaterial to perform its function without eliciting toxic or injurious effects on biological systems and is called biocompatibility, and it influences the mechanical performance of implants/prostheses in the short and long term. Nowadays, the biocompatibility concept includes bioinertia, biofunctionality, and biostability (acute and chronic toxicity of materials to tissues). Biointegration or colonization of implants by neighboring tissues is also framed in the concept of biocompatibility and is an important factor in the long-term biomechanical performance of implants/prostheses that should not be overlooked in clinical decisions.

Synthetic materials mainly metals and their alloys used for implants/prostheses are classified according to their biocompatibility as well as by their mechanical properties, such as tensile, compressive, and shear strength; hardness; stiffness; fatigue resistance to cyclic or acute loading; and creep behavior. The creep concept is a type of metal deformation that occurs at stresses below the yield strength (at elevated temperatures); it defines the stress at which metal begins to plastically deform. Factors such as ease of manufacture, cost, and production quality dictate the potential for the application of a biomaterial in orthopedics.

Load-deformation and stiffness, stress-strain, and elasticity are interconnected concepts to the understanding of the mechanical performance of implants and bone tissue that will be addressed in this chapter.

A thorough understanding of the unique biomechanical properties, characteristics, and behaviors of bone tissue, their alterations in disease, and the implants used in companion animal orthopedic surgery is essential for achieving successful results when attempting to manipulate bone healing. There is a consensus in the field of orthopedic surgery for companion animals that mastering these principles is associated with a low rate of postoperative complications.

A basic understanding of biomechanical principles and biomaterials knowledge is a fundamental skill for the companion animal orthopedic surgeon and forms an important component in the education of surgical trainees.

In this chapter, the information provided is divided into five main areas: biomechanical basic concepts, fracture biomechanics, biomechanics of bone tissue, applied fracture biomechanics to common clinical presentations in small animal osteosynthesis, and biomechanics of implant biomaterials, covering what the authors considered in-depth knowledge of biomechanical principles of bone fracture and applied biomechanics to fracture osteosynthesis in small animals.

The main objective of this chapter is to provide information about biomechanics applied to fracture management in small animals that will help the veterinary surgeon to take more evidence-based decisions with the ultimate goal of surgical success.

2. Biomechanical basic concepts

2.1 Strain

Strain is a local deformation parameter expressed as units of length per length, usually expressed as a percentage, and is therefore dimensionless when the bone is loaded with different force vectors.

The mathematical definition (Eq. (1)) of strain is the change in length divided by the original length.

Formula for strain calculus:

$$\text{Strain } (\%) = \frac{\text{Change in length (mm)}}{\text{Original length (mm)}} \tag{1}$$

Due to the dimensionless characteristic of this parameter, strain provides a clinically useful scaled measure of the displacement of bone fragments and can compare strain values in bones of different lengths. For example, a 1 mm fracture gap displacement is more significant in a 10 mm rat (strain 10%) femur than in a 300-mm dog femur (strain 0.3%) [1, 2].

2.2 Stress

One of the main functions of the appendicular skeleton is to support the body weight in rest or during movement, and consequently, bone is the tissue that supports more mechanical loads. When a force is applied to a bone, this will cause a stress situation [1, 2].

By definition, like pressure, stress is a local force expressed in units of force per unit area (Eq. (2)). The SI unit of force is the Newton, and force is often expressed as N/m^2 or Pascal (Pa) [1, 2].

Formula for stress calculus:

$$\text{Stress}\left(\frac{\text{N}}{\text{mm}^2}\text{orPa}\right)=\frac{\text{Load (N)}}{\text{Cross}-\text{sectional area }(\text{mm}^2)} \tag{2}$$

The damage that the load will cause depends on the area over which it is being distributed: a large force applied over a small area will result in greater stress, the contrary not being true. For instance, that is what happens when a skeletally immature dog falls from a height and supports all the weight on the hind limb. In these cases, the load will be equally distributed proximally and distally to the knee; however, due to the minor dimension of the distal part (tibial crest) of the tibial tuberosity, the fracture is normally located in this point due to stress concentration in a small area [3].

Bone tissue is constantly submitted to mechanical loads, comprehending forces/loads of compression (axial), torsion, tension, bending, and shearing. Usually, these forces act in combination; however, they can be more predominant in an isolated way in certain locations of bones. Conceptually, it is considered that the deformation occurs in the bone tissue when small animals move and will vary between 0.04 and 0.3%, hardly exceeding 0.1%. This interval characterizes the elastic deformation of the bone, which is conceptually an initial response to the establishment of a load in the bone. In this scenario, the deformation/length of the bone returns to the initial dimensions/shape once the load is removed. It is a natural and important process for the homeostasis of bone tissue. An interesting characteristic of bone tissue is related to bone deformation according to the mechanical load that is applied. There are materials conceptually defined as isotropic, which respond to mechanical load regardless of the orientation of the material. By contrast, a material is considered anisotropic if the response to mechanical load varies with orientation. Bone is an example of an anisotropic material with mechanical properties that depend on the orientation of the bone lamellae. Thus, the mechanical properties are not equal in all directions and depend on the direction of the load applied. Long bones are stronger in longitudinal orientation than in tangential or radial orientation, since osteons have a longitudinal orientation in cortical bone. The maximum load that the bone will support is directly related to the direction in which the force is being applied. An illustrative example of this statement is that the appendicular bone supports a greater axial (compressive) load if compared to the transverse load. This difference between maximum strength in different directions emphasizes the anisotropic characteristic of bone. An example of an isotropic material is the stainless-Steel 316L (metallic alloy commonly used in the production of plates, which presents a similar behavior regardless of the direction of the load that is applied, with a similar resistance [3].

From a mechanical point of view, bone is also considered a viscoelastic tissue. Viscoelasticity is the property of materials that exhibit both viscous and elastic characteristics when undergoing deformation. Practically, this represents the ability of the bone to resist deformation without loss of definitive structural integrity. Viscoelasticity of bone is dependent on several factors including water, mineral, and collagen type I content [4].

This concept justifies the resistance of bone to sudden loads/impacts like jumps or falls: if there is some degree of deformation, there is later a return to the original form.

However, bone tissue is not always capable of withstanding all the loads that are applied to it, and this failure occurs when the imposed force exceeds the elastic deformation capacity, which may cause a complete or incomplete fracture. In this last situation, permanent deformation occurs in the tissue even after the load is removed, triggering microfractures and trabecular disruption and preceding macroscopic rupture [3].

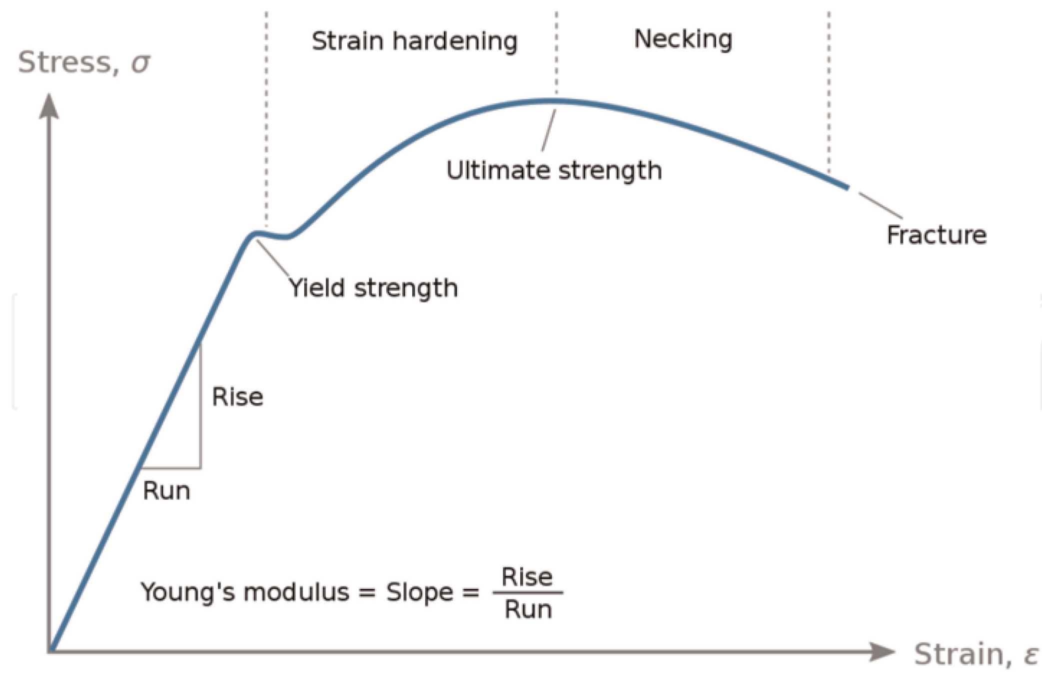

Figure 1.
Graphic representation of the stress/strain curve, and the biomechanical concepts of young modulus, yield, and fracture point.

When a supraphysiological force is applied, the bone will be deformed. The mechanical properties of bone are well represented by the load/stress-strain curve (**Figure 1**). The load/stress response of the bone directly depends on the length, thickness, density, shape, and type of bone (cortical vs. cancellous), so the stress-strain curve represents the structural properties of the object. Bone is characterized by a stress-strain curve with initially a linear response, the so-called elastic deformation. From the point at which the response ceases to be linear to the load and starts to express a curve (physiological limit point), plastic deformation appears and microfractures occur at the structural level, which can culminate in macroscopic fractures if the load is not interrupted or maximum force is reached (fracture point) [3].

2.3 Strength and stiffness

Strength denotes the ultimate load a material can withstand before a catastrophic failure, which is also designated as the fracture point (**Figure 1**) when this concept is applied to bone [1–3].

The stiffness of a biomaterial or bone tissue is the mechanical property that characterizes and quantifies the changes in the original shape when a force vector or a load is applied to it. A graphic that represents stiffness is called a load/displacement curve, and the relationship between stress and strain for materials, or load and displacement for structures, can help us understand these properties better. The slope of the straight part of the curve that ascends represents the elastic modulus or stiffness. The steeper the slope of this part of the curve, the stiffer the material. The strain (or change in shape) in this part of the curve is elastic, which means the material can return to its original shape after the force is removed. There is a point on the curve, called Y, known as the yield point or yield load where the curve stops being nonlinear (**Figure 1**). At this point, the strain exceeds the material's ability to recover from its original shape, and the material gets a permanent change in shape if the load is removed. This point shows where the material changes from elastic to plastic deformation. This point is important for clinical reasons because it means that the bone acquires a different shape from its original [2].

This permanent change in shape, called plastic (instead of elastic) deformation, occurs when covalent bonds break at a molecular level. The point on the curve where the material breaks or fails is called U, or ultimate failure/fracture point (**Figure 1**). The curve also shows how much energy the material can absorb during the loading process [2]. This is called toughness, and it is represented by the area under the curve [2].

2.4 Elastic modulus of young

The elastic modulus, also called the modulus of Young (Y), represents the relationship between stress and strain and is one of the most useful parameters for mechanically comparing biomaterials. It is calculated by applying the formula represented by Eq. (3), where the slope of the stress is plotted versus the strain curve (**Figure 1**). It has the same units as stress (N/mm^2) because strain does not have any units [3]. It quantifies the relationship between tensile/compressive stress. Young's moduli values are normally so large that this parameter must be expressed in gigapascals (GPa) instead of in Pascals. The components of the formula for Young's modulus calculation are: σ (force per unit area) and axial strain, ε (proportional deformation) in the linear elastic region of a material and is determined using the formula (Eq. (3)):

Young modulus formula:

$$E = \frac{\sigma}{\varepsilon} \tag{3}$$

The elastic modulus of Young is a measure of the linear elasticity of a material. This parameter allows grading materials in two categories: flexible and rigid. Materials with higher Y values are considered rigid. The bone tissue, for example, is included in the rigid category with a value of 15 GPa but is less rigid when compared to materials used in the manufacture of orthopedic implants. The elastic modulus of stainless-steel implants is usually 188 GPa, pure titanium is 116 GPa, and titanium alloy (Ti-6Al-4 V) is 113 GPa. A single value of Y assumes a linear relationship, which is true for metals (until their yield point) [3].

2.5 Area moment of inertia (AMI)

The area moment of inertia is a geometric parameter to be considered when the mechanics of implants are studied. The AMI is a measure of the resistance of materials exclusively related to flexion loads. This parameter is only influenced by the geometry and not by the composition of materials. Implants manufactured with bigger AMI have the least probability of structural collapse when submitted to higher flexion loads (higher stiffness to flexion loads). AMI does not take into account material properties, and for that reason, AMI must be only used to compare different constructs of the same material. AMI is a geometric parameter that is calculated based on the dimensions of the structure in the direction of bending. For a circular implant (e.g., a pin or interlocking nail), the direction is not relevant, and the formula used for this particular type of implant is the following (Eq. (4)):

AMI formula for circular implant:

$$\frac{1}{4.p.r4} \tag{4}$$

In Eq. (4), the radius is raised to the fourth power, so a small increase in the diameter of a pin or other circular implant has a large impact on its bending stiffness.

This concept can be illustrated by the following example: if the AMI of a 2.4 mm pin (3/32 inch) is 1.6 mm^4 and for a 3.2-mm (1/8 inch) pin is 5.1 mm^4, an increase of 33.3% of the pin diameter results in an increased AMI value, 3 times larger than an original pin.

For solid rectangular structures, AMI is calculated using a different formula (Eq. (5)):

AMI formula for solid rectangular structures:

$$\frac{1}{3.b.h3} \tag{5}$$

In the AMI formula, *b* is the width, and h is the height in the direction of bending. Plate thickness is an important parameter because this dimension is cubed. However, for bone plates, the presence of the screw holes adds complexity to this calculus. At the screw holes, the AMI is usually less than half the value that would be calculated from its external dimensions [5].

2.5.1 The impact of plate orientation on AMI

If the direction of the bending force of a fracture is known or the vector force can be simplified to a craniocaudal direction, the orthopedic surgeon can also use this concept of AMI to consider alternative plate locations. The classic example of this concept is for distal radius fractures. In this type of fractures, the primary direction of bending is considered to be in the craniocaudal plane; if a 2.7-mm LC-DCP (Limited Contact-Dynamic Compression Plate) is placed on the medial aspect, a higher AMI value (solid section of approximately 111 mm^4) will be measured when compared to a 3.5 mm LC-DCP placed on the cranial aspect (AMI of 30 mm^4), because the height of the 2.7 mm plate in the direction of bending is 8 mm (almost 3 times greater), compared with 3.3 mm for the 3.5 mm plate [2].

Another variant that influences the AMI is the position of the implant regarding the neutral axis of the bone, which is represented by the medullary canal of the bone. If the implant is positioned more distant from the neutral axis, the implant is less efficient to resist the bending forces. For the mentioned reason, the interlocking nail is the most mechanically favored implant to resist bending forces [3].

The use of AMI helps the decision-making process for choosing the osteosynthesis method but is not exclusively based on this parameter. Every long bone has a tension side and a compression side when axial loading is applied to the bone that will cause deformation, promoting bending. When the axial loading is applied and the bone bends, one side will experience tension and the cortices suffer traction. At the same time, the opposite side of the bone and the bone cortices will experience compression. Every long bone has a neutral axis that corresponds to the medullary cavity and that does not suffer compression or traction forces, and also has a tension and compression sides [1–3].

By using the AMI and tension/compression sides defined for each bone, in the decision-making process of orthopedic surgery, two premises must be fulfilled to succeed:

1. Positioning the plate on the tension side of the bone, because this side will be more mechanically challenged by the traction forces. When bending forces are applied to the bone, the tension side will suffer distraction and the plate protects the bone. On the opposite side, the bone suffers compression, which is less mechanically challenged;

2. Fractures not anatomically reconstructed in the compression side (gap fracture) due to a comminuted fracture or by losing far cortical support will suffer an inversion of the former dynamic. During the flexion loading of the bone, the plate side will suffer compression, which predisposes to plate failure. Clinically, in these cases, we must augment AMI by increasing the plate thickness, increasing the working length of the plate, combining implants (orthogonal and bilateral plating and/or intramedullary pinning), or using buttress plates without screw holes in the working length of the plate (biological osteosynthesis plate) [3].

2.6 Working length of plate

In locking plates, the distance between the proximal and distal screw in closest proximity to the fracture is defined as the "working length" of the plate (**Figure 2**). If a plate is compressed against the bone (dynamic compression plate), the working length is the distance between the bone ends of the fracture span by the plate (**Figure 2**). A correlation between plate working length and stiffness of the construct, plate strain, and cyclic fatigue properties of the plate has been shown [5, 7–10]. Another fundamental aspect in high-strain fracture management (e.g., simple transverse fractures) is load sharing between the stabilized bone and the plate; not addressing this aspect in osteosynthesis increases the risk of cyclic fatigue and early failure of the plate.

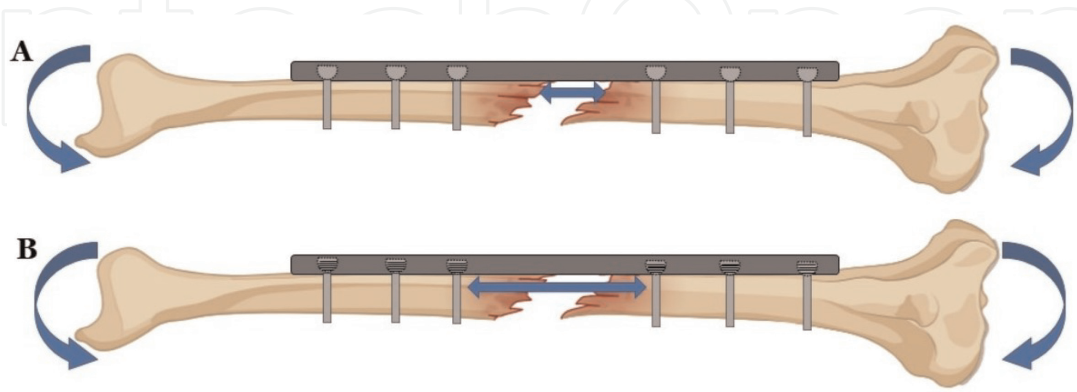

Figure 2.
Illustration of the concept of working length of the plate. The working length of a bridging plate is defined by the type of plate (LCP vs DCP) and the interaction of the plate with the bone. (A) A dynamic compression plate is held to the bone by non-locking screws. When bending occurs, the working length is the distance between the bone ends of the fracture span by the plate. (B) The locking plate spans the same gap, but because the plate is not in direct contact with the bone, the distance between the nearest proximal and distal screw to the fracture line is defined as the working length.

Also, the plate length influences the screw loading. For a given amount of bending moment, a longer plate produces markedly less pull-out force than a short plate due to an improvement of the working leverage for the screws.

The effective application of plate length is another concept. The farthest screws determine the effective usage of plate length and contribute to fracture gap stability. A long plate produces markedly less pull-out force than a short plate.

2.7 Improved anchorage by diverging screws

Locked or non-locked screws with divergent inclinations improve the anchorage considering that a bigger amount of bone is displaced when compared to parallel screws when an equal pull-out load is applied.

2.8 The helicopter effect

The tightening of the first screw (head locking) in one extreme of the plate without stabilization of the other end of the plate will cause the "helicopter effect". This effect also occurs if only two screws were applied in auxiliary plating used in the orthogonal plating technique. In orthogonal plating, the auxiliary plate must have a minimum of two screws applied for the segment to prevent the helicopter effect (**Figure 3**).

2.9 The strain theory

The control of interfragmentary micromotion is the key point for correct fracture stabilization. The knowledge of the factors that dynamically influence the distances between fracture fragments is fundamental to controlling micromotion. The strain theory is the most important concept used in the decision-making process, regarding fracture osteosynthesis from a mechanical perspective.

Essentially, strain or relative deformation is the amount of movement (distancing/approaching) between fracture fragments relative to the original distance (gap).

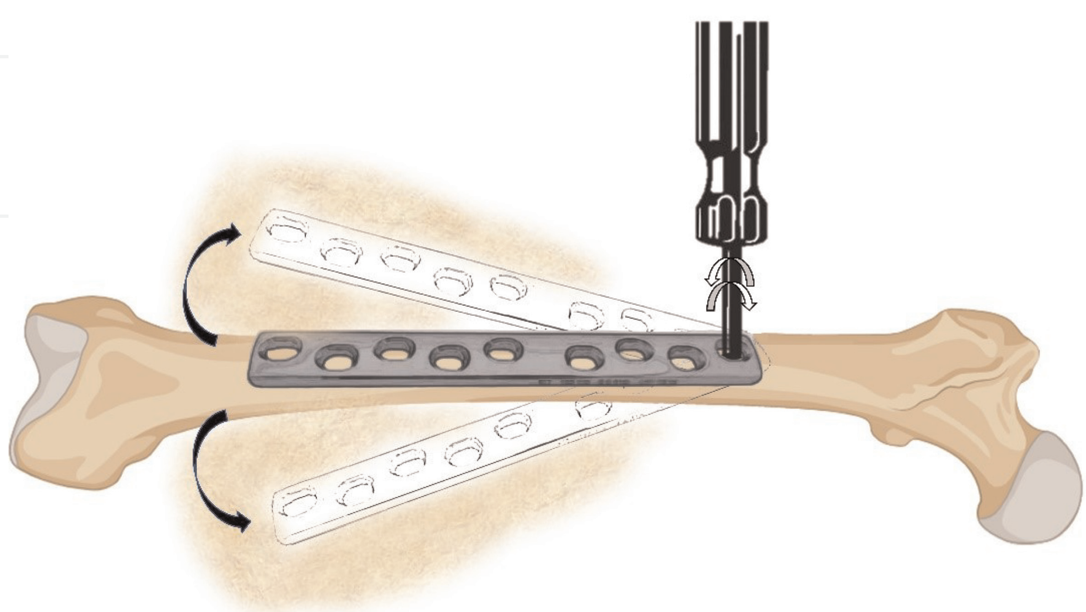

Figure 3.
Illustration of the helicopter effect due to the lack of stabilization of both ends of the plate.

It is expressed as a percentage of movement, that is, movement of gap/original fracture gap when the fragments were subject to mechanical stimulus (weight support, muscular contraction, and passive movement, among others). The calculus of interfragmentary movement in a laboratory environment is obviously more precise than in clinical settings. In clinical scenarios, several factors can influence the magnitude and direction of the fragments in the process of distancing/approaching movement; among these factors are the great variability of fracture patterns and the correspondent mix pattern of strain simultaneously present at the fracture lines [3].

Mathematically, the strain is determined by the formula (Eq. (4)):

Formula for strain determination:

$$E = \frac{\Delta L}{L}$$

where the E is the strain expressed by % value, ΔL is the variation of interfragmentary space (gap variation), and L is the initial gap. By the equation, is possible to infer that if the initial gap is bigger, the final strain value inversely will be

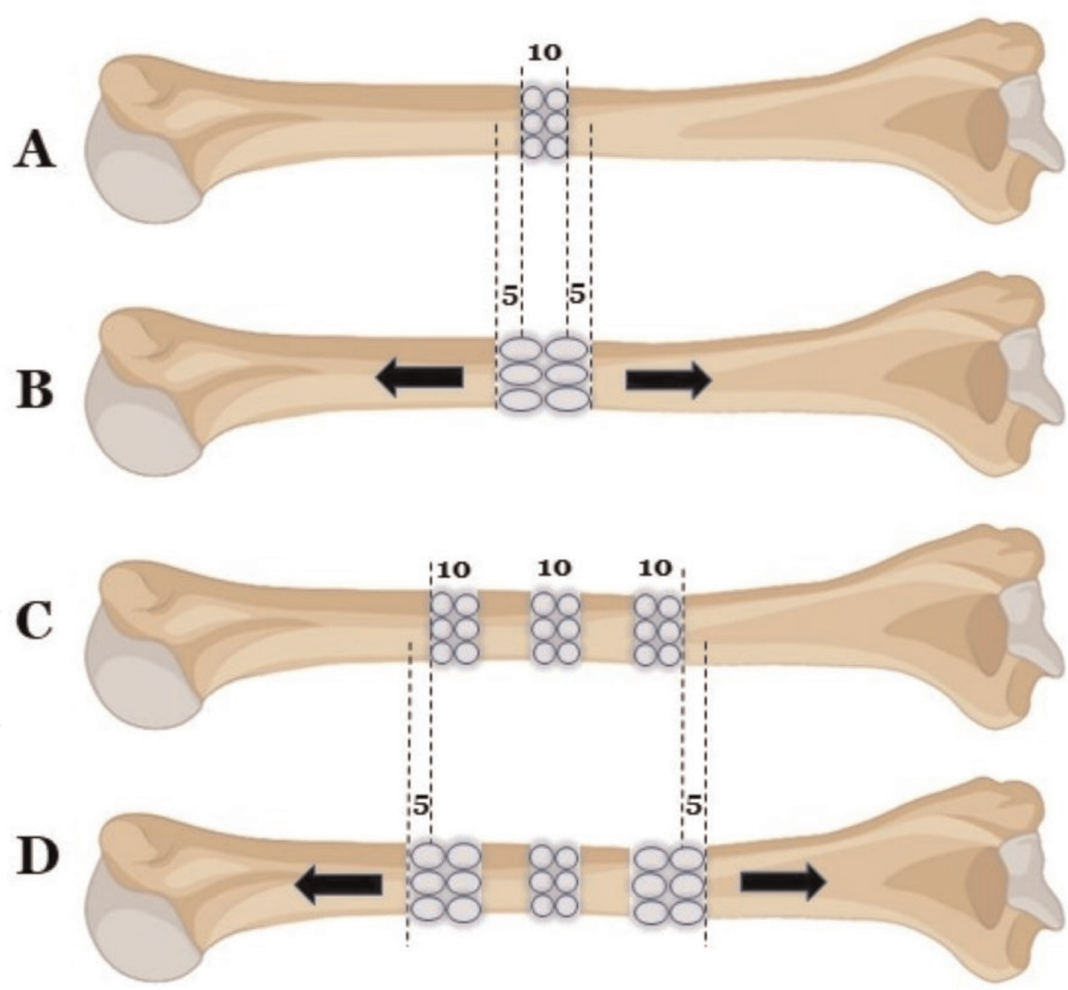

Figure 4.
Illustration of interfragmentary displacement. A and B—simple line fracture with a gap of 10 mm; when the displacement of 5 mm between both fragments occurs, a strain of 100% is produced; C and D—multiple line fracture with a gap between fragments of 10 mm, totalizing a fracture gap of 30 mm; when the same displacement of 5 mm in each major fragment takes place, a total of 10 mm is also added to the fracture gap; however, in this case, the final gap undergoes deformation (strain) of approximately 30%, because the final displacement was distributed over all fragments.

smaller and when a force is applied to a fracture site if the variation of the gap is bigger, the strain also increases.

The illustrative example of two different interfragmentary displacement scenarios is given in **Figure 4**. In the first two images, a simple line fracture with a 10 mm gap between bone fragments is shown. When both fragments are displaced by 5 mm, a displacement (strain) of 100% occurs. In the lower images, a multiline or communitive fracture is depicted, and there is a gap of 10 mm between each fragment, totaling a gap of 30 mm. When the same displacement of 5 mm in each major fragment takes place, a total of 10 mm is also added to the fracture gap; however, in this case, the final gap undergoes deformation (strain) of approximately 30%, because the final displacement was distributed over all fragments.

In the biological context, the strain concept is used to explain the relative deformation and its effects on bone tissue regeneration. In bone callus formation, the tissue can resist a different amplitude of elongation (distancing of fragments). If the movement exceeds the critical value of elongation, there will be dysfunction at a cellular level and, consequently, no delay in the tissue formation. In bone regeneration, the predominant cells in each phase show different tolerance to different magnitudes of elongation movements. As bone regeneration progresses, the tissue is less tolerant, demanding a more rigid mechanical environment (with less micromotion). During the inflammatory phase, the granulation tissue is the most tolerant to movement when compared to cartilaginous or bone tissue in subsequent phases (**Table 1**).

To illustrate the difference in instability tolerance between a simple fracture and a multi-fragmentary fracture, consider the following scenario:

Assuming both fractures have the same initial gap width (5 mm) and overall displacement (5 mm) in (A) and (B), the full displacement (5 mm) is active within a single gap in a simple transverse fracture (A), resulting in a strain of 100%, which is the limit of tolerated strain for granulation tissue. In contrast, in a multi-fragmentary fracture with five gaps (B), the overall displacement is shared among the gaps, resulting in each gap displacing from 5 mm to 6 mm, and the resulting strain is only 20% [3].

Additionally, and from a mechanical point of view, different fracture patterns presented different strain behaviors when subjected to the same stress load. Generally, the larger is the lever arm, the more movement at the interfragmentary interface will be observed; this occurs in single-line fractures (transverse and oblique) in which the fragments are long relative to the fractured section. In the previous scenario, the gap is small and the variation is large, determining a high-strain fracture environment. In simple words, these fracture lines are more sensitive to load/movement forces. In the opposite scenario, multiline fractures, the lever arm is smaller (multiple fragments smaller in length) and the total interfragmentary space (gap) is inevitably larger; with greater gap and equal length variation, with the same stress/load forces applied, the strain value will be smaller (Eq. (4)). For this reason, these fractures are considered low strain or less sensitive to movement or load forces [3].

Tissue type	Tolerance to elongation (%)	Tolerance to shortening (°)
Granulation tissue	100	40
Cartilage	15	5
Bone	2	0.5

Table 1.
Tolerance of tissues of the osteogenic pathway to elongation and shortening.

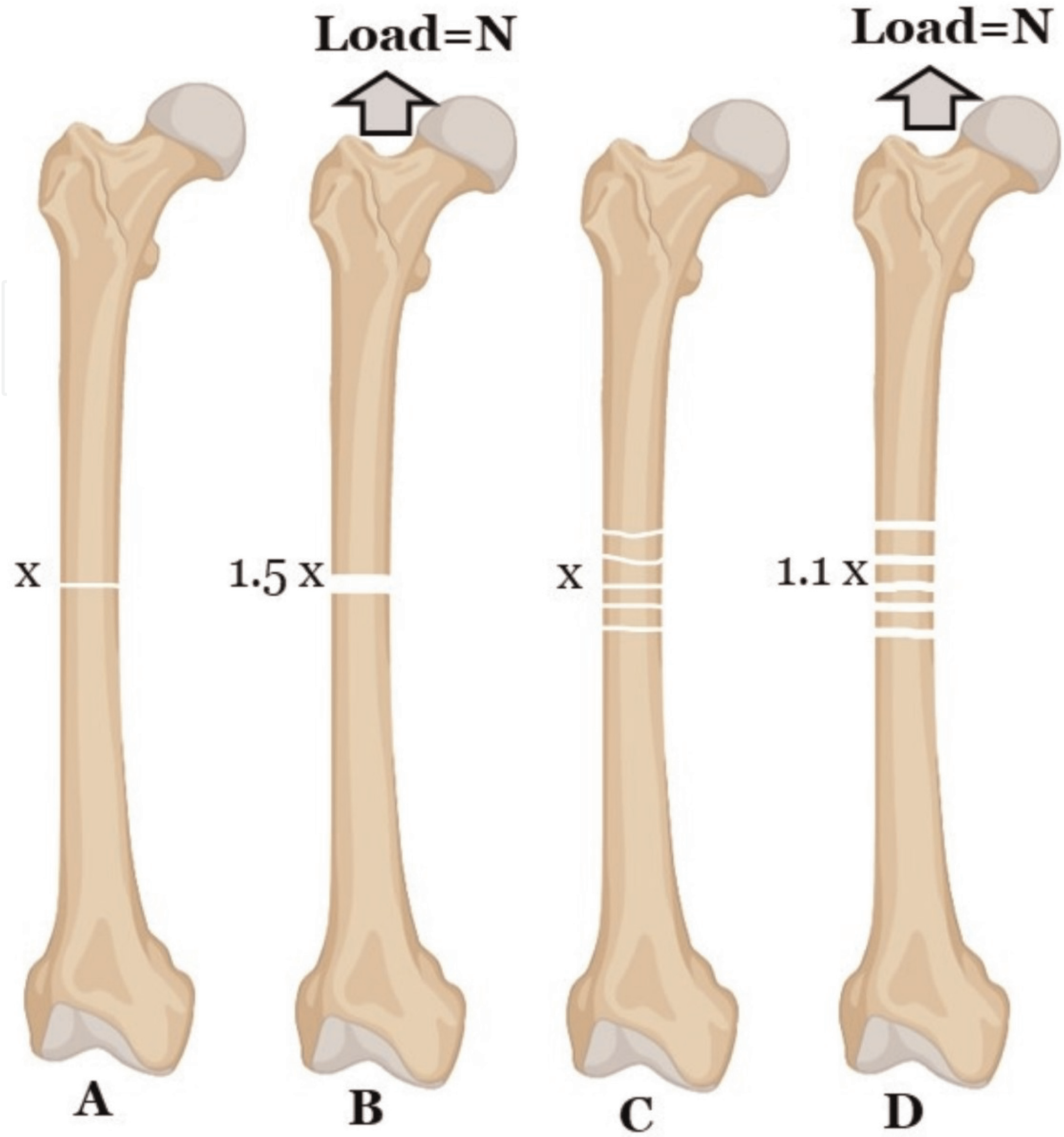

Figure 5.
Illustration of fracture lines with high and low strain. High-strain fracture pattern, A and B—fracture line without load and with a little gap (x); when the bone is loaded, the fracture gap increase by 50% (1.5 x); low-strain fracture, C and D—multiple line fracture with a gap equivalent to fracture line in A; however, when the bone is loaded, the displacement is more subtle among fragments (1.1 x) because the displacement is distributed between all fracture fragments.

Clinically, for the surgical decision-making process, the surgeon should consider two groups of fractures based on the strain theory:

1. High-strain fractures (simple fractures with long fragments) are more sensitive to motion, and interfragmentary movements can easily delate the process of bone regeneration (**Figure 5**). Knowing that, we must choose a method of osteosynthesis which leads to anatomical reconstruction with interfragmentary compression and rigid fixation (reducing the gap and ΔL).

2. Low-strain fractures (multiple line fractures or complex fractures with smaller and short fragments) are more tolerant to movement, allowing more load without major consequences (least chances to overcome tissue tolerance) (**Figure 5**). In these cases, a more elastic and less invasive approach must be

chosen over osteosynthesis methods. This type of fracture is caused by high-energy trauma with more need to preserve the soft tissue envelope of the fracture. Considering this scenario, bone osteosynthesis methods rely on bridging plates (biological osteosynthesis plates) and external fixation.

It is important to remind that several biological and clinical factors influence bone regeneration such as age, time elapsed since trauma, trauma intensity, and soft tissue disruption (open or closed fracture). For this reason, the decision process in bone osteosynthesis is multifactorial and should not be based exclusively on mechanical factors like the strain theory.

2.10 Wolff's law

Bone is a dynamic tissue, and the response to internal and external mechanical stimuli can determine bone density and the organization of bone trabeculae and correlate with the magnitude and direction of compressive and tensile stresses of loading. In the late nineteenth century (1892), Wolff's law was proposed by Julius Wolff, a German anatomist and surgeon, as a mathematical law that described the response of bone to mechanical loading. This law described the functional adaptation of bone to mechanical loading and is supported by several experimental and comparative studies over time. Increases in the loading of bone tissue are known to generate the formation of new bone tissue, which increases mechanical rigidity [6]. Similarly, decreases in mechanical loading, particularly associated with prolonged non-weight-bearing lameness, lead to adaptive resorption or osteopenia of bone tissue conducting to a decrease in mechanical rigidity [6]. One of the classic examples of Wolff's law is the femoral trochlear groove formation by the pressure that the patella exerts on the bone. In dogs with congenital medial patellar luxation, with the lack of pressure by the abnormally positioned patella, the trochlea can be shallow or absent. Another practical example of this law is that bone is generally stronger and stiffer in the direction in which the greatest loads are most commonly imposed (e.g., the long axis of the femur).

2.11 The piezoelectric effect

The piezoelectric effect is a physiological characteristic of certain materials that generates an electric charge in response to a supported mechanical load. The suffix Piezo is derived from the Greek *piezein*, which means to press [7].

Bone has piezoelectric properties because of the highly oriented and patterned structure of collagen type I, and collagen's ability to respond to mechanical loads [7, 8]. When a shearing force is applied to collagen fibers, and the bundles glide past each other, an electric charge is generated. Collagen also has significantly lower elastic moduli than the bone's corresponding mineral component, which makes collagen microfibrils experience the greatest load when strained. Experiencing the greatest load under force deforms collagen fibbers, and this deformation leads to the piezoelectric effect [8]. The role of collagen's piezoelectricity in bone regeneration and remodeling is related to the formation of electric dipoles that stimulates the osteoblasts to promote mineral deposition in the extracellular matrix, increasing bone density. Clinically, when a fracture occurs, the collagen's piezoelectricity is potentiated with an additional mechanism for osteocytes to sense areas with more stress; the generated piezoelectric charge would be greater in stressed areas, which is produced when the bone suffers deformation and negative charges are produced on the tension side and positive charge on traction side, generating bone growing by electric current [7].

3. Biomechanics of bone fracture

Bones from the appendicular skeleton are continuously subject to physiologic and non-physiologic mechanical forces. Physiologic forces are generated through weight-bearing and muscular contraction during physical activity or even at rest. Didactically, it is established that force vectors act isolated in long bones such as flexion or bending, axial compression, tension, and shear and torsion forces; however, clinically, one force vector is predominant. In healthy animals, the physiologic loads applied to bones rarely exceed the yield point, or more practically, physiological forces do not cause plastic (permanent) deformation of bone (**Figure 6**). Nonetheless, when non-physiologic forces are the result of externally applied loads (vehicular trauma, horse kick, fall from height, and gunshot), it is easily exceeding the yield point and load-bearing capacity of the bone is easily exceeding, and as consequence, fracture will occur (**Figure 6**). The resistance of the bone will vary according to the direction of the load, and depending on the intensity and type of forces applied to the bones, different fracture lines will form. In general, oblique fractures originate from supraphysiological axial compression forces; transverse fractures are related to tension (avulsion) and flexion forces applied on opposite sides of the long bone; spiral fracture lines are expected from torsional forces, which create an angular line running around the circumference of the bone and a longitudinal one joining the two ends of the spiral. The combination of forces will give rise to other patterns of fracture lines, such as segmental fractures (butterfly fragment) caused by shear failure on the compression side before the tension fault line; it then propagates throughout the bone, creating the compression side plus a fracture line, usually a single line fragment. The combination of flexion and compression forces potentially generates early failure on the compression side, and a larger fragment will break loose (major butterfly fracture).

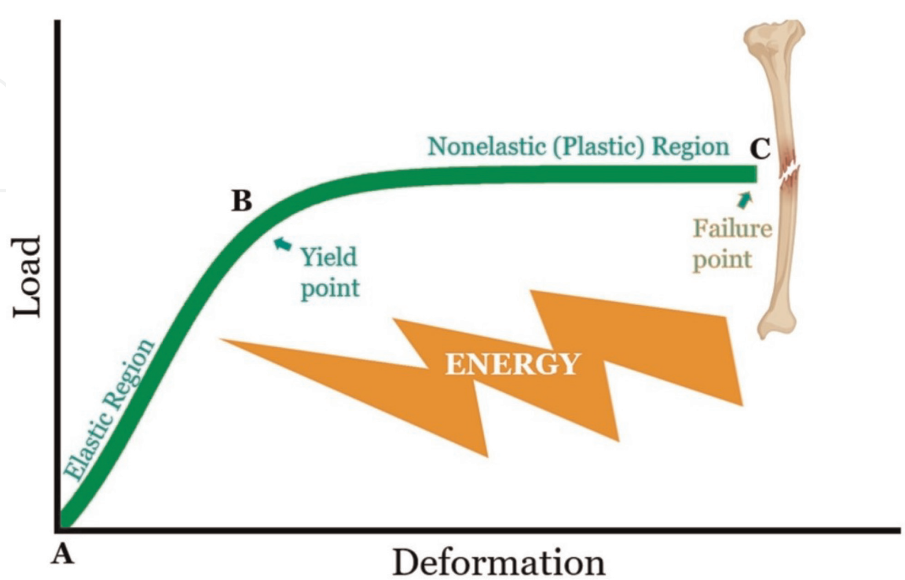

Figure 6.
Load/deformation curve of long bones (A—starting load, B—yield point of deformation, C—fracture or failure point).

The minority of clinically encountered fractures are a result of pure isolated loading; the majority of fractures presented to a veterinary surgeon are caused by more complex loading situations. Most clinical fractures are produced by a combination of three or more loading modes, resulting in a fracture line initiation and progressing in numerous planes. For example, a fracture caused by vehicular trauma is often the result of a combination of bending, shear, and torsional loads. Additionally, the values of the different loads would most likely be different, causing further variations in the fracture patterns observed.

Fractures caused by high-energy trauma (e.g., road traffic accidents) involve the greater accumulation of energy associated with the combination of forces, often resulting in greater fragmentation (comminuted lines). These fractures will also cause greater muscle damage and vascular compromise. The direction in which the energy is applied to the bone is as important as the intensity with which it propagates. Energy is absorbed by the bone and then released with the fracture. Damage applied to soft tissue and bone is proportional to the amount of energy released, and it is concluded that complex fractures are associated with greater soft tissue enveloping lesions.

3.1 Forces acting on bone fracture

3.1.1 Bending

The definition of bending comprehends the axial compressive load that is applied eccentrically (off-center) to the bone column. Studies have demonstrated that 85–89% of the predominant normal physiological stresses that most bones experience during weight bearing are from bending loads due to the curvilinear shape of the femur, humerus, and, in some breeds, tibia e radius [1]. In these cases, the axial compressive load is applied eccentrically during locomotion [1]. The convex face of the bones experiences the maximal load of tension forces, whereas the concave side experiences the maximal load of compression forces. The bone column experiences a gradient distribution of this opposite force perpendicular to the bone, the axis being the center

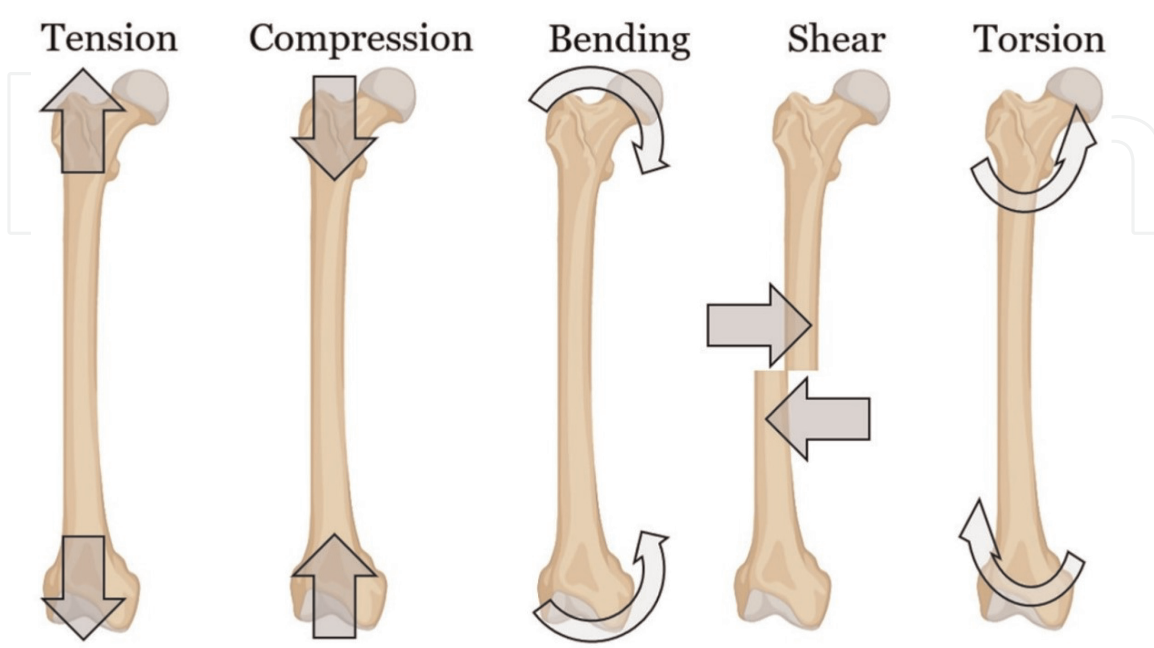

Figure 7.
Different force vectors acting on long bone fractures.

of the medullary cavity, an imaginary point where there are no tensile or compressive stresses and strains. This plane is called the neutral axis or neutral plane of the bone and experiences no axial stress (**Figure 7**) [1]. The magnitude of the compressive and tension forces acting on the bone increases as the distance from the neutral axis increases (**Figure 7**). When a supraphysiological bending load exceeds the yield point and the load-bearing capacity of bone, for example, in extremely soft bone such as immature or diseased bone, when it is subjected to a bending load. A fracture line starts at the tension side (because the cortical bone is weaker in tension than in compression) and propagates to the compression surface, producing, in most cases, a transverse fracture.

If an internally generated shear stress is added to these forces, it results in a short oblique fracture line toward the compression band surface of the bone. Two oblique fracture lines can occur near the compression surface, if the magnitude of shearing forces increases, forming a loose wedge. This fracture pattern is referred to as a butterfly fracture and is a result of two divergent planes of shear stresses near the compression surface.

3.1.2 Compression (axial)

Compression loading, also called axial compression, is produced when equal and opposite loads are applied toward the center and parallel to the longitudinal axis of the bone, causing compressive stress and strain within the bone (**Figure 8**). In long bones, compressive loads cause a decrease in height and an increase in width. Maximum compressive loads occur on a plane perpendicular to the applied load and can be defined by a series of small forces directed toward the center of the bone that

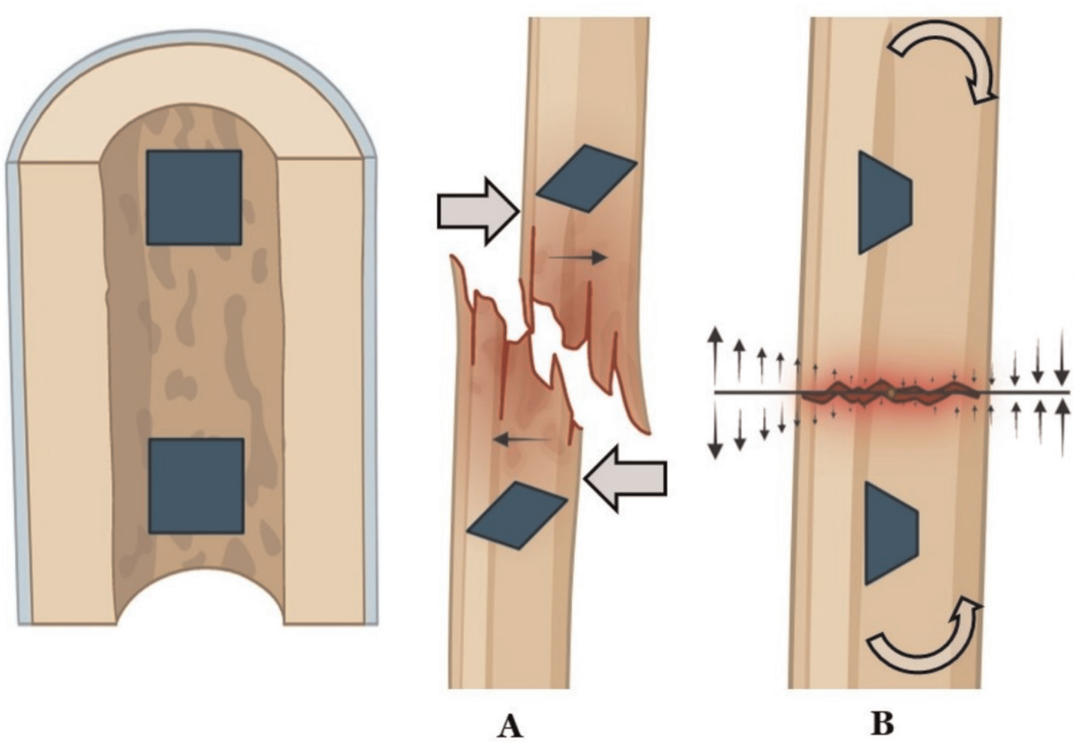

Figure 8.
Illustration of shear and bending forces acting on a long bone, A—shear loading causes angular deformations, B—bending loading induces tensile loading along the convex surface and compressive loading along the concave surface causing a transverse fracture pattern.

potentially can compact or crush the bone. Rationally, we would expect that compression fractures developed perpendicularly to the applied compressive load will crush the bone. Nonetheless, the perpendicular tensile strain is usually not too important, because the expansion of cortical bone is highly unlikely and internally generated tensile strain also develops outward from the center of the bone, perpendicular to its longitudinal axis.

Nonetheless, compression loading also produces internal shear loading that develops oblique to the longitudinal axis and is maximal on a plane of 45° from the axis of compressive loading (**Figure 9**) [9]. Macroscopically, the fracture line of bone loaded under pure compression is typically a short oblique fracture and is created by these internal shear stresses, generated partly because of the bone's anisotropy and the fact that bone is weaker in shear forces and more tolerant in compression loads. These oblique fracture configurations produced by compressive loading are commonly seen clinically with jump or fall injuries of the distal tibia and radius (bones that are loaded along their central axis) [9].

A transverse fracture pattern also can appear as a result of compressive loading and is occasionally seen in vertebral bodies or the growth plates of long bones in young animals, also called impaction or impacted fracture (type V or VI Salter-Harris fracture) [1].

3.1.3 Shearing

Shear loads occur when a force is applied parallel to the bone's surface, causing it to have a tendency to slide past another surface and causing an angular deformation (**Figure 9**). With the shear forces acting in opposite directions on opposing surfaces, shear loads within the bone lead to deforming it in an angular manner (right angles within the bone are deformed to acute or obtuse angles). In general, the bone offers the weakest strength when subjected to shear forces. Therefore, bone fractures along the plane of maximal shear stress. Clinically, fractures developing from shear loading often occur in the metaphyseal region of long bones with high cancellous bone content [9].

A classic example of fracture that occurs in small animals as a result of pure shear loading is the fracture of the lateral aspect of the distal humeral condyle seen in immature animals (Salter-Harris type IV). This fracture occurs as axial compressive forces are transmitted from the foot through the head of the radius to the lateral and/or intercondylar component of the condyle of the distal humerus, resulting in a concentration of shearing forces at these regions of the distal humerus, producing a classical type IV Salter-Harris fracture [1]. Other common fractures created by shear loading would include "T" or "Y" intercondylar fractures of the distal humerus, fractures of the tibial plateau, isolated condylar or intercondylar femoral fractures, fractures of the glenoid cavity of the scapula, vertebral body fractures, and carpal or tarsal bone fractures. As previously described, shear loads also occur in most long bones subjected to pure axial compression, resulting in short oblique fractures along the plane of maximal shear stress [1, 9].

3.1.4 Stress concentration or stress risers

Osteopenia and bone defects on bone structure caused by iatrogenic conditions such as drilled holes (biopsy tract, bone graft collection, or screw removal) or acquired conditions like neoplasia, bone cysts, and bone infection (bacterial or fungal

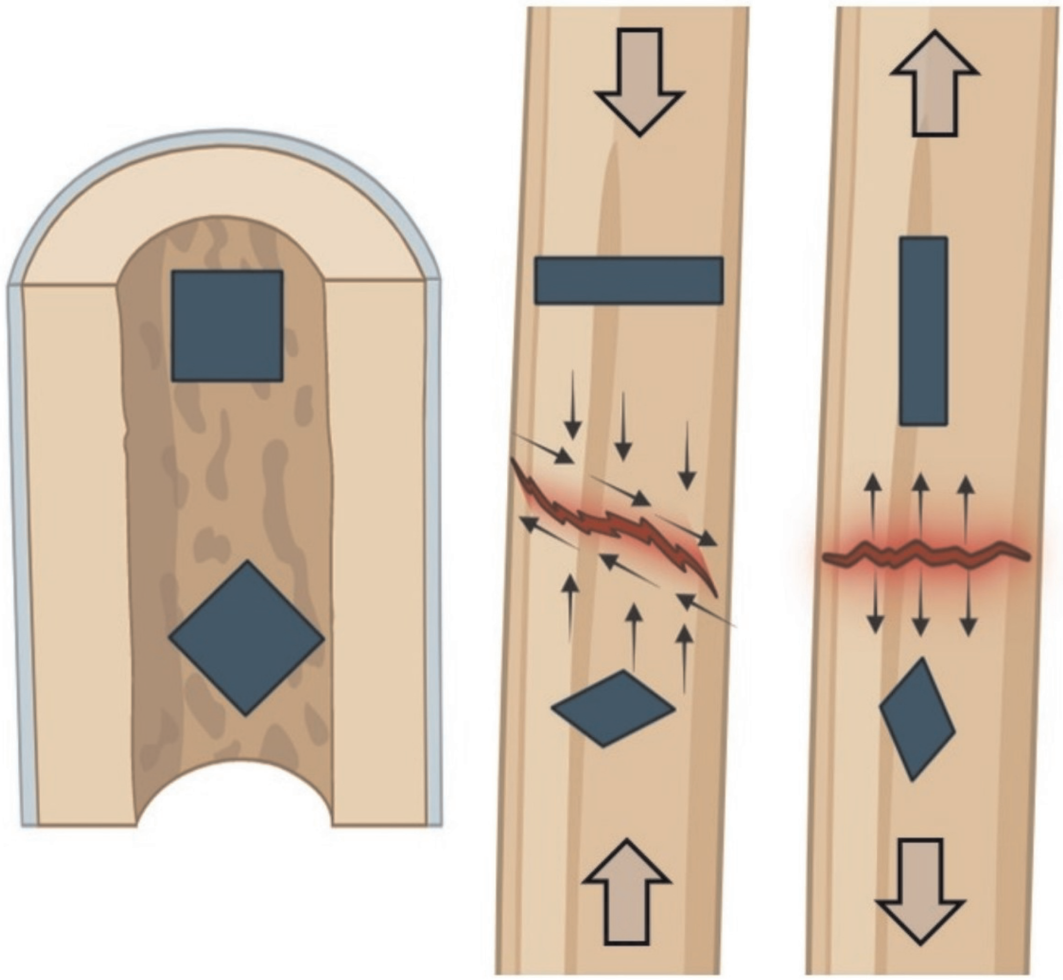

Figure 9.
Illustration of stress and strain produced by compressive and tensile loading. A—Compressive loading induces compressive and shear stresses and strains that, if excessive, may induce a short oblique fracture. B—Tensile loading induces tensile stresses and strains, which, if excessive, induce a transverse fracture.

osteomyelitis) cause stress concentrations in bone that can initiate failures [10–12]. These areas of stress concentration can lead to local stress risers in the bone near the defect, which is many times higher than the stress physiologically applied to the bone. The concept of the stress concentration effect is based on the mechanical phenomenon that physiological loads must flow through the bone and, in a healthy bone tissue, without defects or heterogeneity. The applied forces flow equally through all regions, creating equal stress throughout. However, in bone with defects (e.g., holes from removed screws), the load cannot flow through the areas with defects and thus must flow around the holes. This leads to a concentration of stress in points adjacent to defects and osteopenia areas [9]. The clinical consequence is that bone stress-rising points break at lower loads than homogenous bones. The weakening effect of a stress riser is particularly noted for torsional loading where the decrease in strength may approach 90% and is proportional to the defect size [10]. However, defects smaller than 10% of the bone diameter may be of negligible significance in torsional resistance and may resist under physiological loadings [13].

Another form of stress concentration is illustrated by the difference in the elastic moduli (stiffness) of two materials (e.g., stainless steel and bone) under load. The stress concentration in this example is due to the evidence that the modulus of a material determines its response to an applied force: with high moduli materials, the strain or deformation is inferior to low moduli materials for equal load. As a consequence of that difference in modulus values, the flow of homogenous force is interrupted causing stress concentration. Frequently, clinical cases are reported in bone areas at the limits of the joint prosthesis or stainless-steel bone plating. As the materials are loaded, the bone exhibits greater elastic deformation, creating shear stress at the bone-implant interface [9].

3.1.5 Tension

Tensile loading of bone results when equal and opposite loads are applied away from each other outward from the bone's surface and along its longitudinal axis, as a result of supraphysiological stress, causing the fracture line to be orientated on a plane perpendicular to the axis of loading (**Figure 9**). This mode of loading is primarily due to the contraction of muscles or the effects of ligaments and tendons at bone prominences such as tuberosities, tubercles, and trochanters, where a pure tensile loading is exerted over their cross-sectional area. Clinically, fractures with transverse patterns perpendicular to the applied load are predictably produced and often seen at traction of apophyses such as the olecranon process, tuber calcaneus, and tibial tuberosity (**Figure 9**). Fractures of the patella and avulsion fractures of ligamentous insertion are also exampling where tensile forces predominate and cause a transverse fracture. Because cancellous bone is much weaker under tension than cortical bone, fractures occurring due to tensile loading often occur in regions that have more cancellous than cortical bone, such as bone prominences.

3.1.6 Torsion

When a torsional load is applied to a long bone in such a manner that causes it to twist about an axis (usually the long axis of the bone), that results in the generation of shear, tensile, and compressive forces (**Figure 10**). Specifically, the torsion force causes a shear stress that is distributed throughout the bone. As in bending, there is a gradient in the magnitude of loading, proportional to their distance from the central (long) or neutral axis. Under torsional loading, maximal shear stresses are produced on planes perpendicular and parallel to the central axis. Tensile and compressive stresses are distributed perpendicular to each other and on a diagonal plane to the neutral axis (**Figure 10**). The fracture begins along a plane of maximal shear stress orientated parallel to the neutral axis. The fracture then propagates along the plane of maximal tensile stress creating the typical spiral fracture configuration. Clinically, spiral fractures are commonly seen in the narrow diameters of the distal tibial and distal humeral diaphysis, where the area moment of inertia is relatively small (thus the resultant shear strain from torsional stress is relatively high).

4. Biomechanics of bone tissue

The fracture behavior of bone is influenced by its viscoelastic, anisotropic, and heterogeneous mechanical properties. The stress-strain behavior of bone is dependent on the rate of loading, which is characteristic of a viscoelastic material [3, 14]. If the bone is loaded at a high rate, such as occurs with vehicular trauma or gunshot injury,

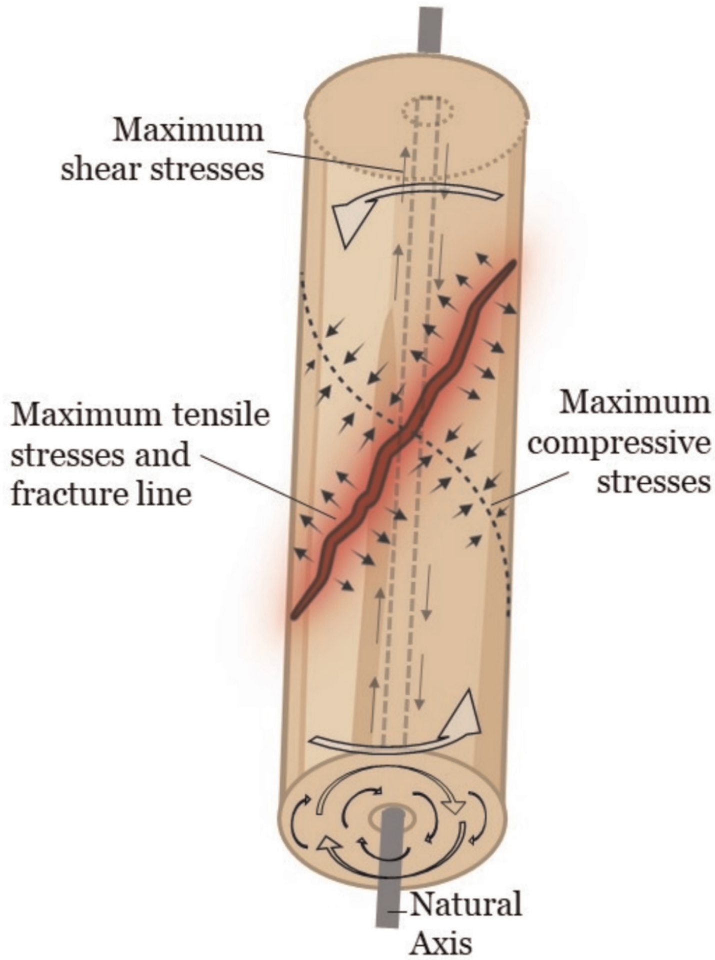

Figure 10.
Illustration of the shear, tensile, and compressive stresses and strains at supraphysiological loads causing a spiral fracture pattern.

its stiffness (Young's modulus), ultimate strain, and energy-to-failure increase. The clinical significance of the high toughness of healthy bone is that if a high-rate loading causes macroscopic failure or fracture, as opposed to just distributed microscopic interfacial failures, the large release of the absorbed energy will cause marked comminution and injury to surrounding soft tissues [15]. Bone is considered a material with anisotropic properties; as a consequence, the values of strength and stiffness are a function of the direction of applied loads regarding bone structure (**Figure 11**) [3, 14].

4.1 Cortical vs. cancellous bone material properties

All bones are composed of a combination of cortical (compact) and cancellous (trabecular) bone. Both cortical and cancellous bones are formed from an inorganic mineralized matrix called hydroxyapatite, which is primarily calcium and phosphate. Hydroxyapatite (HA) is a naturally occurring calcium phosphate mineral characterized by the chemical formula $Ca_{10}(PO_4)6(OH)_2$. HA-like compounds compose approximately 60–65% of bone's dry weight [16]. The inorganic matrix is combined with an organic nonmineralized matrix (35–40% of bone's dry weight) [16]. By contrast, the organic extracellular is significantly more complex and consists mainly of

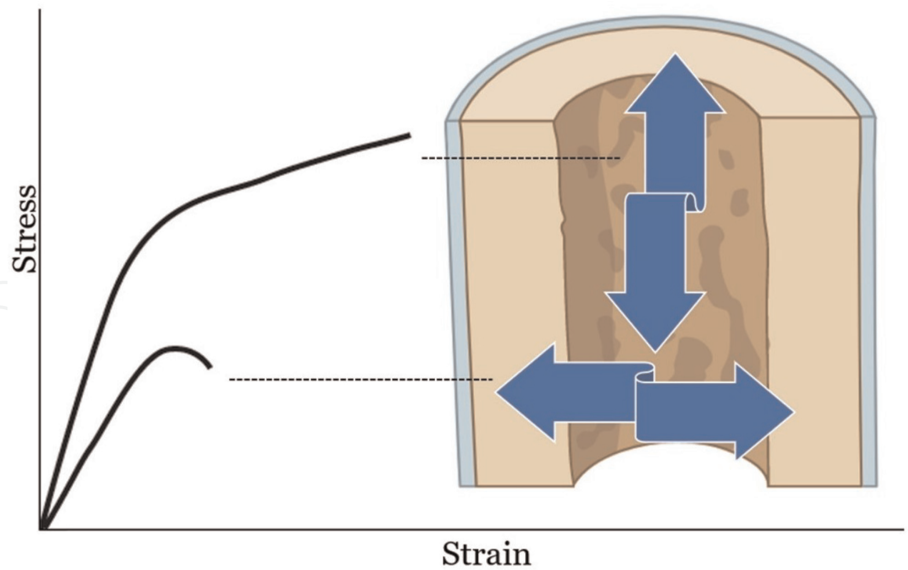

Figure 11.
Stress-strain curve depicting the anisotropic behavior of bone. Load forces of tension were applied in two different orientations: parallel and perpendicular to the longitudinal axis.

collagen type I (90%) and noncollagenous proteins (10%) such as glycosaminoglycans, water, and cellular elements [16]. The inorganic matrix imparts strength and rigidity to the bone, and the organic matrix gives it flexibility and resiliency [1].

The cortical bone always surrounds the cancellous bone; however, the relative quantity of each type varies from one bone to another as well as according to the specific location within a particular bone (diaphysis vs. metaphysis or epiphysis); cortical bone is designed to give strength and stiffness to the bone [3]. From a mechanical standpoint, cancellous bone is designed to absorb a tremendous amount of energy and transmit load [1].

Both cortical and cancellous bones have inorganic and organic components; however, one of the primary differences between both bone types is the different percentages of organic versus inorganic matrix of each type. Structurally, this difference influences the porosity and apparent density and consequently the mechanical behavior of each type of bone when submitted to loads.

Porosity is defined as the volume of bone occupied by nonmineralized tissue. Cortical bone is composed primarily of inorganic mineralized matrix and therefore has low porosity. The porosity of cortical bone has been estimated to vary from 5% to 30% and in the cancellous bone, it can vary from as little as 30% to as much as 90% [17].

Apparent density is a measurement related to porosity and is directly related to its inorganic mineral content, being the mass of the bone tissue divided by the bulk unit volume of bone tissue, including mineralized bone and marrow space [17]. Cortical bone typically has a higher apparent density than cancellous bone tissue [17].

The differences in porosity, or apparent density, between cancellous and cortical bone dramatically affect their behavior when the two types of bone are submitted to loads (**Figure 12**). Cancellous bone initially exhibits elastic behavior followed by a yield, which occurs as bone trabeculae begin to fracture. After the yield point, a long plateau of plastic deformation occurs as a result of progressive fracture and collapse of additional trabecular bone and marrow spaces (**Figure 12**). Once the entire marrow

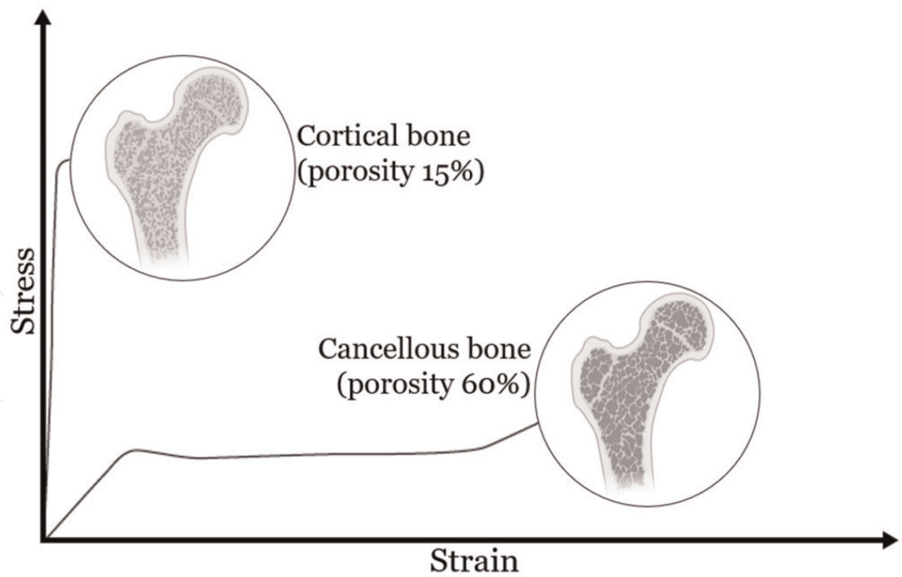

Figure 12.
Different stress/strain curve profiles for cortical and cancellous bone.

space has filled with debris from fractured bone trabeculae, which is referred to as pore closure, there is a marked increase in stiffness before the ultimate failure point of cancellous bone is reached. Under compression loading, cancellous bone exhibits a stress-strain behavior similar to that of soft porous metal. When compression loading is applied, cancellous bone can absorb a large amount of energy (when compared to cortical bone) and can tolerate strain values up to 7% before structural failure·.

On the contrary, cortical bone, due to its low porosity, presents a brittle behavior when subject to compressive loads, similar to glass. Cortical bone is characterized by a decreased plastic deformation phase before failure, absorbs less energy, and tolerates lower strain values (<2%) before fracture as compared with cancellous bone (**Figure 11**). However, cortical bone has greater ultimate strength and increased stiffness and can tolerate more force loads before fracture than cancellous bone.

The clinical implications of the relationship between bone's apparent density and its mechanical behavior are evident when large changes in the strength and modulus of bone can result from small changes in its apparent density. In the clinical setting, the reduction of apparent density is evident on radiographs only when lost by 30–50%, and consequently, the reduction in bone density detected on radiographs is associated with greatly reduced stiffness and strength [1]. Conversely, greatly enhanced fracture zone stiffness and strength may be present even with minor increases in fracture zone density observed in radiographs.

5. Applied fracture biomechanics to common clinical presentations in small animal osteosynthesis

When the concepts of fracture biomechanics are applied to clinical situations, in simple terms, it is possible to define strain as movement and stress as force or a magnitude of load that is applied to the bone and/or the implant.

5.1 Strain

The plate strain is the strain (movement) experienced by a plate when a load of the force vector is applied to it. More specifically, it is the amount of movement that the plate experienced with a certain force (proportional to the original length). Areas of high strain on the plate are areas of high stress. Areas of high plate strain should be avoided because small increases in stress on a plate decrease the fatigue life of an implant. The majority of implant failures after small animal orthopedic surgery are fatigue failures.

It is generally accepted that strain/movement at the fracture gap needs to be within the tolerable levels for tissues. The fracture stability dictates the type of healing that will occur. With strain below 2%, primary bone healing can occur, whereas at 100% strain, the only tissue that can form is granulation tissue. In secondary bone healing, the initial tissue at the fracture site in the inflammatory phase of bone regeneration is granulation tissue. The tissue then progressively stiffens until cartilage can form. Cartilage has a strain tolerance of around 10%. *In vitro* data suggests that callus is stimulated at strains of around 5–10% and bone is stimulated at strains between 1 and 5%. The bone formation starts in lower strain zones at the periphery near the periosteum before spreading inward across the entire fracture gap. .

5.2 Stiffness

The concept of stiffness can be thought of as the magnitude of movement when a force is applied (it is the slope of the stress/strain curve). If the implant is stiff, it does not move when force is applied. One of the determinants of stiffness is working length. If the working length is increased, the stiffness decreases. This means more movement of the plate and higher stress (and higher strain). However, if the working length of the plate decreases, the stress and the strain will be concentrated in a smaller area, which can also predispose to plate failure.

5.3 The strain paradox

Stoffel et al. found that in an *in vitro* situation of a 1 mm simple fracture gap, the strain experienced on the plate in tension bending was lower with a long working length [18]. However, if you have a more flexible plate, the fracture ends touch and suddenly load sharing is produced and therefore less movement and lowered strain, however, only in tension bending. Although with a stiffer plate, the plate does not bend in tension bending. Basically, in the situation of a 1 mm gap, the strain was paradoxically decreased with a longer working length. However, this phenomenon can be explained by the fact that fracture ends touched when the bone is loaded, preventing further movement of the plate in the '*in vitro*' situation. If strain and stress on implants are increased, the fatigue life of a plate decreases. If the fatigue life of a plate from 100,000 cycles is reduced to 10,000 cycles, this could be the difference between the fracture healing before implant failure and catastrophic failure requiring surgical revision. .

5.4 The concept of micromotion

It is widely recognized that micromotion contributes to fracture healing by stimulating the formation of bridging calli. Osteosynthesis methods that are based on relative stability allows micromotion creating a biomechanically optimal construct for secondary bone healing by promoting bone callus formation and has already been associated with early bone healing in several high-risk cases [19]. On the other hand, delayed unions resulting from insufficient mechanical stability, or hypertrophic non-union, may also be associated with large callus formation.

The concept of micromotion is also applied to joint prostheses at the bone-implant interface Excessive micromotion of an implant in bone renders bone ingrowth impossible and reduces osteointegration of prosthesis. The tolerated minimal movement within an interface has been reported to be 28–150 μm, and repetitive higher displacements values allow only the ingrowth of fibrous tissue to avoid osteointegration [20]. Micromotion magnitude is primarily a function of implant stability, although is influenced by the differences in the elastic modulus of bone and implants.

Axial micromotion can be created with circular external skeletal fixators (because the wires allow motion at the fracture site that is axial in direction), with some configurations of interlocking nail and with special plate designs. However, when using a locking plate with a long working length, the micromotion observed at the fracture site is characterized by not only an axial vector; it is also multidirectional.

Besides the influence of the magnitude of micromotion, the characteristics of interfragmentary micromotion are also influential in bone healing. Applying cyclic interfragmentary micromotion for short periods has been shown to influence the repair process significantly [21]. In a study by Goodship et al., it was reported that interfragmentary cyclic micromovement applied for the short term at a high strain rate produced a greater amount of periosteal callus when compared to the same stimulus applied at a low strain rate. It was also shown if a high-strain-rate stimulus is applied later in the regeneration period, this physiological process was significantly inhibited [21]. The beneficial effect of this particular biophysical stimulus early in the healing period may be related to the viscoelastic nature of the differentiating connective tissues in the early endochondral callus. In the early endochondral callus, high rates of movement induce a greater deformation of the fracture fragments because of the stiffening of the callus [19, 21].

An experimental study proved that stimulation of new bone formation by dynamization with micromovement was effective mainly in the early healing phase (4 weeks postoperatively), while dynamization had no significant influence in the late healing phase (8 weeks postoperatively). The beneficial effects of micromotion are hampered by the influence of the gap size in the healing process [22]. From that evidence, with dynamization, the negative effects related to a large gap size overcome the positive effects of dynamization [22]. If a flexible fixation of a simple diaphyseal fracture is performed in clinical practice, the fracture gap should therefore be reduced to as small as possible. But if for some reason a large fracture gap cannot be avoided, dynamization (i.e., enabling micromovement) of the fracture should be performed very carefully and only in the first weeks postoperatively [23]. A large callus formation does not necessarily lead to greater mechanical stability [23, 24]. From that conclusion, was not the size of the radiological evident callus, but the amount of newly formed bone of the peripheral callus that was important for gaining mechanical stability. After the early healing phase, a large amount of new bone is formed, which is mainly responsible for the biomechanical stability of the fracture line.

The amount of callus, more specifically the periosteal callus, is, to some extent, related to the flexural rigidity of the fracture. Research that has found a consistently positive effect of interfragmentary movement on the mechanical stability of regenerating bone has applied only small and controlled interfragmentary movements in the early healing phase [25] or allowed larger movement and loads in a later phase [26].

5.5 Excessive stiffness of implants

The concept that stiffer implants delay bone healing assumes that a callus cannot be formed when strain conditions are too low. In a situation where strain is 0%, potentially this could delay healing. However, it is an unlikely situation in clinical scenarios. When animals use the limb, the amount of force applied always causes a strain value at the fracture site over 0%, and for that reason, there will be an unrealistic complication in small animals. Under optimal stiffness repairs are much more common in veterinary patients, often delayed, and nonunions are a consequence of inadequate addressing of fracture mechanics and/or poor biology versus too stiff implants. Low-strain environments created by stiffer implants facilitate haversian canals and faster bone regeneration.

5.6 The concept of elastic osteosynthesis

This concept is very specific to juvenile dogs and cats. Different breeds of dogs reach skeletal maturity at different ages; it is considered that the physiological process is finished between 5 months (toy breeds) and 18 months (giant breeds) through a very rapid and biphasic growth rate. During the initial growth phase, both structural and material properties of immature bone are considerably different from those of adult bone and are characterized by lower strength and stiffness, as well as lower yield stress and elastic modulus. Additionally, the diaphyseal cortices are thinner but have a more robust periosteum in young animals compared to in adults. As a consequence, immature canine bone is more predisposed to implant failure due to screw pull-out. In addition, due to the rapid initial growth phase and the natural flexion angle of the elbow and knee, postoperative immobilization of these joints in young dogs will inevitably lead to ankylosis secondary to adhesion formation and muscle contracture. In the hind limb, if the functional recovery does not happen early on after osteosynthesis, fracture disease leads to irreversible loss of function due to muscle contracture even after a few days of immobilization. To prevent this debilitating complication, early osteosynthesis is recommended to promote controlled postoperative mobilization, which can lead to implant failure due to the hyperactive nature of non-leash-trained puppies.

The use of overly rigid fixation in juveniles can lead to concentrated forces at the screw-bone interface. In the situation of a standard cortical screw, in this poor-quality soft juvenile bone, this could result in poor screw purchase, screw loosening, and subsequent implant failure, mostly due to screw pull-out. This situation is less common with locking screws, as for a locking screw to fail, it needs to cut through the bone.

Regardless of the osteosynthesis technique chosen and used in juvenile or pediatric dogs, physes must be preserved at all cost. This absolute requirement contraindicates the use of any intramedullary implants (e.g., pins or interlocking nails) especially during the first, rapid growing phase where the physes are more sensitive to traumatic closure. The external fixation is not the technique of first choice for the osteosynthesis of humeral or femoral diaphyseal fractures in young dogs due to mechanical and biological reasons. Namely, the outward position of the external fixator construct, away from the neutral axis of the bone, elevates the bending stresses at the pin/bone interface, promoting a stress riser point. These osteosynthesis technique is also prone to early failure due to implant pull-out, and the use of positive profile transfixation pins does not reduce this complication. From a biological standpoint, the transfixation of the thigh or arm musculature increases the exudation at pin/soft-tissue interface due to excessive movement and generates postoperative pain avoiding free range of motion (ROM) at the knee or elbow. The resulting loss of ROM potentially leads to muscle contracture. Due to the potential complications associated with intramedullary pining and external fixation techniques, plate osteosynthesis remains the treatment of first choice for diaphyseal fractures in juvenile dogs However, if the AO principles of anatomical reduction and rigid internal fixation were used routinely in early growth phase, it can result in catastrophic implant failure via screw pull-out, which leads to the creation of elastic plate osteosynthesis technique (EPO). The technique relies on the increased overall compliance of the bone/plate construct to reduce the risk of focal failure of the screw/bone interface. EPO is used in conjunction with minimally invasive surgical strategies (MIS) favoring restoration of alignment rather than anatomical reconstruction and percutaneous sliding plate techniques to further decrease postoperative morbidity and stimulate early functional recovery. The plates used in EPO were mainly veterinary cuttable plates preferably with locking screws used in a bridging function without anatomical reduction and hematoma disturbance due to their favorable effects on indirect bone healing. Indirect fracture reduction is accomplished by traction on the distal fragment with small fragment forceps and/or using the plate. Large fragments or an oblique fracture should be reduced with the aid of pointed reduction forceps but without attempting a precise reduction. Since anatomical reduction is not attempted; restoration of the bone length is achieved by determining the appropriate plate length from radiographic views of the contralateral intact bone. Since the fracture site is not exposed, it is beneficial to verify proper alignment via intraoperative radiography or fluoroscopy.

The plate is cut to the desired length according to the anticipated position of the screws relative to the growth plates and inserted epiperiosteal through two proximal and distal small incisions. Cortical screws are placed in the two most proximal and the two most distal holes of the plates without tapping to increase bone adherence. In order to decrease pull-out complication, the screws axis should always be oriented in diverging planes in relation to bone longitudinal axis.

The preservation of the strong periosteal sleeve, and the use of an undersized implant such as veterinary cuttable plates (VCP), allow controlled motion at the fracture site, which in turn promotes secondary bone healing via fast callus formation. The flexural or bending deformation of the bone/plate construct is controlled, in part, by the working length of the plate dimension. EPO guidelines recommended that the central plate span without screws should be as long as possible and include no less than 3 consecutive empty screw holes to increase compliance and reduce stress riser effect. This pattern of screw distribution increases the working length of the plate and therefore its compliance. As a result, it decreases the stress riser effect of a single empty screw hole, thus reducing the risk of implant fatigue failure. Furthermore, the enhanced compliance of the bone/plate system lowers the stress on the interface between the bone and screw, thus decreasing the possibility of screw pull-out. Another strategy to decrease screw pull-out complication would be the use cancellous screws instead of cortical screws. The cancellous screw has larger threads and a higher pitch as compared to the cortical screw, which makes its use indicated in metaphyseal bone, osteoporotic bone, or low-porosity bone as found in young patients.

The use of minimally invasive (percutaneous) plate osteosynthesis in conjunction with EPO further reduces postoperative morbidity and promotes early use of the fractured limb and a rapid functional recovery. With this method of osteosynthesis, bone union was achieved as early as two weeks and in all cases at four weeks postsurgically [27]. Surgical complications related to implant failures, such as screw pull-out and plate plastic deformation, were not reported. Radiographically, callus remodeling could be visualized two months postoperatively, and the bony union was completed in four months [27]. Diaphyseal growth occurred without complications, and angular deformation was not observed in either epiphysis.

Although weight-bearing and ROM are recommended immediately after surgery, high-impact activities (jumping and rough play), while difficult to be truly controlled, should be avoided. In contrast, controlled physical activities such as leash walking, trotting, and swimming are beneficial to bone regeneration and should be stimulated.

5.7 Osteosynthesis in toy-breed dogs

In toy dog breeds, complications related to osteosynthesis were more frequently reported than in the general population [28]. Delayed or nonunion and stress protection have been documented in long bone fractures of toy breeds as the most frequent complications, with a special focus on the radius and ulna [29]. Refracture after plate removal is a common complication after stabilization of the radius and ulna fractures. Patient factors such as poor intraosseous vascularity and limited periosseous soft tissue coverage predispose small-breed dogs to healing complications [30].

Biological osteosynthesis techniques decreasing iatrogenic surgical trauma while yielding appropriate construct stability would appear to be advantageous for facilitating the healing of these fractures. External skeletal fixation can be used in toy-breed dogs; however, the radius is a very narrow bone, in addition to its elliptical cross section, which makes the placement of transosseous ESF pins technically challenging [28]. Piras et al. reported the use of circular external skeletal fixators (CESF) in radius and ulna fractures in 16 toy-breed dogs, all of which achieved union despite reporting a 40% minor complication rate, including pin and wire tract discharge [31]. Plate osteosynthesis classically is considered a successful surgical option despite the report of major complications in 18% of cases in one study [32]. Nevertheless, more recently published studies have described a reduction in complications overall or implant-related. Hamilton et al. reported a series of 14 toy-breed dogs treated with a T-plate, all of which healed uneventfully [33]. Regarding function assessment, it was graded as excellent in six dogs, good in four, and fair in two dogs. Vallefuoco et al. only reported 9% of implant-related complications with the use of LCP plates, which could explain the lowering of complications over time [34]. Despite MIPO being recommended in this group of dogs due to poor intraosseous vascularity and limited periosseous soft tissue coverage, recent studies have shown that conventional plate fixation of these fractures is not associated with such a high complication rate when fractures are treated with an appropriately sized bone plate. Pozzi et al. reported a retrospective study that radius and ulna fractures managed with MIPO had similar alignment, reduction, and time to union as fractures managed with ORIF [29]. Arburn et al. also reported a low rate of complications (3%) when ORIF for distal radial fractures was used [35].

In toy breeds, any implant has the potential to lead to stress protection, which can cause osteopenia, especially in radius and ulna fractures. This does not mean that the use of flexible implants is an absolute indication for toy breeds. For the same reasons as above, plates without the appropriate stiffness will fail in the same way as for any dog, especially if the anatomical reduction is required and the fracture line is not uniformly compressed, leaving the transcortices without contact subjecting the plate to bending stress and more prone to fatigue failure.

Excessively rigid plate fixation has historically been considered to be associated with stress protection and subsequent osteopenia, which may in part be responsible for increased refracture risk in these breeds [36, 37]. Osteopenia induced by stress-protection has been reported as a frequent (7.1–20%) complication after plate osteosynthesis of distal radial and ulnar fractures in miniature and toy-breed dogs [28, 32, 38]. A low incidence (1.5%; 1/65) of osteopenia was reported in the study published by Aikawa et al., in part because of the selection of appropriate plate size and type (DCP vs. LCP) with a proper technique [39]. Stress protection-induced osteopenia can only be detected by long-term plate application follow-up [37]; therefore, long-term annual radiographic evaluations are needed to diagnose this complication. On the other hand, a recent study has assigned vascular compromise of the bone cortex as the main cause of osteopenia [40]. Stress protection may not be the cause of osteopenia in distal radial and ulnar fractures, and routine plate removal is not necessary when fractures, provided that plates of appropriate size and type are used and soft-tissue handling atraumatic not overlooked [28, 32]. The diameter of the screws used is another factor to be considered. If they occupy more than 30% of the width of the bone radius (as the maximum size allowed), the bone may have reduced bone strength or have impaired vascular supply, and this can be a reason for osteopenia development [41].

Implant-induced osteoporosis (IIO) or osteopenia can be caused by osteonecrosis of the bone occurring just below the plate that causes cortical bone thinning of about 40%, occurring at 24 weeks after dynamic compression by plate placement [42].

IIO is evolved by biphasic changes and is attributed to inadequate blood supply at 8–12 weeks and reduced mechanical bone stress at 24–36 weeks [37]. IIO is a relatively common complication in small dogs, caused by a process of insufficiently developed bone microvessels, after internal fixation with a conventional plate [43].

LCP plates are reported to preserve blood flow to the periosteum and enable angularly stable fixation, leading to increasingly used in small animal orthopedic surgery [44–46]. In contrast to DCP/LC-DCP in which stability is provided by frictional forces between the plate and bone, locking plates allow the plate to be placed away from the periosteal surface and do not require compression of the periosteum, preserving periosteal blood flow and achieving secondary bone healing due to relative stability [46]. Preserving periosteal blood flow during fracture treatment is an important factor for bone regeneration; as long as the blood flow is preserved, the risk of infection and IIO is reduced. LCP plates due to reportedly small periosteal contact areas reduce the risk of early postoperative osteoporosis and should be the main option for distal radial fractures in toy breeds [47].

Regarding the material used for plating, the comparative studies for the most common alloys used (titanium vs. stainless steel) did not show different results regarding stress shielding [48, 49]. However, titanium alloys produced more flexible plates compatible with the modulus of elasticity of bone. This flexibility is inductive of fracture healing in areas where higher strain values are needed to promote bone regeneration. Additionally, titanium alloy is reported to be more resistant to cyclic load and notch sensitivity when compared to stainless steel and from a theoretical point of view should be the first-choice material for implants used in this type of breed [50].

Plate removal is indicated if osteopenia or IIO is diagnosed due to the predisposition to refractures after implant removal. This procedure should be staged in two to three surgical procedures [51].

5.8 Minimally invasive plate osteosynthesis

Minimally invasive plate osteosynthesis (MIPO) is a surgical approach to fracture treatment using bone plates, following principles that include (1) the use of indirect, closed reduction techniques; (2) epiperiosteal plate insertion through small incisions remote to the unexposed fracture site; and (3) minimal reliance on secondary implants and bone grafts [52].

This surgical approach emphasizes soft tissue preservation over anatomic reconstruction/absolute mechanical stability and is specially indicated for low-strain fractures. In most fractures repaired by MIPO techniques, the bone heals in conditions of relative stability. Relative stability relies on the use of implants that provide flexible fixation, allowing an acceptable degree of strain compatible (<2%) with bone regeneration. Osteosynthesis methods that are commonly used in MIPO are plates or plate-pin combinations applied in bridging function to span a bone defect not anatomically reduced, resulting in a relatively stable environment.

This technique is applicable in the treatment of most diaphyseal, metaphyseal, and periarticular fractures. The use of an intramedullary pin, particularly recommended in comminuted diaphyseal and metaphyseal fractures, is beneficial in facilitating the reduction and restoration of alignment [53]. The minimum recommended diameter for the IM pin is 30% of the medullary canal diameter at the bone isthmus [54].

MIPO is a surgical approach that often ends up with a long working length plate; however, this is because we have chosen to sacrifice the mechanics of our implant, to preserve the biology. This approach can favor the biological factors of bone regeneration, but the increased working length decreases the stiffness of the construct and therefore the fatigue life of the plate. The primary factors affecting the stiffness of the plate are the modulus of the material used, the AMI of the construct, and the working length. The factors influencing gap strain are gap width and the magnitude of motion between the fragments. Fatigue failure are determined by factors such as the yield bending strength of the construct and the cumulative load/number of cycles that are suffered by the plate. The rationale of the MIPO approach is to improve biological factors at the fracture site to speed up healing, preventing plates from prematurely failing due to fatigue failure.

In MIPO, the plate is applied as a bridging function; for that reason, the selection of an implant of appropriate length is a crucial step. With this surgical approach, it is recommended to use longer plates as possible for improving screw-working leverage and to distribute bending forces well along the plate, thereby lowering pull-out forces on screws. If the surgeon chooses the MIPO approach, selecting the adequate plate length in preoperative planning is a crucial step for bridging osteosynthesis. Two parameters are used to determine the plate length: the plate span ratio and the plate screw density. The plate span ratio is the quotient of plate length and segmental length of fractured/comminuted bone. The plate screw density is the quotient of the number of screws inserted and the number of screw holes available. For comminuted fractures, which are commonly treated with MIPO and bridging osteosynthesis, the plate span ratio should be greater than two to three. For simple fractures, this ratio ranges between eight and ten. In comminuted fractures, the plate working length may not be the distance between the screws closest to the fracture, but rather the unsupported area of the plate corresponding to the length of the fracture gap.

Plate screw density or screw-hole-ratio should be smaller than 0.5–0.4 in comminuted fractures and at least two to three screw holes empty over the bone defect [55]. For simple fractures, a value of 0.4–0.3 is recommended. Because this ratio is usually applied to the whole plate, it may not be as applicable for highly comminuted fractures in shorter animal bones. Also, the screw density can be different in different bone segments due to the diversity of lengths, being higher in shorter segments and lower in longer segments. Mechanically, there was a poor advantage of adding more than 4 screws per fragment. Within a fragment, the guidelines advise placing 1 screw close (near) to the fracture and 1 at the very end of the plate (far) and then a minimum of 2 additional screws evenly spaced over the remaining span. Adding more screws offers no mechanical security but does add surgical damage to the bone [2].

The recommended ratio of plate length to bone length [Plate-Bridging Density (PBD)] should be less or equal to 0.91 ± 0.05 [56].

Beyond location, the number of monocortical and bicortical screws in the construct is also influential on its biomechanical properties. Less torsional stiffness is provided with monocortical screws compared to with bicortical screws. When using LCP, a minimum of one screw must be placed bicortically in each major bone fragment due to a significantly increased torsional stability, based on the scientific evidence of a biomechanical study using bone models [18, 57].

Additionally, long plates enable plate insertion incisions to be created far from the fracture site. Surgical planning should include the exact location and sequencing of insertion of the screws to be placed. It is recommended to start inserting the first screw distally to center the plate in the distal segment. To align the bone and stabilize the fracture, the most proximal screw is next inserted into the proximal fracture segment. Additional screws are inserted and used to reduce the bone to the plate. When using a pre-contoured locking plate, it is recommended that a cortical screw be placed in both the distal and the proximal bone segments to frame the bone to the plate, further aligning the bone in the sagittal plane. After stabilizing the fracture with the 2 non-locking screws, locking screws are sequentially placed in the aforementioned order. Preoperative bone plate contouring is advisable to decrease surgical time with the MIPO technique. Preoperative plate contouring can be performed using contralateral bone radiographs or 3D printing models if the contralateral bone is not fractured [58].

An important factor to be considered is the alignment between the bone axis and the plate. Due to poor visualization of the bone surface caused by a limited surgical approach, malalignment between the bone axis and plate leads to an eccentric plate position can occur. At the proximal or distal end of the plate, a monocortical screw will not anchor in the bone [57]. To overcome this problem of insufficient anchorage of a monocortical self-drilling screw, a long bicortical self-tapping screw can be inserted or a standard screw allowing angulation in the plate hole [57]. However, this procedure can also cause iatrogenic fractures [59].

Dynamic compression plate (DCP), limited contact-dynamic compression plate (LC-DCP), or locking compression plate (LCP) systems have been used with success for MIPO procedures. Nowadays, the MIPO technique is almost performed in the majority of cases using a locking plate-screw interface, such as the LCP, due to the angular stability provided by this system, which by definition increases the load-carrying ability of the construct. The angular stability originates from the threaded screw heads being locked into the threaded plate holes, thus forming a fixed-angle construct. Another important advantage of locking plates for use in MIPO is the minimal contouring needed for the application of the plate in contrast to DCP or LC-DCP, which requires optimal contouring to maintain the reduction of the fracture. Locking plates are considered internal fixators and therefore do not displace the fracture segments during screw tightening regardless of the precision of contouring.

The major disadvantages of using monoaxial locking implants are the inability to vary the angle of screw insertion through the hole (unless using a polyaxial locking plate system) and the increased cost of locking implants compared with that of standard plates and screws [58].

On the other hand, non-locking bone plates for MIPO offer other advantages, in radius and ulna fractures where the plate can be used to reduce and align the fracture segments in the sagittal plane. The relatively flat cranial surface of the radius allows precise reduction of the proximal and distal fracture segments as long as the plate has been preoperatively contoured. Many locking plates also allow the insertion of non-locking (cortical) screws into the plate holes (combi holes). If a locking plate is pre-contoured and initially applied to the bone using a cortical screw in the proximal and distal fracture segment, then the locking plate can be used to align the radius in the sagittal plane similar to a non-locking plate. Once sagittal plane alignment is achieved, the remaining screws inserted should be locking screws, to take advantage of the angular stability provided. The cortical screws that were initially inserted may be left in place or replaced by locking screws [58].

LCPs also have the advantage of preserving periosteal vessels. The periosteal blood supply beneath locking plates is not damaged because compression between the plate and the bone does not occur because is not a plate-bone friction base system which improve and hastens bone healing and simultaneously reduced the risk of cortical bone necrosis and infection. Malunion or delayed union are infrequent complications when using this type of implant in MIPO. Regarding infection rates, when MIPO and ORIF are compared, there is a lack of evidence in veterinary studies, but in the human side, evidence showed lower infection rates when MIPO techniques are used in long bone fractures [29, 60–62].

Further advancements with intraoperative imaging such as fluoroscopy have the following aims: maximized biology due to a more limited surgical approach allows placing implants with a longer working length and improve alignment. Alignment of the main bone segments and the articular surfaces without torsional and angular deformities is also one of the main objectives of MIPO. Intraoperatory fracture alignment can be assessed by two methods: intraoperative diagnostic imaging and clinical evaluation. Intraoperative imaging is not always available in clinical practice, and for that reason, precise perioperative planning is a critical point for bone alignment in MIPO.

Fracture reduction under the plate (FRUP) is a technique that was developed by Cabassu et al. to improve bone alignment on MIPO without intraoperative imaging but requires precise preoperative contouring of the plate and extensive preoperative planning [63].

With the FRUP, the first step of surgical planning is to obtain radiographs of the fractured and contralateral bones, under sedation or general anesthesia. Two orthogonal projections of contralateral bone digital radiographs were obtained using a radiopaque marker (of known dimensions) to calibrate images for plate contouring. The choice of the type of fixation is based on fracture location/classification and biological and clinical factors. After calibrating the radiological image, the craniocaudal or mediolateral image of the long bone is used to contour the plate. Ideally, the plate length is selected to span from the proximal to the distal metaphysis of the bone when possible or based on a plate length/fracture length ratio of 3 (MIPO guidelines) [57]. The placement of the plate on the digital radiograph is oriented by anatomical landmarks that could be externally identified intraoperatively such as patella, medial tibial malleolus, femoral greater trochanter, ulna styloid process, lateral epicondyle, and greater tubercle of the humerus. The plate is then anatomically contoured to adapt to the bone surface (e.g., the lateral face of the femur diaphysis and the medial surface of the tibial diaphysis). Fracture line(s) is drawn on the intact bone, which allows planning the number and the position of the screws to be inserted. First, the site to place the screws near the fracture is chosen. According to the MIPO guidelines, at least three empty screw holes should be respected over the fracture site [57].

When managing long oblique or comminuted fractures that have a significant gap, it is recommended to place one screw in each fragment as close to the fracture as possible. In comminuted fractures with a smaller gap, screws are placed with a minimum of three holes' space between them. In the outermost plate holes, one screw is placed in each proximal and distal fragment. Depending on the case, a third screw may be inserted between the inner and outermost screws [63]. The location for each screw is predetermined and identified by its hole number from proximal to distal. The type of screw, whether locking or cortical, is then selected. At least one cortical screw is used on the distal and proximal fragments to allow for fracture reduction, and these screws are placed first in these bone segments [63]. These screws are inserted in the diaphyseal segment of the bone in diaphyseal fractures or close to the fracture site in metaphyseal fractures. Afterward, the surgeon will then subjectively decide whether to place locking or cortical screws based on the screw location and angulation relative to the joint. The order of screw insertion is then selected, starting with the cortical screws used to reduce the fracture. Generally, the first screw inserted in the femur is in the proximal segment, while on the tibia, it is in the distal segment. The reason is that plate location was easiest to determine on these fragments. The cortical screws that were initially inserted may be left in place or removed and replaced by locking screws. Screw length is measured during preoperative planning as well as screw angulation (this is possible using a variable angle locking plate system) to avoid articular penetration. The plate is then sterilized the day before surgery or during patient preparation and draping. Specially designed "L," "Y," or "T" plates have proven to be very useful for MIPO stabilization of distal diaphyseal or metaphyseal fractures of the long bones (especially in radius fractures), which would normally be difficult to stabilize using straight plates [58].

Two skin incisions are made to the level of the bone surface away from the fracture, and an epiperiosteal tunnel is created, and the plate is slid onto the bone surface [64]. Anatomical references are identified, flowing by the alignment of one bone segment with the plate using bone-holding forceps; immediately after this step, the plate is fixed to the bone fragment using the first cortical screw, which is inserted perpendicular to the bone surface [63]. The opposite bone fragment is then distracted using bone-holding forceps to gain length, and alignment in torsional and axial planes, and temporarily stabilized to the plate to maintain alignment and length. Anatomical landmarks on the opposite fragment relative to the plate are checked, and the second cortical screw is inserted to obtain a reduction under the plate. The second cortical screw is then inserted to obtain a reduction under the plate. Alignment is assessed intraoperatively by evaluating the range of motion and alignment of adjacent joints in axial and frontal planes. When an intramedullary pin is used, the fracture is temporarily aligned under the plate and stabilized using bone forceps only to facilitate the intramedullary pin insertion. The pin is then inserted, and correct insertion is assessed by releasing the distal fragment from the plate. If the placement of the pin is evaluated as incorrect, the pin is removed from the distal fragment, and the fragment is manually mobilized to allow placement of the pin in the distal medullary cavity. The plate is then fixed in the same way as without using an intramedullary pin, and other screws are inserted respecting the order of preoperative planning using additional intermediate incisions if necessary. Other screws are then inserted respecting the order of preoperative planning using intermediate incisions when necessary. Intermediate incisions can be used if necessary to verify the alignment of the caudal tibial cortex and medial/lateral cortices of the radius with the plate. When an intramedullary pin is used, the pin is then cut to the appropriate length. At the end of the surgery, postoperative orthogonal radiographs are obtained respecting the preoperative radiographic protocol for the fractured limb.

In conclusion, it is accepted that:

1. More flexible implants increase strain at the fracture site;

2. An increase in working length creates a more flexible implant;

3. If strain and stress on implants are increased, fatigue life decreases;

4. Strain needs to be at tolerable levels for bone formation, and this tends to be very low.

6. Biomechanics of implant biomaterials

Orthopedic implants are commonly used for different types of surgical procedures to gain optimal function and provide stability to bone tissue. When inserting these implants, the characteristics of the material are important for surgical success, and the ideal implant must be biocompatible and nonallergenic from a biological point of view. However, when contoured an implant to the bone surface, its resistance can change significantly. Implants can be temporary or permanent in the body, and metal possesses properties that make it acceptable for bone repair. In orthopedic implants, metals and their alloys were the first materials used in their production, primarily due to their superior strength and biocompatibility. The metals used for implant production include nickel, iron, cobalt, titanium, vanadium, and aluminum. Metal alloys aim to achieve specific properties in the final mixture, such as ductility, strength, elasticity, and corrosion resistance [65]. Ductility is the ability of a material to absorb energy and plastically deform without fracturing. The term ductility is sometimes used to encompass both types of plasticity: tensile (ductility) and compressive (malleability). Current alloys used in orthopedic metal-based implants include stainless steels, cobalt-based alloys, and titanium-based alloys.

6.1 Stainless steel

Stainless steel 18-8 (18% chromium, 8% nickel) is the most common alloy. It has superior corrosion resistance obtained through compositional modifications by using additional metals, especially Cr [66]. The inclusion of Cr allows Cr_2O_3 promotes the formation of a strong and adherent layer that is beneficial for healing. Stainless steel is commonly used in removable orthopedic devices, such as plates, screws, and intramedullary pins, due to its affordability [50, 67]. Currently, the new stainless steel-based alloys contain Co-Cr, Mn, Ni, and a high nitrogen content. Stainless steel alloys have high resistance to corrosion due to their high chromium content (more than 12 wt%), which enables the formation of a strongly adherent, self-healing, and corrosion-resistant coating of Cr_2O_3 oxide. Different types of stainless steel are available for implant production, and the most widely used is austenitic stainless steel. Austenitic stainless steel, which contains austenite-stabilizing elements such as Ni or Mn, is the most commonly used type of stainless steel for implant manufacture. AISI 316L is the most widely used stainless steel in clinical applications, containing 0.03 wt % C, 17–20 wt% Cr, 12–14 wt% Ni, 2–3 wt% Mo, and minor amounts of nitrogen, manganese, phosphorus, silicon, and sulfur [68].

When compared to bone tissue, stainless steel alloys are significantly stiffer and have proven to be durable enough for osteosynthesis [69]. Additionally, stainless steel is relatively inexpensive and biologically well-tolerated, with a smooth surface from electropolishing. It is also ductile enough to allow for contouring of the plate without breaking [69].

6.2 Titanium and titanium-based alloys

Titanium (Ti) and its alloys were initially used in the field of aeronautics but later gained significant interest in the biomedical field due to their remarkable properties. These properties include a moderate elastic modulus of about 110 GPa, good corrosion resistance, and low density (around 4700 kgm^{-3}) [70].

For orthopedic devices, Ti may be used alone or in alloys with other metals, most commonly commercially pure (CP)-Ti and Ti-6Al-4V alloy; this designation refers to its chemical composition of almost 90% titanium, 6% aluminum, 4% vanadium, 0.25% iron (maximum content), and 0.2% oxygen (maximum content). They both provide stable fixation and a low risk of implant loosening [70].

The report of the osseointegration phenomenon for Ti implants by Branemark [71] led to the development of dental and surgical applications of Ti alloys. This property enables titanium and its alloys to tightly integrate with bone, resulting in the improved long-term behavior of the implanted devices, which in turn reduces the risks of loosening and failure.

CP Ti, grade 4 (ASTM F67) and Ti6Al4V (ASTM F136) are the titanium alloys most commonly used for orthopedic implants. For CP Ti-based implants, four grades are currently available varying their oxygen content. CP Ti grade 4 is the type having the highest amount of oxygen (up to 0.4%) and, consequently, the highest tensile and yield strengths [72].

The use of pure titanium has the following advantages: low weight and very good corrosion resistance, especially in saline solution. Ti and its alloys possess outstanding corrosion resistance, which can be attributed to the creation of a robust and adherent TiO_2 oxide layer on their surface. About the surface properties, namely, wear, the performance is poor due to the low shear resistance of Ti and Ti alloys.

The ability to become tightly integrated into the bone greatly improves the long-term mechanical behavior of the implant as well as reduces the risk of loosening and failure of the device [73–75].

CP Ti, with a single-phase alpha microstructure, is currently used for dental implants production, while Ti6Al4V, with a biphasic alpha-beta microstructure, is mostly used in orthopedic implants and prostheses. The Al and V alloying elements stabilize the alpha-beta microstructure and improve the mechanical properties of CP Ti (typically twice the yield and ultimate strength values of CP Ti). Mechanical properties of CP Ti and their alloys can be altered by heat treatment and mechanical working. Although Ti and Ti alloys are characterized by an array of excellent properties (e.g., favorable mechanical characteristics, corrosion resistance, fatigue-corrosion resistance, low density, and relatively low Young modulus), their processing is complex whether it is by machining, forging, or heat treating.

CP Ti and Ti alloys, on the other hand, more closely matches the modulus of elasticity of bone. This flexibility may be more conducive to fracture healing in points where more strain is required for a bone regeneration to develop. Titanium alloy is also more resistant to cyclic loading and notch sensitivity.

6.3 Cobalt-based alloys

Cobalt-based alloys are superior to stainless steel in terms of strength [76]. However, cobalt alloys have better biocompatibility and are more corrosion-resistant. But these alloys are more expensive to produce. Cobalt-chromium-molybdenum alloy variants are specifically used for hip prosthesis implants due to their high abrasion resistance [77, 78].

6.4 Fatigue failure and cyclic loading of implants

Fatigue failure and cyclic loading are two important concepts for guiding the choice of orthopedic implants to avoid construct failure. Clinically, acute deformation or catastrophic failure by a single applied load is a rare event. Several factors can influence the fatigue failure phenomenon such as the magnitude of the applied load (by consequence generates stress within the implant), the geometry of the implant, the material and how it was handled and manufactured, and the local environment of the fracture.

Experimental determination of the fatigue behavior of a material involves creating an S versus N curve, where S represents the applied stress and N represents the number of cycles required for failure (plotted logarithmically). If the applied stress is greater than the yield stress of the implant, the material fails in a few cycles, such as repeatedly bending a paper clip. The number of cycles to cause failure increases as the applied stress is reduced. The stress level at which a material can withstand an infinite number of cycles without failure is called the endurance limit, which is approximately 50% of the ultimate tensile stress for most metals [2]. A similar process can be used to characterize a fatigue behavior of a structure such as a bone plate, applying a load versus number curve. After determining the yield load, a series of progressively decreasing peak loads are established, and the number of cycles required to reach a defined failure point, such as breakage or reduced stiffness, is recorded. The number of cycles required to reach failure increases as the applied load is reduced. An implant's performance may be considered adequate if it survives a clinically relevant number of cycles, which is often set at 10^6 [2].

The response curve for implant construct may be more complex to interpret because geometry, material, and manufacturing factors may all interact. Factors, such as plate screw holes, may cause local stress concentrations that accelerate fatigue failure. The degree of cold working and even the purity of the production process may vary among different manufacturers of similar implants. Macroscopically visible small imperfections and cracks can trigger the implant failure cascade. Surgeons should also be aware that small notches on the surface of a structure can significantly decrease the endurance limit because it is a stress riser and should prompt intraoperative replacement.

In clinical practice, fatigue failure can be avoided by selecting implants of appropriate strength and dimension for the weight and bone size of the animal, minimizing notching, and, through good client compliance to discharge indications, reducing the magnitude and frequency of the applied loads.

Optimizing the rate at which the one regenerates at the fracture gap consolidates also helps avoid fatigue failure because stress in the plate decreases during the regenerative phase of bone due to progressive load sharing.

The local environmental factors related to fractures also can influence fatigue failure of implants. Factors like load sharing between plate and bone, when anatomical reconstruction of fracture is possible (e.g., simple fracture of long bones), highly reduce the cyclic loading magnitude and early failure of the implants [79]. Technical factors related to the correct application of DCP or LC-DCP plates are also crucial to fatigue failure. An illustrative example is the use of DCP plates without pre-bending or overbending of at line fracture. In this case scenario, there will be compression under the plate and distraction on the opposite cortex, causing failure of load sharing and altering strain distribution over the fracture and increasing the magnitude of cyclic loading and fatigue failure of implant more probable [80]. The correct magnitude of pre-bending of the plate is 2 mm prior to a fixation on a convex side of long bones and provides the most compression at the far cortex and consequently the load sharing between bones and plates [80].

LCP has over the years progressively replaced the use of DCP and LC-DCP plates. One of the main advantages of applying this type of implant is the possibility of applying those without adequately contoured and affixed directly to the bone for stable internal fixation of the fracture. For this reason, it has been used in minimally invasive osteosynthesis modalities such as in MIPO and supports biological osteosynthesis by functioning as an internal fixator, rather than as a full (DCP) or limited contact bone plate (LC-DCP) [18, 81]. Additionally, it was reported that LCPs were more resistant to cyclic loading in different force vectors than DCP and LC-DCP [82]. However, to maintain biomechanical advantages, it is advisable that LCP must not be more than 2 mm away from the surface of the bone [81–83].

Bone regeneration in high-strain fractures occurs only if the interfragmentary strain is less than 2%. According to Claes et al., transverse line osteotomies can tolerate up to 2 mm of micromotion without causing harmful damage to bone regeneration [24]. In this type of fracture, anatomical reconstruction is necessary and the strain at the fracture site caused by different force vectors must be neutralized by the implants during the reparative phase of bone regeneration to avoid complications such as delayed or nonunion [84]. With high strain rates, the magnitude and frequency of loading cycles are also greater because the animal will use the limb very early, and the implants will endure a greater number of loading cycles predisposing to fatigue failure. On the other hand, the reparative phase of bone regeneration develops over time, alleviating the magnitude of load cycles due to load sharing.

In low-strain fractures, the interfragmentary movement is not very harmful to the repair process woven bone can tolerate 2–10% of interfragmentary strain [85]. The main objective in this type of fracture is the indirect reduction of bone fragments with bridge plating or external fixation, aiming to re-establish the mechanical axis and bone length and promote secondary bone healing by relative stability [86]. The great advantage of this method is the possibility of a minimally invasive application, and therefore, it is appropriately used in MIPO, where the preservation of the fracture environment is maximized, and bone healing is optimized and even faster than in open reduction and internal fixation (ORIF) [13, 56, 61, 87]. On the other side, from a mechanical standpoint, plates experienced a greater magnitude of strain, increasing the risk of fatigue failure. However, there are surgical options for a sparing effect on

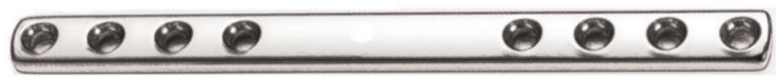

Figure 13.
Image of biological healing plate with a central section without screw holes.

plate site that bridges fracture site and reduce strain. The use of an intramedullary rod (IMR) also helps the restoration of alignment, a substantial challenge in MIPO inherent to the lack of fragment observation and biological healing plates [53, 88]. These former implants are designed to support high strain values for bending and torsional force vectors by possessing a central section without screw holes (**Figure 13**). The screw holes located at the plate's outer section allow the implant to be fixed to the intact proximal and distal fragments, which avoids the need for anatomical reduction of the diaphysis. Additionally, the use of a locking version of this type of plate can improve the performance of the implant by decreasing the pull-out of screws and the need to exactly contour the plate to the bone surface. By applying a MIPO approach, the soft tissue disruption can be minimized, improving biological factors at comminuted fracture sites and hastening bone regeneration.

7. Conclusion

Mastering the concepts of biomechanics in fracture management is an essential tool for the small animal orthopedic surgeon. The application of these concepts in the selection of implants, surgical technique, and fracture healing and their interaction can reduce the rate of postoperative complications. With the rise of minimally invasive osteosynthesis, the knowledge of the most common fracture pattern and the interaction and how the force vectors act on fracture sites determines the choice of implants. On the side of the implant, the knowledge of AMI and the working length of implants determines the yield of the construct and the ability to support the forces before implant failure occurs. Gap strain management is vital for vascular ingrowth and tissue differentiation along the osteogenic pathway. The recognition of the strain pattern at fracture (low strain fracture vs. high strain fracture) is a key element to implant choice and by the influence of the magnitude of the strain at the tissue differentiation (during the osteogenic pathway) also influences fracture healing. Strict adherence to guidelines for implant placement is another pathway to fulfilling evidence-based biomechanics in orthopedic surgery. Finally, an important part of the postoperative assessment of constructs is for surgeons to use their understanding of these mechanical parameters to predict the weakest point and have this guide patient management decision.

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Conflicts of interest

The authors declare no conflicts of interest regarding the publication of this article.

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