

Physics behind the implants used for high strain fractures: literature review

A física por trás dos implantes utilizados para fraturas de alto strain: revisão de literatura

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Abstract

Whenever bone fractures occur, external forces produce continuous interfragmentary motion and a stabilization method is necessary. It is known that the mechanical conditions at the fracture site influence bone callus formation during the healing process. To achieve primary (direct) bone healing, absolute stability at the fracture site is necessary. Perren's concept of strain determines that relative deformation at the fracture gap depends on the original gap's conformation. Simple fractures (without comminution) are considered high strain fractures since a small force applied to the fracture site results in great movement with deleterious effects on the healing process. The purpose of this study is to review the available literature regarding factors that influence the mechanics of high strain fractures in veterinary medicine, its treatment methods and implants available. Each fracture configuration requires special attention and critical care in choosing the osteosynthesis method and the type of stability required for consolidation to occur within the expected time. One must know the strain theory to become an orthopedic surgeon.

Keywords: Fracture fixation. Physical stimulation. Veterinary orthopedics.

Resumo

Quando fraturas ósseas ocorrem, forças externas produzem movimentação interfragmentária contínua, e um método de estabilização se faz necessário. É sabido que as condições mecânicas no local de fratura influenciam a formação de calo durante o processo de cicatrização óssea. Para que cicatrização óssea primária seja obtida, é necessário estabilidade absoluta no foco de fratura. O conceito de *strain* de Perren determina que a deformação relativa ao foco de fratura estabilizado depende do tamanho da lacuna de fratura original. Fraturas redutíveis (sem cominuição) são consideradas de alto *strain*, pois uma pequena força aplicada à linha de fratura resulta em grande movimentação com efeito deletério ao processo de consolidação. Este trabalho revisa a literatura disponível a respeito de fatores que influenciam a mecânica de fraturas de alto *strain* em medicina veterinária, seus métodos de tratamento e a física por trás dos implantes disponíveis. Cada configuração de fratura requer atenção especial e cuidado crítico na escolha do método de osteossíntese e no tipo de estabilidade necessária para que a consolidação ocorra no tempo esperado. Conhecimento da teoria do *strain* é mandatório para a formação de cirurgias ortopédicas.

Palavras-chave: Fixação de fratura. Estimulação física. Ortopedia veterinária.

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Literature review

Fracture is an acute disruption of the continuity of bone tissue, where force transmission through this tissue is no longer feasible. External forces act continuously in long bones fractures producing interfragmentary motion and a stabilization method is needed (AUGAT et al., 2005). There are two types of bone healing, defined according to the level of stability of bone fragments during the healing

process: primary or direct healing (fragments are rigidly fixed with no movement at the fracture site), and secondary or indirect healing (when there is relative motion between bone fragments, with callus formation) (BETTS; MÜLLER, 2014). Several studies have shown that mechanical stresses at the fracture site influence callus formation during the healing process, particularly regarding stability of the implants used for osteosynthesis (JAGODZINSKI; KRETTEK, 2007). Mechanical stimuli modulate the healing period, change the proportions of types of generated tissues in the fracture area and regulate gene expression patterns of bone repair cells (PALOMARES et al., 2009).

The concept of strain on fracture gaps and its influence on bone healing was introduced in the late 1970s (PERREN, 1979). Strain is the relative deformation at the fracture gap divided by the original gap's width. Tissue cannot exist under strain conditions that exceed its tolerance to motion and elongation at rupture (HAK et al., 2010). Tissues are only built on environments with ideal mechanical conditions. Cortical bone accepts only 2% elongation before rupture and requires strong and rigid fixation with compression between the fragments to heal directly without callus formation (RAHN, 2002). Trabecular bone, on the other hand, tolerates up to 10% strain, and whenever relative stability is achieved, secondary healing occurs. If more than 10% deformation is still present, non-union might occur (HAK et al., 2010). Strain of the fracture gap depends upon the type of fracture. Reducible fractures (without gross comminution) are considered high strain fractures since small forces applied to the fracture area result in large relative deformation with deleterious effect on the healing process (AUGAT et al., 2003). The purpose of this study is to review the available literature regarding factors that influence the mechanics of high strain fractures, methods of treatment and the physics behind the available implants.

Bone biomechanics

Bone has measurable characteristics such as resistance, hardness, moderate elasticity, and limited plasticity. The stiffness of bone corresponds to approximately 10% that of steel (SCHATZKER, 2002). It attenuates sonic and ultrasonic waves, dissipates energy and is ideal for standing and moving. Cortical bone exhibits viscoelastic behavior, modifying its mechanical properties and deforming as a response to the loads the tissue is submitted to *in vivo* (GIBSON et al., 2008). It has the function of load transmission and is subjected to constant compression, deflection, torsion and shear loads besides avulsion loads exerted by tendon insertions (RAHN, 2002).

Osseous tissue regenerates and repairs itself, thus altering its morphology and those mechanical characteristics. It also exhibits piezoelectric properties (AN et al., 2000). Bone is an anisotropic material; therefore, it responds to loading according to the direction on which the load is applied. Bone remodeling occurs in consonance to Wolff's law – osteoblasts lay bone down where needed and osteoclasts resorb where it is not required, responding to mechanical stresses placed on the osseous tissue (SCHATZKER, 2002). Whenever forces are applied to any object, it suffers distortion from its original dimensions and internal forces are produced within this body (CARTER; SPENGLER, 2002).

Stress is the load per unit area and strain represents the fractional change in dimension (deformation) of a loaded body (SHARIR et al., 2008). It is possible to correlate load suffered and deformation caused by the exponential response curve generated named stress/strain curve. This curve is affected by the material's properties, geometry and structural morphology (RAHN, 2002). Strain constitute the geometric change occurred in response to determined load (deformation per unit time) (AN et al., 2000).

Constant applied stress and its gradual augmentation exhibits, at first, a linear relation with strain; the material deforms elastically proportional to the tensile force and returns to its original form when the stress is removed. This linear correlation of elasticity is known as Hooke's law (SHARIR et al., 2008).

At a second instant, the curve is no longer linear, and the strain lengthening becomes larger for the same load applied (SHARIR et al., 2008). Once a specific point (yield point) is achieved, any further load increment results in a non-linear and irreversible (permanent) plastic deformation. If loading is continued, the mechanical strength limit is crossed, leading to bone failure (AUTEFAGE, 2000). Bone failure relates more to loading rate than to load itself. The faster the load is applied, the more energy is accumulated, and explosive failure occurs, and if this force is not dissipated within the fracture itself, it is absorbed by the muscle envelope (SCHATZKER, 2002). The ratio between stress applied (σ) and correspondent strain (ϵ) is the coefficient of proportionality represented by what is named Young's modulus (E) or modulus of elasticity, characterized by the formula:

$$\sigma = E \cdot \epsilon$$

This modulus shows the material's stiffness and is represented by the slope of the linear portion of the curve generated by

normal loading (AUTEFAGE, 2000; SHARIR et al., 2008). Stress produces loads of axial compression, torsion and bending. Both rate and loading type determine fracture configuration. Bone shows resistance to compression and usually fails by shear forces (RAHN, 2002).

Pauwels (1960) first proposed that tissue differentiation within bone callus was determined by mechanical stimuli. His theory was that cartilage was formed because of local hydrostatic pressure causing mesenchymal stem cells to become chondroblasts, while bone and fibrous tissue resulted from shear deformations which would cause mesenchymal stem cells to differentiate into osteoblasts and fibroblasts, respectively. Mechanical loads applied to bone propagate at a level which cells are affected, resulting in alterations over the tissue. Many studies investigate the effects of loads applied to bones *in vivo* and quantify tissue response and their mechanical skills (BETTS; MÜLLER, 2014).

Perren and Cordey (1980) studied the maximum mechanical stimulation which bone tissue may be submitted to while still being able to heal, analyzing the gradual changes in the mechanical properties of repairing tissues and its replacement for more resistant cell types (RAHN, 2002). In the initial phase of wound healing, particularly when soft tissue is present, a fracture tolerates higher strain than a later stage when the callus mainly contains calcified tissue. The manner in which mechanical factors influence fracture healing is explained by Perren's strain theory (PERREN, 1979; BETTS; MÜLLER, 2014).

The strain theory

Strain (ϵ) is a material's relative deformation (for example, granulation tissue within a fracture site) when a given force is applied. Normal strain is the change in length (ΔL) in comparison with the original length (L) when a certain load is applied. Thus, it has no dimensions and is often expressed as a percentage, as in the following formula (PERREN, 1979; PERREN, 2002):

This theory suggests that the strain which causes healthy bone tissue to fail is the upper limit of tolerable deformation for this tissue's regeneration (JAGODZINSKI; KRETTEK, 2007). The amount of deformation that animal tissues can withstand while

$$\epsilon = \frac{\Delta L}{L} \times 100$$

maintaining its function varies widely. Intact bone can withstand 2% strain (before fracturing), while granulation tissue has a plasticity of 100%. Thus, a fibrous callus will not stabilize a fracture gap when there is too much movement between the fracture ends (PERREN; CORDEY, 1980). With the expansion of callus volume, a decrease in the local tissue tension occurs to a level that allows bone bridging (RAHN, 2002). This adjustment mechanism is not effective when the gap fracture is considerably reduced so that most of interfragmentary motion occurs within the gap itself, yielding a high strain environment. In this case, fracture overload with too much motion between the fragments, in a more advanced stage of bone healing, is not tolerated by the bone cells (CLAES et al., 1998).

In long-bones fractures, normal forces that act on the living animal cause fracture displacement, resulting in interfragmentary motion (RAHN, 2002). To achieve stability while keeping the fracture fragments in place, rigid and strong implants should be used. Furthermore, the choice of the fixation method must take into consideration many patient factors such as general health, age, weight, presence of concomitant lesions, expected level of physical activity (according to the patient's labor routine), owner's ability to take appropriate post-operative care (HOULTON; DUNNING, 2005).

From the mechanical point of view, the fracture should be assessed in order to determine its acting forces, configuration and the advantage to perform biological fixation (HAK et al., 2010). Fracture pattern is decisive for choosing the fixation method. Simple fractures (reconstructable), whenever possible, should be treated by direct reduction with fragment compression, especially articular fractures. Absolute stability of the fracture site allows primary (direct) healing (HOULTON; DUNNING, 2005). Without movement, there is no development of bone callus and healing occurs through Haversian remodeling of cortical bone (BETTS; MÜLLER, 2014).

This is a unique case in the spectrum of bone healing. Contact between bone fragments is required to maintain the stability and cause friction counteracting torsional and shear forces, while the transmission of compression is shared with the implant (PERREN, 2002; RAHN, 2002). In a small fracture gap the ratio between the change of width to the total width is high, meaning that the total deformation (strain) is high and should not exceed 2% to achieve direct bone healing (RAHN, 2002).

In direct bone healing process, at first granulation tissue is formed within the small fracture gap through angiogenesis. Then, loose connective tissue is formed at the gap's edges while in the center there is deposition of lamellar bone until the entire gap is filled, which occurs if stable mechanical conditions are maintained (CLAES et al., 1998; RAHN, 2002). Complete remodeling takes a few months up to years, depending on the animal's species and health conditions (PERREN, 2002; RAHN, 2002).

Bone implants that promote absolute stability

Different implants can be used to reduce high strain fractures. Lag screws, cerclages wires, tension bands, compression and neutralization plates are the recommended options (SCHATZKER, 2002).

A screw is a very effective implant for fracture reduction by means of interfragmentary compression (KOCH, 2005). Screw purchase in the bone depends on the bone-implant interface. The goal is to achieve the largest possible contact area in a sufficiently stable implant with the smallest possible size. In veterinary medicine, usually cortical and cancellous screws are used. Cancellous screws have larger outer diameter, deeper thread and a larger pitch between the threads and are used in the metaphyseal and epiphyseal bone. Cortical screws are designed for bone shaft (PERREN, 2002; KOCH, 2005).

The simplest way to compress two bone fragments is through the use of a lag screw, but it often requires auxiliary fixing methods because it does not provide sufficient strength alone to withstand physiological loads (SCHATZKER, 2002). Lag screws convert torque into compressive forces. They must be positioned perpendicular to the fracture to prevent fragment slippage when a force is applied (KOCH, 2005).

Lag screws are partially threaded to engage only the trans cortex. However, the lag effect can be achieved by use of a regular cortical screw (threaded throughout its length). This is done by drilling a gliding hole of a diameter equal to, or moderately larger than, the outer diameter of the screw thread in the near cortex. An opening in the trans cortex is drilled, corresponding to the core diameter of the screw, the full length of the drill hole is measured and the drill hole in the far cortex is tapped. Thus, the fully threaded screw is applied with a smaller threaded hole in the far cortex and slides within the near cortex, achieving compression between the fragments. The lag

screw must be inserted in the middle of the fragment to prevent cracks and to achieve maximal interfragmentary compression (SCHATZKER, 2002; KOCH, 2005). There are cannulated screws with a central hollow core that are inserted over Kirschner wires that act as a guide for use in cancellous bone. They can be also employed as lag screws and are indicated for the reconstruction of the epiphyseal or metaphyseal fractures such as those happening in the distal humerus, or proximal femur (KOCH, 2005).

It is known that the strength of bone tissue reduces as the size of the screw increases; therefore, it is recommended that the screw diameter does not exceed 40% of the diameter of the bone that it is applied to (KOCH, 2005; BOUDREAU et al., 2013). Screws can be self-tapping or not, according to the manufacturer's needs. Self-tapping screws are designed in such a way that once a pilot hole has been drilled into bone, they can be inserted by plainly screwing them in. Indications for lag screws include avulsions, epiphyseal or metaphyseal fractures, butterfly fragments reduction and intraarticular fractures (SCHATZKER, 2002).

Cerclage wires are a malleable form of 316L stainless steel. They come in a range of diameters, from 0.5 to 1.5 mm. The larger the wire's diameter, the higher its yield bending and tensile strength, but also the more difficult it is to manipulate the wire. It is up to the surgeon to decide the appropriate size for each situation (ROE, 1997). The wire must be in intimate contact with the bone surface. The periosteum should be elevated. If there is soft tissue between a wire under tension and the bone, it will surely become necrotic and be resorbed, reducing the effective bone diameter. Even with very tight wires, slight reductions in diameter (< 1% of the bone diameter) will cause the wire to loosen and potentially become ineffective (ROE, 1997).

A study compared the use of two lag screws alone, two lag screws associated with double cerclage and double cerclages without screws in torque and axial load in plastic bone models with standard midshaft butterfly fractures. The authors concluded that the use of double cerclage is more stable than lag screws in torque, but there was no statistical difference between lag screws and double cerclage under axial load (KANAKIS; CORDEY, 1991).

Another treatment option is dynamic compression plates (DCP®), first launched in the market in 1969. They allow the creation of axial compressive force by eccentrically positioning screws on the plate

(PERREN et al., 1969; KOCH, 2005). When the compressive screw is tightened, the bone is displaced relative to the plate, causing compression at the fracture site. It is possible to use one or two compressive screws, one on each side of the fracture gap. There are two drill guides for each DCP® plate size, a gold drill guide that produces an off-center hole 1.0 mm away from the fracture, and another green drill guide for neutral (concentric) holes (KOCH, 2005).

In order to achieve interfragmentary compression, the plate must be stabilized in one of the fragments with a screw in the neutral function; then, in the other fragment, a screw should be inserted in compressive function to slide the fragment underneath the plate and cause interfragmentary compression (PERREN, 2002; KOCH 2005). At first, both cortices are equally compressed, but with increased pressure bone tends to bend slightly towards the plate. Straight plates need to be pre-contoured prior to application, to keep intimate contact with the bone surface (SCHATZKER, 2002; FERRIGNO et al., 2016).

It is recommended to use the longest plate possible to the bone in question, as a long plate is more efficient than a shorter plate in neutralization. It increases the working length of the implant and distributes the forces over a larger surface. The ideal distance between the nearest screws to the fracture line must be 4 to 5 mm or to the exact size of the diameter of the screws used for that plate (KOCH, 2005; PIERMATTEI et al., 2006).

The DCP® plates were further developed by the limited contact dynamic compression plates (LC-DCP). They were created following the trend to reduce the area of the plate–bone contact (the plate “footprint”) sparing the capillary network of the periosteum under the plate. They are available in both stainless steel and titanium (KOCH, 2005).

The LC-DCP’s holes are symmetrical, allowing eccentric screw placement in either direction. Additionally, the plate holes are evenly distributed over the entire length of the plate and the screws can be inclined in any direction to a maximum of 7° and in the longitudinal direction up to 40° (as opposed to the maximum of 25° for DCP® plates, an obvious advantage). This feature allows for compression at any level along the plate (KOCH, 2005). The geometry of the plate results in an even distribution of stiffness, making contouring easier and minimizing stress concentration at any of the screw holes (SCHATZKER, 2002).

The locking compression plates (LCP®) that have a “combined” plate hole that accommodate either a conventional screw or the new locking head screw

(LHS), which has a conical threaded head that locks on the plate. They are also designed to reduce the plate/bone contact, minimizing bone necrosis (PIERMATTEI et al., 2006). The new combination hole comes in two parts: the first one has the design of the standard DCP/LC-DCP compression hole and accepts a conventional screw (that can perform axial compression or allow placement of an angled lag screw). The other part is conical and threaded to fit the locking head screw (thus providing angular stability) (PERREN, 2002; KOCH, 2005). In order to achieve dynamic compression on such plates one has to use cortical screws in eccentric position (SCHATZKER, 2002; HAMMEL et al., 2006).

The use of LCP for high-strain fractures can compromise stability through micromovement on the trans cortex (HAK et al., 2010). The axial stiffness of the LCP is mainly influenced by the plate’s working length (STOFFEL et al., 2003). The working length of a plate is defined by the distance between the two screws placed closest to the fracture line. One disadvantage of placing screws near the fracture gap is the potential to concentrate stress and occur implant failure (plate breakage). The smaller the working length, the greater the stiffness (HAK et al., 2010).

Placing screws farthest from the fracture site can better distribute tension along the plate length and reduce the risk of failure. However, this maneuver can allow for greater relative deformation in the fracture gap interfering with consolidation, particularly for high strain fractures. Screws should be positioned closer to the fracture gap for increase in axial stiffness (HAK et al., 2010).

Long bones such as the femur, humerus and radius are eccentric loaded (SCHATZKER, 2002) and plates applied to those bones may also act as tension bands, converting eccentric load into compressive forces if they are applied to the convex side of a curved bone (PIERMATTEI et al., 2006). Pre-bending the plate helps to enhance force conversion and achieve further compression between the fracture fragments.

Plates may also be used in neutral position, when a primary tutor is positioned performing interfragmentary compression (such as a lag screw or cerclage). In this case, the plate protects the interfragmentary compression achieved by the other implant from all rotational, bending, and shearing forces and carries about 70% of the load applied to the bone, reducing the stress on the fracture site (HULSE et al., 2005).

Plates and screws can damage the bone fragments' vascularization during their positioning and application. In part of the canals, resorption around the damaged vessels occurs, but it does not include the full size of the osteon (HORSTMAN et al., 2004). A new osteon originates in the filled gap as in a process of internal renewal (RAHN, 2002). A reduction of bone mass in the area under the plate occurs when there is excessive stability, a phenomenon called "stress protection" (PERREN, 2002; RAHN, 2002).

Another type of implant used to achieve interfragmentary compression is tension bands. They convert tensile force into compression force. Generally, they are suitable for fractures/avulsions of patella, olecranon, greater trochanter and greater tuberosity. The first fragment is stabilized using two or more K-wires, or a lag screw (KOCH, 2005). Then, one cerclage wire is placed to counter rotational forces. To take advantage of the resultant force, pins should be oriented perpendicular to the fracture plane (STIFFLER, 2004; KOCH, 2005).

In general, all the techniques applied to high-strain fractures allow load sharing between the implants and the bone, reducing the risk of failure.

Some of the implants available in Brazil are made in similar fashion to those popular in North America, but there is still lack of controlled biomechanical studies as well as laws and regulations for the development of proper national implants (AZEVEDO; JÚNIOR, 2002).

Effect of mechanical stimuli on high-strain fracture healing

Numerous animal models and mechanical stimulators have been used to test the effects of interfragmentary movement in fracture healing (GOLDZAK et al., 2014). The latest studies show that mechanical stimulation

improves callus formation for fracture healing in its early stage (BOERCKEL et al., 2009).

In two-dimensional and three-dimensional cell culture systems, undifferentiated and differentiated cells are used to study the *in vitro* response to mechanical loads (JAGODZINSKI; KRETTEK, 2007). Differentiated cells can be stimulated to proliferate and synthesize specific components of the cellular matrix as fibroblasts (ZEICHEN et al., 2000) and chondrocytes (MAEDA et al., 2001). Osteocytes, in turn, have different proliferation and differentiation responses (JIN et al., 2001).

The strain theory predicts that fracture healing only occurs if the interfragmentary strain (interfragmentary motion divided by gap width) is less than 2% (PERREN, 1979). Claes et al. (1998) concluded that transverse line osteotomies tolerate about 2 mm of micromotion without deleterious damage to bone healing.

Conclusion

Each type of fracture requires special attention regarding its critical evaluation and care in choosing the osteosynthesis method and type of stability necessary for consolidation to occur at the expected time and to foster rapid limb function restoration. Knowledge of the strain theory and its applications in different species is mandatory for surgeons who aim to surgically treat orthopedic conditions. In high strain fractures, absolute stability at the fracture gap (up to 2% strain) by means of interfragmentary compression is a mandatory factor to achieve primary bone healing and should be the standard treatment of high strain and intra-articular fractures, always respecting biological aspects to cause the least possible damage to the local vascularization.

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