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. This paper provides examples to reinforce basic structural mechanics applied to medical devices.

Keywords force; geometry; materials; mechanics; moment; orthopaedics; stress; structural support; trauma

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 to the patient. Many of these implants were originally produced empirically, based on a 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.

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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.

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 at its 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 to 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 one to two orders of magnitude stiffer than bone. Additionally, the Elastic modulus of medical Titanium alloys ($E \sim 110$ GPa) is half that of Stainless Steel or Cobalt Chromium ($E \sim 200$ 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

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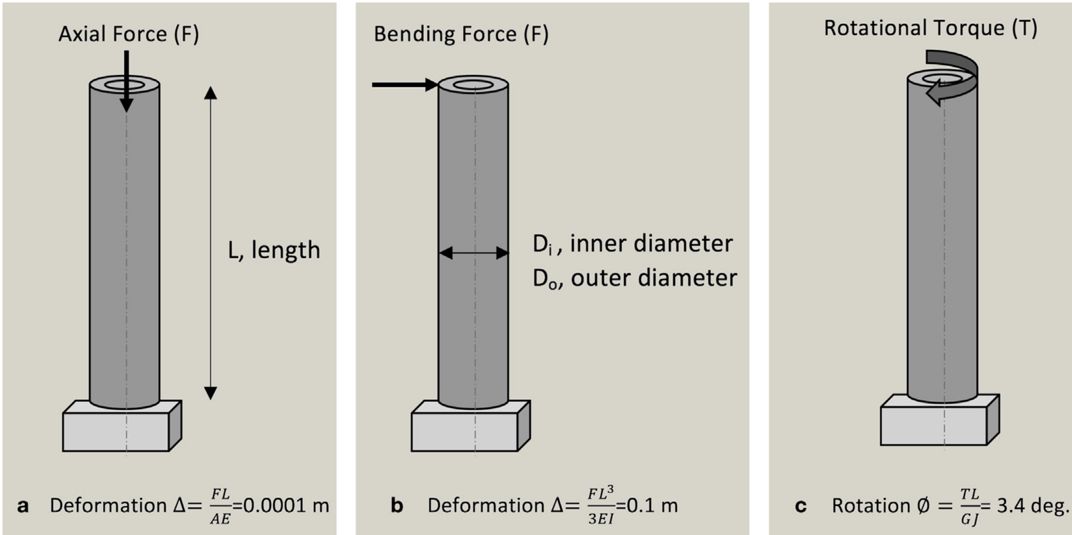

Figure 1 Comparison of Axial, Torsion and Bending for an annular cross section. where, $I = \frac{\pi(D_o^4 - D_i^4)}{64} = 4269\ mm^4$, $J = \frac{\pi(D_o^4 - D_i^4)}{32} = 8537\ mm^4$; representing the moment of inertia of area and polar moment of inertia respectively. Assuming: L = 0.5 m, D_i = 30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

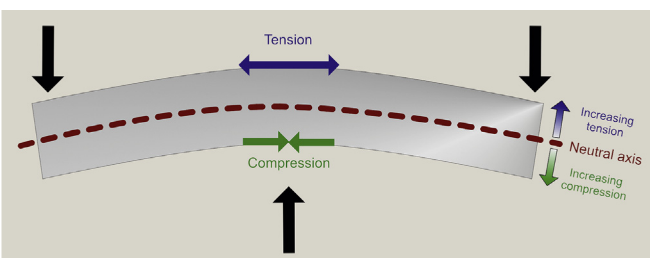

Figure 2 Example of the neutral axis under bending load. Note that material further from the neutral axis is subjected to increasing tension on the convex side and increasing compression on the concave side. The material along the neutral axis is subjected to zero net force along the axis.

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 inertia 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.

Different mathematical formulae describe 'I' for objects with different cross-sectional shapes under different types and directions 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 circle's 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]

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 four, hence for any given cross-sectional area I and 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 or J would decrease by a factor of five ($I_{anular}/I_{solid} = 5$). 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 9 mm intramedullary nail would be approximately 50% more rigid than an 8 mm nail. This effect is seen in ageing osteoporotic bones, where the amount of osseous material available is decreased. Again, under the effect of Wolff'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

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