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Mechanics of Musculoskeletal Repair Devices

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Abstract

This paper applies the mechanics of engineering science and materials to the understanding of clinical devices used in Orthopaedics and Trauma. The rigidity of devices is described to be a function of material stiffness and its geometry relative to the loading axes. Structures are more rigid under loads that are applied along their long axes and are more flexible under bending and torsion which increases with length. This may be applied to an individual plate, screw or bone and to the entire construct. Increasing the thickness of a plate greatly increases rigidity as a third power relationship exists between these variables. Similarly, increasing the diameter of a rod increases its rigidity by a fourth power relationship. A hollow cylindrical cross-section, as found in long bones, provides the most effective rigidity to weight ratio when complex stresses are applied. The paper provides examples to reinforce basic structural mechanics applied to medical devices.

Keywords

Mechanics, Materials, Geometry, Structural Support, Stress, Force, Moment, Trauma, Orthopaedics.

Conclusion

To promote bone healing medical devices are intended to provide a specific level of strain or deformation to the fracture site. An understanding the fundamentals of stress is important in the pre-operative planning /selection of these devices to complement the specific support that is intended.

Introduction

Clinical Trauma and Orthopaedics involves the use of many implants that either replace the function of or provide temporary or permanent support to bones to facilitate healing. The general aim being to remove pain and restore mobility of the patient. Many of these implants were originally produced empirically based on basic understanding of materials science, mechanics and bone healing, evolving into an effective device with less successful designs being abandoned. Increasing understanding of bone healing and biomechanics alongside advances in design technology have led to increasingly specific designs that address particular problems from conception. The aim of this paper is to provide the reader with an understanding of the function of structural support devices used in orthopaedic surgery. An introduction to the basic science of materials is firstly considered followed by some practical examples.

Introduction to Basic Science

The rigidity of a structure defines its deformation under loading and therefore its ability to provide support. This is a function of the materials used and the shape of the device, with the shape being surprisingly just as important as the material. In Engineering terms, calculation of a structure's rigidity will also change with the direction and type of load applied as shown in Figure 1, which shows an example for hollow tubes.

Figure 1

From Figure 1, axial loads generally result in small deformations, whereas tangential loads cause bending moments that produce deformations that will be orders of magnitude greater and potentially far more important. Torsional loads produce rotational deformation that increase with length. Additionally, the deformation of bones, particularly from bending forces, is restricted in the body by the action of muscles that attempt to keep bones under axial compression only where bone is the strongest. This can create extremely complex loading regimes, and clinically it would be rare to experience a single form of loading as anatomical loading is multidirectional. It is therefore critical to understand the type and magnitude of forces that an implant will be expected to withstand when implanted into the body. For these reasons simple axial load models or tests are seldom appropriate. The important contributors to the mechanical performance of an orthopaedic implant are therefore the stiffness of the material used, its shape and the type of load applied.

Under bending forces, as shown in the equations above, stiffness is calculated as the product E I.(1) Where E is the elastic modulus of the material, a material constant, and I is the moment of inertia of area, a variable that is determined from the shape of the object. The elastic modulus varies from 1-10 GPa in polymers and bone, to 100-200 GPa in alloys and steel, hence metallic engineering materials used in implants or trauma devices are 1-2 orders of magnitude stiffer than bone. Additionally, the Elastic modulus of medical Titanium alloys (E~ 110GPa) is ½ that of Stainless Steel or Cobalt Chromium (E~200 GPa). Implant materials should therefore be selected with their mechanical properties in mind, based upon the forces they will be expected to encounter, alongside their geometry as described below. It will become clear that a Titanium plate for example could be produced with similar mechanical properties to a stainless steel plate by simply altering its thickness.

For purely axial loads (Figure 1), the shape of the object in cross section is not important, as deformation is a function of the cross sectional area alone. However, for loads applied in bending or torsion, shape is much more important and there is much greater scope to alter mechanical performance by changing geometry than by selecting different materials. The moment of inertia of area (I) describes the distribution of material in cross section around a theoretical neutral axis (Figure 2). Bending of an object occurs about this axis, thus when stress is applied to objects with a solid symmetrical cross-section bending occurs about the centre line with tension occurring on the side that is stretched and compression on the other side that becomes shorter. The neutral axis describes the point in the structure that is effectively subjected to zero net force – thus material on the bending axis has very little stress applied to it. The presence of a neutral axis means that hollow materials can be more effective at supporting loads than solids, as the material can be concentrated where it is needed. The further the material within the object is located away from the neutral axis generally the greater the stiffness. This is evident in building construction in steel beams with an "I" cross section that have top and bottom plates separated by a thin vertical rib; these beams push material away from the neutral axis to increase the moment of inertial of area. If the direction of bending is known, like in a civil engineering supporting beam, the material within the beam can be specifically positioned, however, in the case of a long bone where forces can act in many planes, often at the same time, the most effective cross section becomes a circular annulus.

Figure 2

Different mathematical formulas describe I for objects with different cross sectional shapes, under different types and direction of loading. This is important because, as described above, the material further away from the neutral axis in the plane of deformation will have more effect on rigidity. The equivalent property under torsional loading is known as the polar moment of inertia of area (J). Examples for common cross sections encountered in orthopaedic implants are shown in Figure 3. For objects with complex cross-sectional shapes, the overall moment of inertia of area can be calculated by adding or subtracting the moment of inertia of the simple shapes for the material that is or is not present respectively. A good example is that is of a hollow tube where I or J of the overall cross section is equal to that of the outer diameter circle minus the inner circles diameter (Figure 3). If the bending axes are not coincident in the shapes, it is necessary to use the parallel axis theorem; this is beyond the scope of this paper.(1)

Figure 3

The body takes great advantage of material distribution in long bones such as the femur by creating an annular hollow structure under the control of Wolff’s law.(2) The value of I (bending) and J (torsion) are a function of the radius to the power of 4, hence for any given cross-sectional area I,J are greatest if the internal material is removed and placed more peripherally (further from the neutral axis). For the case of a solid shaft with the same cross-sectional area as the example in Figure 1 (D_{solid}=19.9 mm) the magnitude of I ,J would decrease by a factor of 5 (I_{anular}/I_{solid} =5). Thus, when using the same amount of material, a hollow tube will be more rigid under bending or torsion than a solid rod as the material is placed away from the neutral axis where it has more effect. A long bone is therefore adapted to its function as a tube will be lighter than a solid bone with the same diameter. Similarly increasing the diameter of a rod by a small amount greatly increases its bending rigidity. For example, a 9mm intramedullary nail would be approximately 50% more rigid than an 8mm nail. This effect is seen in ageing osteoporotic bones where the amount of osseous material available is decreased. Again under the effect of Wollf's law the medullary canal of the bone expands leaving the outer diameter large, creating the most rigid, strong structure possible with the given material.

Typical values of bending rigidity for common structural shapes are shown in Table 1; note the cross sectional area (proportional to the mass assuming length is maintained) is constant for all of the objects shown and only the shape has been altered. This is the equivalent of forming different objects using the same amount of material, of the same length, but with different shape. For comparison to the annular cross-section shown in Figure 1, the moment of inertia of area of a rectangular section (as in a plate) is equal to $I_{plate} = \frac{(wh^3)}{12}$, where w is the width and h is the thickness. Note that, as the moment of inertia of area is proportional to the thickness of the plate to the power three, small increases in plate thickness result in large increases in bending rigidity. Increasing the plate thickness from 2 to 3mm would lead to the bending rigidity increasing approximately 3 fold.

Table 1

As the cross-sectional area (mass) of all of the shapes in Table 1 is identical, under an axial load all of the beams would deform the same amount. However, under a bending load their stiffness and thus their ability to resist deformation varies by up to 16 times. These examples also demonstrate that altering geometry has a much more important effect than altering material. For example, the plate on its edge (ii) is 9.6 times more rigid than the same plate on its side (i), whereas changing material would only change the rigidity by 2. The most rigid object in Table 1 is the annular cross-section, it is also the thinnest, demonstrating that the distribution of material in a particular shape is most important. The annular tube (v) has a stiffness 5.5 times greater than a solid rod (iv). This is adopted in modern bicycles that utilise very thin aluminium or carbon tubing that have a very large diameter compared to the smaller diameter, thicker steel tubing used historically.

The structure of the body's long bones varies significantly. The reasons for this can be determined by considering the forces encountered under physiologic loading and, again, considering Wolff's law. In the metaphysis, adjacent to the joint, limited bending forces are encountered; friction is low, impact forces can be high and muscle action will tend to move the joint rather than apply a bending force to the bone. Hence, the metaphyseal bone is porous and cancellous. This bone can absorb proximal impacts and transmit the resulting stress along the length to the cortex. The trabeculae are aligned with the direction of load transfer and supported additionally by cross braces much like a step ladder, to prevent the slender trabeculae from buckling. Bending and torsion increases along the length of the bone, therefore, the structure of bone changes to become a dense cortical annulus. Hence, the bone changes naturally to support the load applied to it. Designers can copy this by preferentially reinforcing materials to increase their moment of inertia of area in areas of high stress by varying their thickness and orientation to the bending axis. It must however be remembered that if constructs are excessively rigid they will shield the bone from forces. This can result in poor fracture healing in the short term and resorption of bone with resultant failure in the long term due to localised osteoporosis and bone weakening.(3) This is commonly seen following total hip replacement as the proximal femur is shielded from the complex stresses applied to it by the essentially rigid femoral stem leading to bone loss proximally and bone growth distally.

Clinical Examples;

Internal fixation plates

Internal fixation plates and rods are fastened in place with bone screws of varying type, some providing rigid fixation and others providing a degree of controlled movement. The necessity for screws requires the plate/rod to have a range of access holes to accommodate anatomical variation. From an engineering perspective there are general common sense design rules that are useful to consider. Holes, in particular, should be avoided in any construct as they reduce the cross sectional area locally and therefore act as a point of weakness. If required they should be placed where possible more than 1 diameter from an edge of the plate and more than 1 diameter from the next adjacent screw hole. This is because features such as holes, notches, grooves, threads, or any sudden changes in geometry act to provide stress concentrations that accentuate the localised stress to be up to an order of magnitude greater than if the feature was not present.(1) Thus an important aspect to consider in design is to keep stress concentrations away from the location of maximum bending moment where stresses are at their greatest.

In internal fixation devices, the absolute application of engineering principles are sometimes disregarded due to practical constraints, particularly for trauma applications. The constructs cannot ordinarily be manufactured for each individual situation and therefore must be versatile enough to be used in different injury patterns in different individuals. A good example of this is the fact that many plates will have holes for screws throughout their length. It may therefore be impossible to avoid having features (such as holes) at the location of maximum bending moment. Some more modern designs try to take this into account, examples are shown in Figure 4 A and B. The operating surgeon must otherwise take account of any short-comings in implant design when faced with a particular situation. This will allow synthesis of constructs which will mitigate these issues, examples are shown in Figure 4 C and D where addition fixation has been added to neutralise particular forces.

Figure 4

Where fractures are slow to unite and / or such issues are not taken into account, implant failure can occur. The redundant screw holes in Figure 5 are a prime example of this. Due to their proximity to the fracture site, these holes cannot be used, they therefore represent a point of weakness in the area of maximum bending stress opposite the unstable fracture site, contributing to the ultimate failure of the device. Had the plate not had holes in this particular location it would likely not have failed.

Considering the plate construct in Figure 5, as screws on an individual side of the fracture site get closer together, the force applied to them increases as jointly they form a force couple. A force couple is defined as two points of load that together produce a bending moment. The two forces within a force couple have equal magnitude but opposite direction, hence, one is in compression and one in tension.(1) The overall resistance to bending is the magnitude of one force multiplied by the distance between the screws. The bending moment is a function of the load applied to the limb, thus as the screws get closer together on a specific side the force on each screw increases linearly. Furthermore, if the applied moment changes direction, the force in the screws swaps from tension to compression. It is important to note that screw holes provide stress concentrations to the bone and therefore, screws should not be placed too close to the edge of a bone/fracture. It is also crucial that there are at least 2 screws on each side of the fracture site as an individual screw, whilst being able to resist shear, will not resist a bending moment applied to the plate. By convention at least 3 or 4 bicortical screws are applied dependent on the anatomic location to improve pull out strength of the plate.

An additional consideration is the stability/movement at the fracture site. The working length of the plate is defined as the distance between the nearest screws on each side of the fracture. Screws further from the fracture site (greater working length) allow greater bending to occur within the plate and thus provide greater movement/instability. Increased working length also places more stress on the plate as the plate deforms and a greater pull-out force on the nearest screws. It should be noted that this only applies when there is bending or torsional loading as under a purely axial load the screws would be in shear and the working length would be less important.

Figure 5

Conventional plates obtain fixation by compressing the plate against the bone as the screw gains purchase, thus stability is achieved by friction of the plate against the bone. (Figure 6 A and B) For this to occur the bone must be of sufficient quality for screw purchase to provide this compression, this can be problematic in osteoporotic patients. Screws can toggle in the hole in the plate and pull out along the line of least resistance. Each screw – plate interface acts as an independent point of fixation and can move and therefore fail independently. This means that the screw subjected to the greatest force can fail first, resulting in the subsequent screws being subjected to increasing force and failing sequentially. These plates frequently fail by screw cut out in this mode. Newer generations of plating systems include locking plates. These systems obtain fixation by a completely different mechanism (see Figure 6 C and D). The screws have a thread on their head with a corresponding thread in the plate hole, therefore after insertion the screw head locks into the plate. This affords angular stability to each screw and in effect the construct will act as a single structure once completed. This has several important implications for construct mechanics. Locking plate systems obtain stability by 3-point fixation via each screw, these being the locking mechanism between the screw and plate, the screw purchase in the near cortex and the screw purchase in the far cortex. Insertion of several screws therefore potentially generates an extremely stable construct in this regard. This also means that the plate can provide stability without being in contact with the bone, acting as an internal fixator. In unstable fractures, angular stability is inherent to the system and this can improve maintenance of reduction and reduce the risk of implant failure. As the screws are fixed into the plate and cannot toggle, it is less likely for screws to fail individually as on angular loading all of the screws will resist cut out. It is therefore less common that these locking plates fracture, however, if the fracture does not unite rapidly enough the plate may experience long term repeated loading and fail by subsequent fatigue failure, as has occurred in Figure 5.

Figure 6

Internal fixation nails

The mechanics of intramedullary nails in regard to their cross sectional geometry is discussed above. It is important to remember that due to the 4th power relationship between the radius and the strength and rigidity of the nail, small increases in nail size will have a profound effect on mechanics. An important aspect to consider is also the fit of the nail to the cortical bone support within the medullary canal (Figure 7.). Nails are generally long, with fixation at either end by a cluster of locking bolts. This leads to long moment arms and therefore the potential for large bending forces transferred to this fixation. When loaded in bending, a similar long plate and screw construct would result in very high stress in the screws which would likely fail due to the proximity of the screws to each other and the long moment arm. However, dependent upon the way that the nail has been inserted, the inner geometry of the bone generally provides support and limits the amount the rod may bend thus reducing the bending stress applied to the nail. Therefore, the nails primary function is to support axial and shear loads only. The stress within the screws, however, may be very complex. The mechanics generated will depend upon how close the fit of the nail is to the medullary canal and can be understood in terms of the nails working length. A narrow nail inserted into a large medullary canal will potentially not achieve any contact fit at the isthmus of the long bone. The working length in such a case is between the two set of locking bolts and large force transfers as described above can occur. By reaming the medullary canal, not only can a larger nail be inserted, but as a cylindrical channel is created the fit of the nail will potentially be much closer. If tight fit is achieved either side of the fracture, then the working length is effectively shortened between these points. If there is a close fit between the nail and the bone, the fixation bolts will generally need to resist only simple shear, as designed. However, as the fit gets wider, leading to a gap between the bone and the intermediary nail, the screws act as a cantilever at each end with complex bending taking place. This may lead to fatigue failure over time, as portrayed in Figure 7.

Figure 7

When designing or applying any implant for bone stabilisation, it is important to consider the effect of design and technique of application not only on the mechanical environment at the fracture or implantation site but also the forces in the implant itself. As alluded to above, the way that a device is applied may profoundly alter the stresses that it will be subjected to on loading. According to Newton's third law, every action has an equal and opposite reaction, this means that constructs which apply large forces to the implant will also potentially apply large reciprocal forces at the bone implant interface. This may lead to failure at this point. Perhaps the most profound example of this relates to the working length as described above. The same effect is illustrated in Figure 8 in the context of a plate. Here we see that moving fixation close to the focus of instability (fracture) reduces the working length of the implant, this will increase rigidity, reducing motion at the fracture site. This means that the implant must be sufficiently strong to withstand these forces if the implant is to be used in this mode. Fracture site mechanics influence healing and the effect of implant design and method of implantation must be considered taking both the effect on the fracture and the implant into account. This is a more complex situation than might initially be apparent as most constructs have degree of inherent stability imparted by the bone itself dependent on configuration. A well reduced fracture simple fracture with stable fixation will result in a load sharing construct with the majority of the force transfer being between the bone fragments rather than the implant itself. This protects the implant and results in a much lower risk of failure. A poorly reduced fracture or a more complex injury without inherent stability relies entirely on the implant which will act in a load bearing manner. This is described more fully elsewhere.(3)

Figure 8

External fixation

External fixation systems are in general highly modular, their mechanical performance is therefore dependent on their components alongside the construct that is assembled. Simple mono-lateral fixators utilise pins inserted into the bone connected outside the body by a series of rods and clamps. With regards to construction, perhaps the most important aspect to consider is that of working length. In general, this means having 2 points of fixation wherever possible and maximising moments and lever arms where they are advantageous and reducing them where they might increase force transfer and instability (increasing span over stable segments and reducing span across areas of instability). This leads to the concept of near-far fixation which is covered more fully elsewhere.(4) The structural rigidity will depend upon the material from which they are made and their design. Most external fixator pins are stainless steel. Whilst titanium pins exist, they are less rigid and offer little advantage in other respects for this application. Older systems employ hollow metal connecting rods, with almost all newer rods being made of carbon fibre composite. These are more rigid, lighter and are safe to be placed inside a magnetic resonance imaging scanner should the patient require this. The majority of important components are simple rods, which as described above, will have a bending rigidity proportional to the 4th power of their radius. Therefore, increasing the diameter of bone fixation pin from 4 to 5mm will result in 2.5-fold increase in its rigidity; a 6mm pin is twice as rigid as a 5mm pin. Large 12mm interconnecting bars found in modern fixators are more than 5 times more rigid than older 8mm equivalents. Selecting modern systems with more rigid components allows stabile constructs to be constructed with more simple designs and fewer parts.

Multi-planar frames, particularly those utilising fine wires such as the Ilizarov fixator, have more complex mechanics. The function of the pin or tensioned wire is to provide support to the relative bones whilst healing (Figure 9). This demonstrates that transfixation wires will potentially generate a beam-loading mechanical environment when axial loads are applied. Pre-tensioned wire mechanics are complex as the wires are flexible and the tension will additionally increase when the frame is loaded resulting in the wires self-stiffening. This is traditionally understood to generate favourable fracture healing mechanics by permitting a degree of axial motion whilst preventing shear.(3) Uniplanar half pins generate different mechanics by creating a cantilever bending mechanical environment, this leads to different forces at the fracture site and may induce unfavourable shear (Figure 10). Pins and wires are often applied asymmetrically to the limb to minimise soft tissue trauma. This creates a non-linear deformation during gait as the bone is held more rigid relative to a given plane (closer to the edge of the frame). Combinations of pins and wires in different planes lead to complex mechanics and there is the potential for generating shear by cantilever bending creating shear when half pins are included in constructs. However, when utilising these in particular configurations this appears to be at levels unlikely to be of clinical importance.(5)

Figure 9

Pins may fatigue at the point of maximum bending moment and maximum stress concentration, both of which exist at the junction with the bone, the stress concentration coming from the presence of threads. Wires are thin (~2mm) and tensioned so experience primarily an axial load similar to a stretched cable uniquely allowing the bone to move relative to the wire; there is also a knurled version of the wire (olive wires) that is textured and designed not to move in the bone. The pin/wire provide support to the bone with the shorter pin or greater tensioned wire providing the most support. The absence of pins on a single side means that there is locally less restriction to movement and the fracture is more likely to toggle about the more rigid side.(5)

Figure 10

The frame that supports the pins and wires and hence the fracture, is usually constructed from individual modular components. Each frame is therefore a unique structure with its form depending on the characteristic of the fracture and the patient that restricts assembly. Traditional frame structures are generally far stiffer than the pins and wires and can be considered almost rigid in comparison.(5) The amount of bending or torsion that occurs within the frame is a function of the position of the axial support rods relative to the axes that the bending moment is applied. Hence if the rods are further away from the bending axis (neutral axis) the stiffness of the frame will increase. Straight rods generally provide more resistance to bending than hexapod rods that are positioned at an angle. Under torsional loads however, a hexapod arrangement provides more torsional support than straight rods. This occurs because under bending, straight rods experience greater axial compression/tension and under torsion the straight rods act more like cantilever beams. Thus the more axially the load is applied to an individual component the greater the resistance to deformation of that component (Figure 1). It must be noted that there are many components within an Ilizarov frame so the ultimate support provided is complex. The actual load applied by the patient will produce a combination of axial, bending, and torsion that will be different in the pin/wire, frame and rod and will be furthermore dictated by patient weight and the particular activity they are doing.

Conclusion

To promote bone healing medical devices are intended to provide a specific level of strain or deformation to the fracture site. An understanding the fundamentals of stress is important in the pre-operative planning /selection of these devices to complement the specific support that is intended.

Acknowledgements

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Glossary

Stiffness – a material characteristic expressing deformation under load. Proportional to the gradient for the linear part of the stress-strain curve for that material.

Rigidity – characteristic of a construct expressing the deformation under loading. Determined by the materials stiffness, the geometry of the construct and the direction and type of load applied.

Neutral axis – Theoretical axis in a construct that is subjected to zero net force when a load is applied in bending or torsion.

Area moment of inertia – expresses a structures material distribution in cross section in relation to bending strength and stiffness.

Polar moment of inertia – expresses a structures material distribution in cross section in relation to torsional strength and stiffness.

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Figure and Table Titles

Figure 1 – Comparison of Axial, Torsion and Bending for an annular cross section. where, $I = \frac{\pi({D_o}^4 - {D_i}^4)}{64} = 4269mm^4$, $J = \frac{\pi({D_o}^4 - {D_i}^4)}{32} = 8537mm^4$; representing the moment of inertia of area and polar moment of inertia respectively. Assuming: L = 0.5m, D_i = 30 mm, D_o = 36 mm (Cross sectional area "A" =311 mm^2), E = 10 GPa, G = 1 GPa, F = 1000 N, T = 10Nm.

Figure 2 – Example of the neutral axis under bending load. Note that material further from the neutral axis is subjected to increasing tension on the convex side and increasing compression on the concave side. The material along the neutral axis is subjected to zero net force along the axis.

Figure 3 – Example formulae for the moment of inertia and polar moment of inertia of commonly encountered orthopaedic implants.

Table 1 – Bending stiffness (EI in GPa m^4) of common object cross-sectional shapes made of steel or titanium (cross sectional area = 311 mm^2). E Steel = 200 GPa, E Titanium = 110 GPa. Note: the bending axis is perpendicular to this cross-section with the load applied vertically.

Figure 4 – Examples of different trauma implants with design features which attempt to mitigate stress concentration in common areas of weakness. Note the lack of features in the areas of maximal bending stress in typical injury. A) Plate for distal radial fracture – note that the plate is thickened and there are no holes in the area of the plate where bending loads will frequently be applied in unstable fractures. B) Dynamic compression plate – note the scalloping of the plate to reduce stress concentration around features and allow smooth contouring of the implant. C) Intramedullary nail to stabilise a transverse osteotomy which is inherently unstable – note that a plate has been added to help neutralise shear and bending forces and reduce loads at the locking bolts. D) An intramedullary hip screw – note that the nail is thickened proximally at the area where maximal load is usually applied and to help mitigate for the large hole for the screw in the femoral head. The surgeon has added a circlage wire at the fracture in an attempt to reduce bending forces on the nail and maintain contact. This was ultimately unsuccessful (inset) and the nail underwent fatigue failure at the large hole proximally.

Figure 5 – Clinical example of a fractured plate and basic schematic of plate / screw configuration with resultant stresses. Note that the plates have both undergone failure at screw holes which corresponds to points of stress concentration at the fracture site.

Figure 6 – Comparison of fixation method for non-locked and locked plating systems. A – each screw exerts a compressive force on the plate, maintaining stability by friction of the plate against the bone. B – Forces are applied to each screw and as these move they fail sequentially resulting in cut out. C – Locked plates have screws which lock into the plate itself – these obtain multiple point fixation and form an extremely rigid construct in the intact bone segment (red boxes). D – this transfers force to the plate and therefore these systems will tend to fail by implant fracture rather than screw pull out.

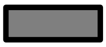
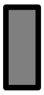
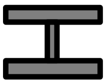
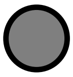
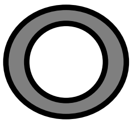
Figure 7 - Schematic of femoral nail fit and clinical fracture. With close fit screws experience shear only, whereas a wide fit leads to cantilever bending.

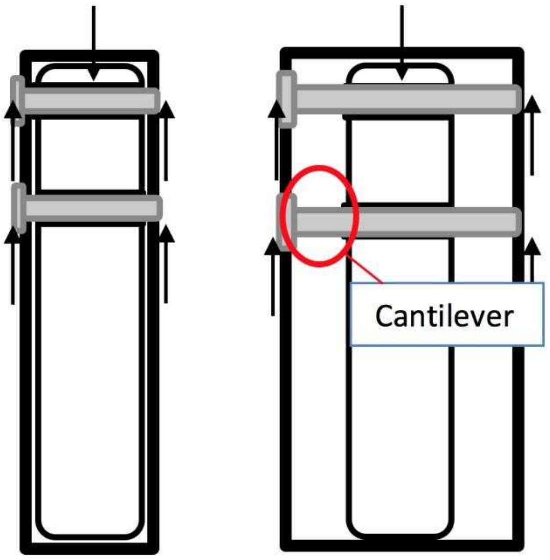
Figure 8 – Working length of a plate illustrating the effect of moving fixation relative to the fracture site. A – Fixation near and far to the fracture site increases working length, reducing fracture site motion and increasing stress concentration in the plate. B – Moving fixation away from the fracture site increases working length, increasing fracture site motion and reducing stress concentration in the plate.

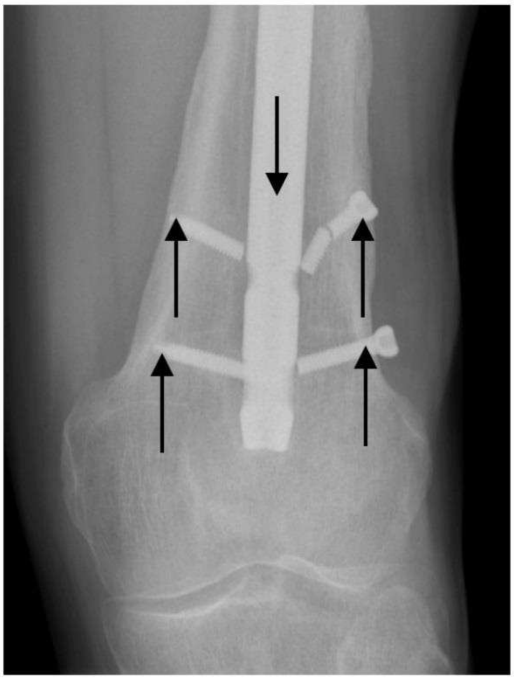
Figure 9 - Example of Hexapod (A) and Ilizarov frames (B) with pins and tensioned wires.

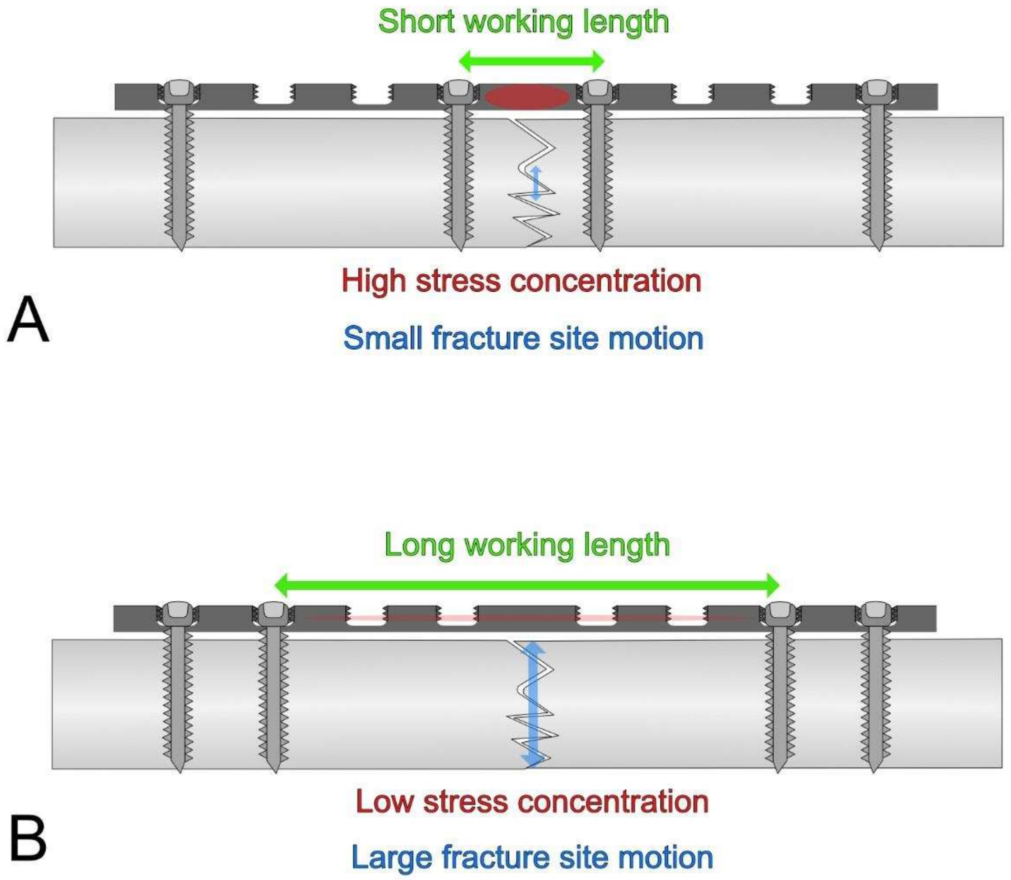
Figure 10 – Mechanics of ring and mono-lateral external fixators. A - ring fixator with tensioned fine wires. On loading the wires are tensioned further and beam loading is generated. This results in controlled axial motion at the fracture site. B – mono-lateral fixator with rigid half pins. On loading a torsional force is generated and cantilever bending occurs. This results in shear at the fracture site.

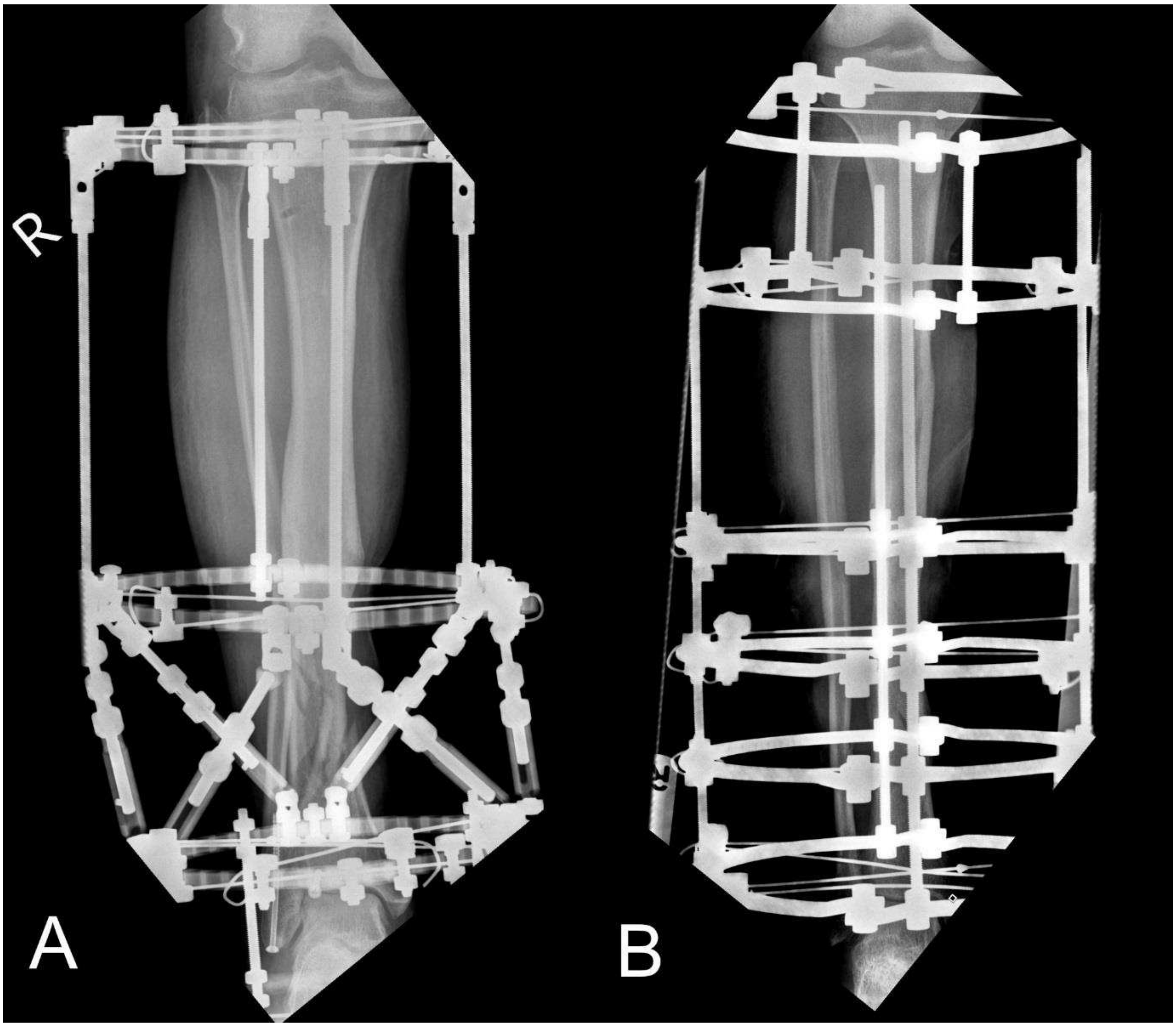
Table 1 – Bending stiffness (EI in GPa.m^4) of common object cross-sectional shapes made of steel or titanium (cross sectional area = 311 mm^2). E Steel = 200 GPa, E Titanium = 110 GPa. Note: the bending axis is perpendicular to this cross-section with the load applied vertically.

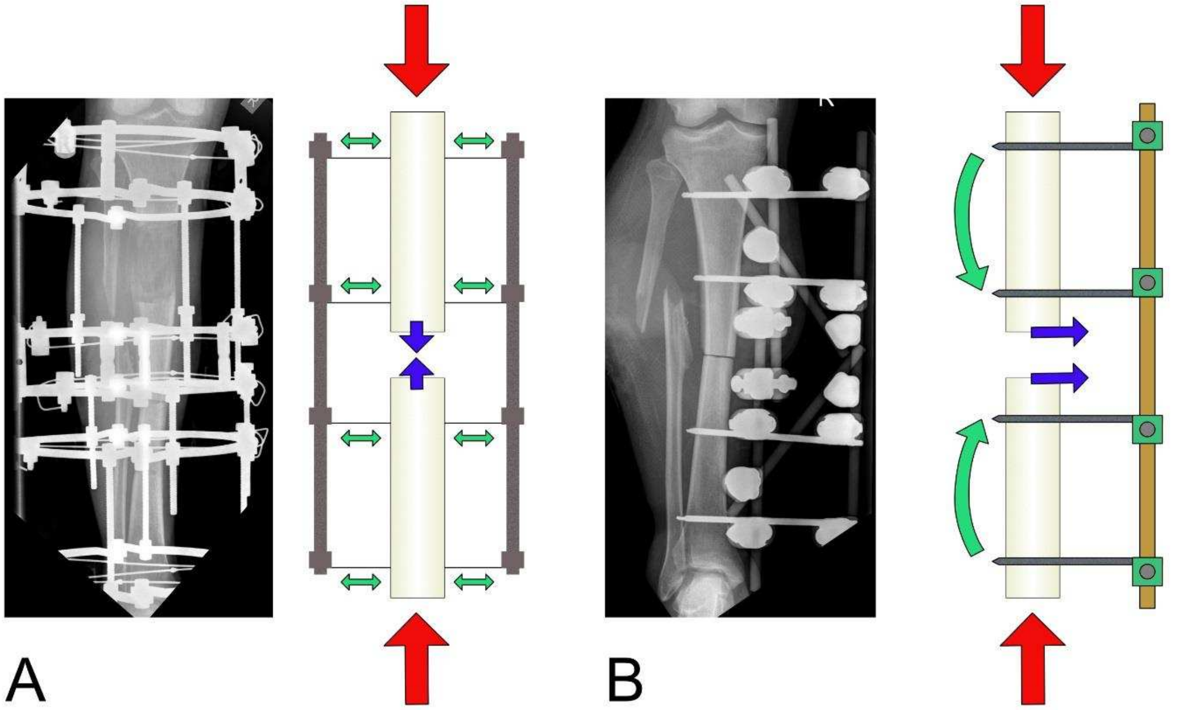
	Beam	Cross-section	I, moment of inertia of area mm^4	EI_{steel} GPa.m^4	$EI_{titanium}$ GPa.m^4
i	31mm wide rectangular plate 10mm thick		I = 31 x 10^3/12 2583.3 mm^4	516	284
Ii	10mm wide rectangular plate 31mm thick		I = 10 x 31^3/12 24825.8 mm^4	4965 9.6 x i	2730
Iii	Two stacked rectangular plates (31 mm wide x 5 mm thick) separated by 10mm I = [$I_{outside}$ - I_{inside}]		I = [(31x20^3/12) - 31x10^3/12] 18083.3 mm^4	3616 7 x i	1989
iv	19.9 mm diameter Solid rod.		$I = \frac{\pi(19.9^4)}{64}$ 7698.1 mm^4	1539 3 x i	846
v	36 mm diameter hollow tube, 3.0mm thick (Figure 1).		$I = \frac{\pi(36^4 - 30^4)}{64}$ 42687.2 mm^4	8537 16 x i 5.5 x iv	4695

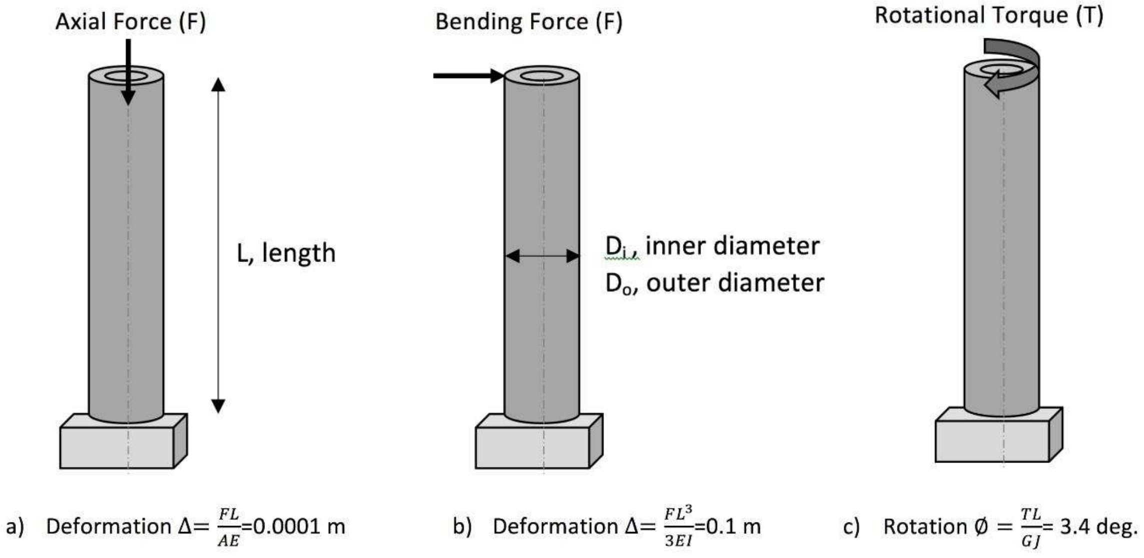

Cantilever

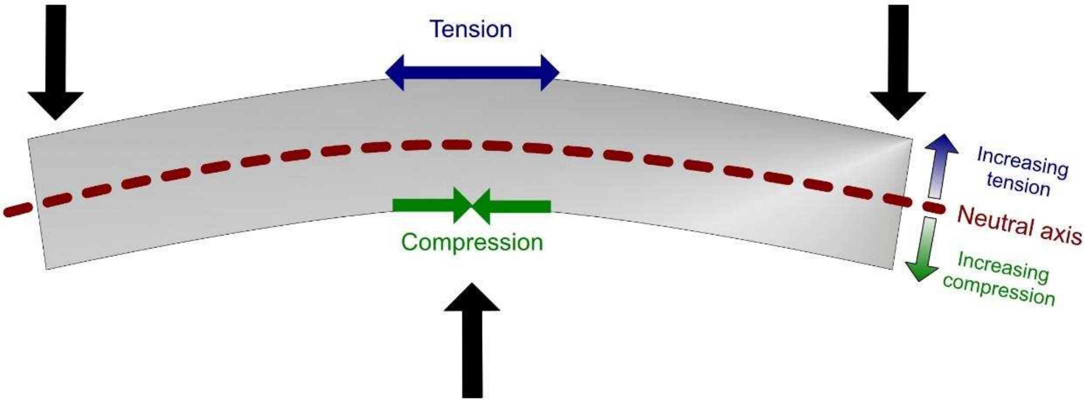


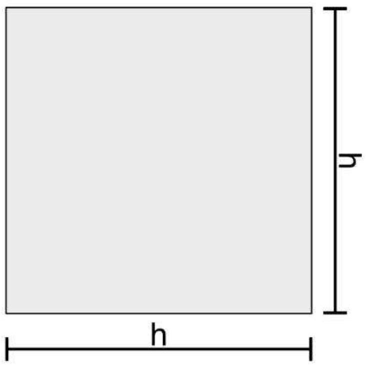

Short working length
High stress concentration
Small fracture site motion
A
Long working length
Low stress concentration
Large fracture site motion
B


R
A
B


A
B


Axial Force (F)
L, length
Bending Force (F)
D_i, inner diameter
D_o, outer diameter
Rotational Torque (T)
a) Deformation $\Delta = \frac{FL}{AE}$=0.0001 m
b) Deformation $\Delta = \frac{FL^3}{3EI}$=0.1 m
c) Rotation $\emptyset = \frac{TL}{GJ}$= 3.4 deg.

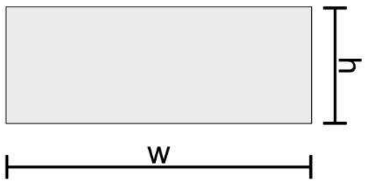

Tension
Increasing tension
Neutral axis
Increasing compression
Compression



Square prism

Moment of inertia (I) - $h^4/12$

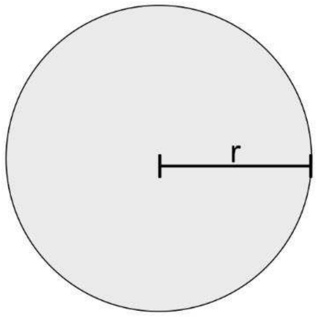
Polar moment of inertia (J) - $0.141h^4$



Rectangular prism (plate)

Moment of inertia (I) - $wh^3/12$

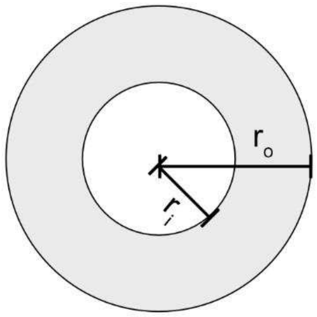
Polar moment of inertia (J) - $wh(w^2+h^2)/12$



Solid rod (solid nail)

Moment of inertia (I) - $\pi r^4/4$

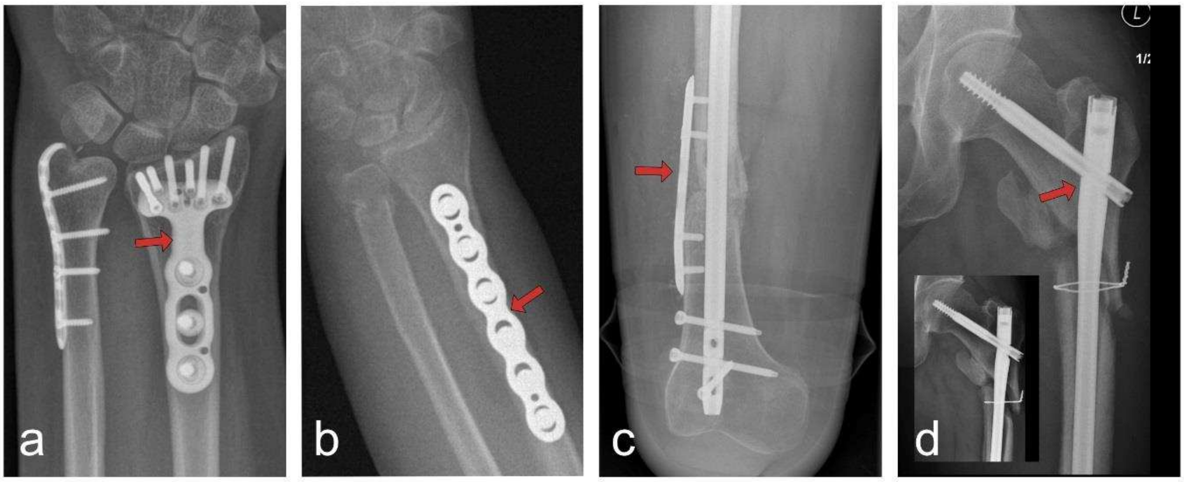
Polar moment of inertia (J) - $\pi r^4/2$

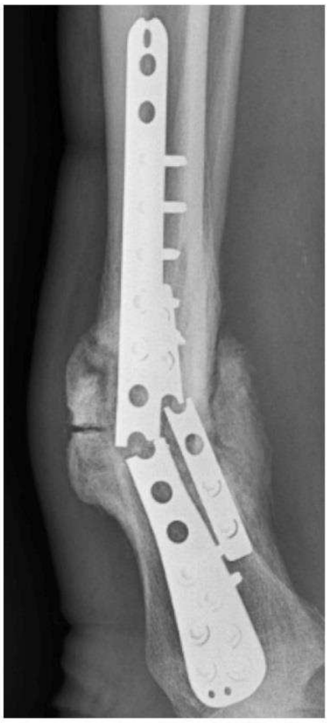


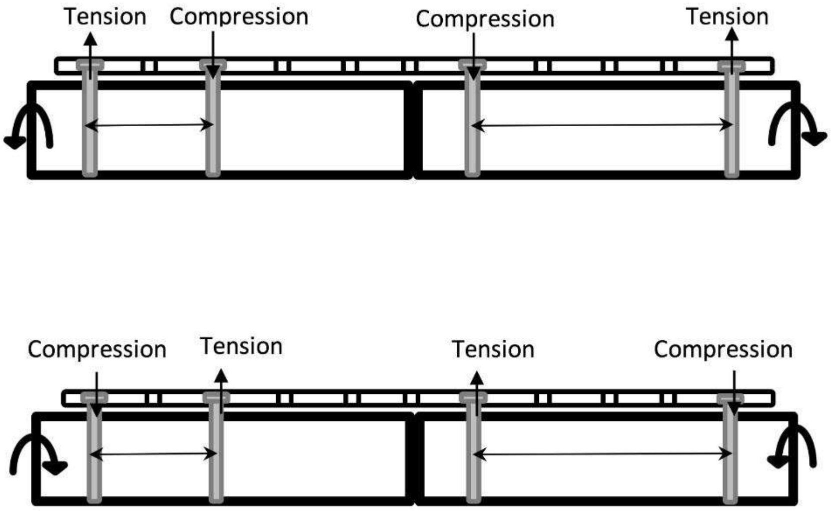
Tube (cannulated nail)

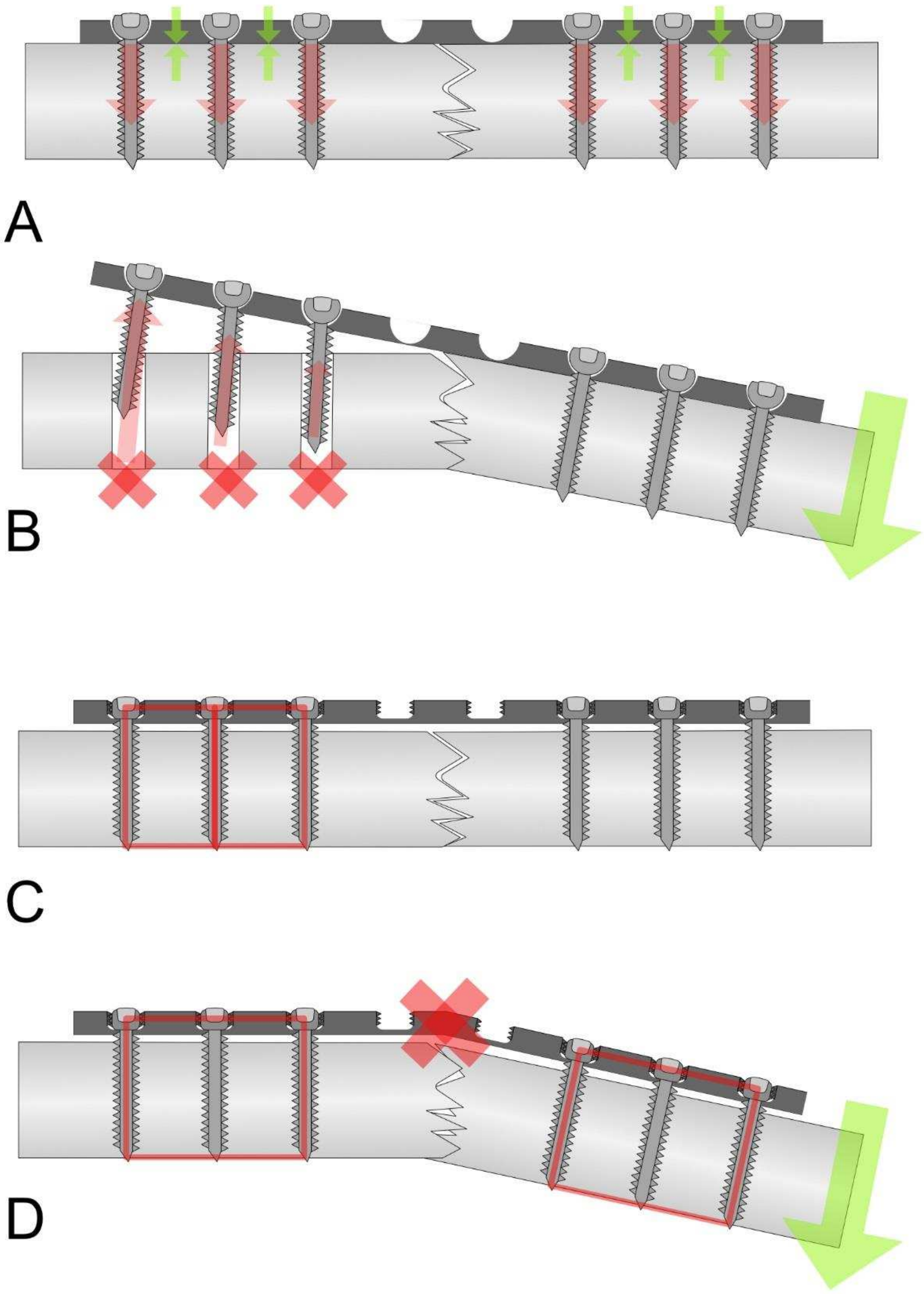
Moment of inertia (I) - $\pi(r_o^4-r_i^4)/4$

Polar moment of inertia (J) - $\pi(r_o^4-r_i^4)/2$


a
b
c
d
L
1/2




Tension
Compression
Compression
Tension
Compression
Tension
Tension
Compression


A
B
C
D