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BASIC BIOMECHANICS AND BIOMATERIALS
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ABSTRACT

This paper outlines the basic knowledge that should form an integral component of a training programme in basic biomechanics and biomaterials for orthopaedic residents. For a comprehensive learning package the reader is directed to the substantive textbooks listed in the suggested reading section.

1. INTRODUCTION:

A basic understanding of the principles of biomechanics and biomaterials has become a prerequisite for practicing orthopaedic surgeons and forms an important component in the education of orthopaedic trainees. Biomechanics is a sub-branch of the general field of mechanics and, as such, it embraces the fundamental principles of this discipline and applies them to the human body considered as a load bearing structure. Similarly the term biomaterials refers to the study of the ability of both natural and synthetic materials suitable for implantation to withstand internally and externally generated loads resulting from physiological activities.

2. SCALAR AND VECTOR QUANTITIES

Two basic types of quantities scalars and vectors are used in biomechanical analysis and these are defined as:

Scalar - a quantity that is completely defined by its magnitude (e.g. temperature, mass, volume, time…): described by a number.

Vector - a quantity that is defined by magnitude, direction, sense and point of application (e.g. force, displacement, velocity…) described by an arrow which identifies the point of action and direction and the length of which defines the magnitude.

Operations on vectors V_1 and V_2 be performed geometrically in order to determine the direction and magnitude of the resultant vector as illustrated in Figure 1.

Fig 1

3. BASIC PRINCIPLES OF BIOMECHANICS:

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

Kinematics: The science of pure motion, considered without reference to the objects moved, or to the force producing or changing the motion (e.g. trajectory, velocity, acceleration). Motion is defined as a change in position with time. Any complex motion comprises a series of translations and/or rotations. The term translation is reserved for a type of motion in which all the points in the body are subject to the same displacement vector while rotation refers to a type of motion characterised by the fact that different points in the body are subject to different displacement vectors (Figure 2).

Fig 2

Statics: The branch of mechanics that deals with the forces acting on an object at rest (or moving with constant velocity with zero acceleration). The forces which act on an object can be made up of direct forces or moments, a pair of equal, parallel and opposite forces separated by a distance from one another and acting on the body. The application of forces and moments to a body has the effect of changing its state of rest. The action of a force on a body will induce a change in the state of rest by producing a linear acceleration of the body, i.e. a change in its velocity; the action of a moment on a body will induce a change in the state of rest by producing angular acceleration of the body, i.e. a change in angular velocity.

The underlying principle of static analysis is that of equilibrium. A body at rest (or moving with constant velocity) is said to be in equilibrium if the sum of all applied loads (i.e. forces and moments) is zero. With reference to human joints the applied forces are typically the external loads such as the body weight component, and internally generated loads such as the muscle forces generated to maintain

the joint in equilibrium. The concept of equilibrium is used in the static analysis of joint loading. In this case, the joint of interest is considered in isolation from the rest of the body and all forces and moments acting on it are identified, thus establishing the free body diagram for the joint. The equilibrium condition is then applied in order to establish the resultant joint reaction force. An example of this is given in Figure 3 where the hip joint reaction force is calculated for a subject in single leg stance.

Fig 3

Dynamics: The branch of mechanics that deals with the forces acting on an object and the changes they produce on the motion of the object. Essentially, dynamics incorporates both the principles of statics and kinematics in that it examines the actions of forces and the resulting accelerations and motions of the object. In the context of orthopaedic biomechanics, dynamic analysis frequently is applied to activities such as gait studies. The approach used in this application is to determine, at any point in time, the acceleration of the body parts under consideration and the forces required to produce these accelerations; to build up the resultant forces at each time interval using the static analysis methods described above thus obtaining the resultant forces over the range of movement of interest.

4. LOAD-DEFORMATION AND STIFFNESS, STRESS-STRAIN AND ELASTICITY:
When a structure is subjected to loads this produces a deformation. The plot of load, F, versus deformation, d, is a function of the material properties and the geometry of the structure, therefore changes in either the material or the geometry of the structure or both will be reflected in changes in the load-deformation diagram (Figure 4a). The slope of the linear part of the load-deformation plot is known as *structural stiffness, F/d*, while the area under the load-deformation curve is a measure of the *energy* absorbed by the structure as it is being deformed. According to the shape of the load-deformation curve different regions of interest can typically be identified. The elastic region is characterised by a linear relationship between load and displacement; within this region the deformation induced by the application of the load is totally recovered on removal of the load. In the plastic region the linear relationship between load and displacement is not maintained, in this region some permanent deformation associated with yielding of the structure occurs, this deformation cannot be recovered upon removal of the applied load. The final region of interest is characterised by the fact that the integrity of the structure is lost resulting in failure.

Fig 4

In order to characterise the stiffness of the material it is necessary to utilise quantities which are independent of the shape of the structure (Figure 4b). This is achieved through what is known as structural diagram of a material, a plot of the stresses induced as the material is strained. The stress-strain diagram is a characteristic of the material and is independent of the actual shape of the object tested. Stress is the ratio between the applied load and its area of application. Depending on the relative position of the line of action of the applied load and the area of application, stresses can be identified as normal or shear. Normal stress, σ, is defined as the load, F, divided by the area perpendicular to the line of application of the force (eq.1). Normal stresses can be *tensile* or *compressive* depending on the direction of the applied load. Shear stress, τ, is defined by load, F, divided by the cross sectional area, $A*$, parallel to the line of application of the force (eq. 2).

$$\sigma = \frac{F}{A} \text{ (eq. 1) and } \tau = \frac{F}{A^*} \text{ (eq. 2)}$$

Stresses have units of N/mm^2 or Pa**.**

Strain is the ratio between the deformation induced by the applied stress and the un-deformed object. Similarly to stresses, strains are distinguished into normal and shear according to the nature and origin of the induced deformation, thus the normal strain, ε, is induced by a normal stress, and is a measure of the change in length, ΔL, divided by the original length, L, (eq. 3). Equally, shear strain, γ, is induced by a shear stress, τ, and is a measure of the change the angle of two lines originally at right angles two each other (eq. 4)

$$\varepsilon = \frac{\Delta L}{L} \text{ (eq. 3) and } \gamma = \frac{\Delta \phi}{\phi} \text{ (eq. 4)}$$

Strain is a dimensionless quantity and is expressed as a percentage strain or typically as microstrain.

The slope of the linear part of the stress-strain curve in the case of normal stress and strain is termed the Young's modulus or Elastic Modulus, E, (eq. 5). In the case of the shear stress and shear strain plot, the slope is termed the modulus of rigidity or Shear Modulus, G, of the material (eq.6).

$$E = \frac{\sigma}{\varepsilon} \text{ (eq. 5) and } G = \frac{\tau}{\gamma} \text{ (eq. 6)}$$

The units of the Young's Modulus and the Shear Modulus are the same as stress (Pa). The moduli are properties of the material and are totally independent of the geometry of the sample. Typical values of Young's Modulus are around 200GPa for stainless steel or cobalt chrome, 100GPa for titanium, 20GPa for cortical bone and 2GPa for PMMA bone cement.

5. STRESS ANALYSIS:
The distribution and magnitude of the stresses generated within a structure due to the application of external forces is determined by the cross-sectional geometry and the properties of the material characterising the structure under investigation and the nature of the external load. The study of the stress field induced within a structure is known as stress analysis.

5.1 Axial load, axial stress and axial stiffness
An axial load consists of an external force vector F acting along the longitudinal axis of the structure (Fig 5). Axial stresses arise within a structure subject to an axial load. The magnitude of the axial stresses thus generated is constant (Fig 5) and is proportional to the modulus of the applied force and inversely proportional to the cross-sectional area of the structure:

$$\sigma_a = \frac{F}{A} \text{ (eq. 7)}$$

Axial stiffness is the resistance of a structure to axial deformation and is a function of both the material and the way in which the material is deployed. It is essentially proportional to the slope of the load deformation plot.

Axial stiffness proportional to EA

5.2 Bending moment, bending stress and bending stiffness
A bending moment consists of an external force couple, M, acting perpendicularly to the longitudinal axis of a structure (Fig 5). Bending stresses arise within a structure subject to a bending moment. The distribution of bending stresses within the cross-section is non-uniform (Fig 5), with tensile, positive, stresses on the convex deformed side and compressive, negative, stresses on the concave deformed side. As the stress varies from positive to negative there exist a region within the cross-section that is characterised by a null stresses, this takes the name of neutral axis of the structure. At each point within the cross-section the magnitude of the stresses is proportional to the applied bending moment, M, proportional to the distance from the neutral axis, y, and inversely proportional to the second moment of area, I, a measure of the way the material is distributed about the neutral axis (Fig 6):

$$\sigma = \frac{My}{I} \text{ (eq. 8)}$$

For a rectangular beam of width b and depth h the second moment of area is given by the expression:

$$I = \frac{bh^3}{12} \text{ (eq. 9)}$$

therefore small changes in the depth of the beam, h, have a profound effect on the second moment of area and hence on the stresses.

Bending stiffness is the resistance of a structure to bending deformation and, again is a function of the material and the way the material is distributed in the structure and given by the slope of the load deformation plot.

Bending stiffness proportional to EI

5.3 Torsional load, torsional stress and torsional stiffness

A torsional load consists of a force couple, T, acting along the longitudinal axis of a structure (Fig 5). Torsional stresses arise within a structure subject to a torsional load. The distribution of torsional stresses within the cross-section is non-uniform (Fig 5). The magnitude of torsional stresses is proportional to the applied torsion, T, to the distance from the axis of the structure, r, and inversely proportional to the polar moment of area, J, a measure of the way the material is distributed about the axis (Fig 6):

$$\tau = \frac{Tr}{J} \text{ (eq. 10)}$$

For a solid rod of diameter d the polar moment of area is given by the expression:

$$J = \frac{\pi d^4}{32} \text{ (eq. 11)}$$

therefore a small change in the rod diameter has a profound effect on J and hence on the shear stresses.

Torsional stiffness is the resistance of the structure to torsional deformation determined by the slope of the load deformation plot.

<div align="center">

Torsional stiffness proportional to GJ

</div>

<div align="center">

Fig 5

Fig 6

</div>

5.4 The importance of structural stiffness

As shown previously, the structural stiffness is significantly influenced by the cross-sectional geometry of the structure and the properties of the material. For example, two hip stems characterised by the same geometry but made from two different materials such as, for example, titanium and cobalt chrome will be characterised by structural bending stiffnesses directly proportional to the two different values of Young's modulus for these materials, i.e. the cobalt chrome stem will have twice the structural stiffness of the titanium stem. However, small changes in cross-sectional geometry will result in significant changes in stiffness. For example, in the case of a hip stem of rectangular cross-section, an increase of 25% in the medio-lateral dimension will, in effect, almost double it's bending stiffness. This is easily understood by increasing the value of h by 25% in equation 9.

<div align="center">

i.e. $I = bh^3/12$ for $h=1.25$: I is proportional to $1.25^3 = 1.95$

</div>

A 95% increase in bending stiffness for the same material!

The structural stiffness of implants is of paramount importance in orthopaedic applications as it determines the extent of load sharing between the implant and the host tissue. Given an implant characterised by much greater structural stiffness that the supporting host it will carry a higher proportion of the shared load. This phenomenon is known as stress shielding and can contribute to bone loss.

6. BIOMATERIALS

The term biomaterials can be applied to the natural materials found within the body, such as bone, connective tissue, cartilage etc. and to the synthetic materials used to replace the natural ones such as stainless steel, cobalt-chromium alloys, titanium alloys, polymers, ceramics and composites.

Synthetic materials are classified according to their biocompatibility and their mechanical properties, such as tensile, compressive and shear strength, hardness, toughness, fatigue resistance and creep behaviour. Factors such as ease of manufacture, cost and production quality will dictate the potential for application of a biomaterial in orthopaedics.

6.1 Static mechanical properties

Mechanical strength is typically evaluated from the stress-strain curve obtained from a tensile (or compressive) test. The overall shape of the stress-strain curve depends on the nature of the material being tested. For example, in the case of metals a typical stress-strain curve will be characterised by a linear, elastic region followed by a region of plastic deformation where the relationship between stress and strain is no longer linear (Fig 7). The point that demarcates the deviation form linearity is known as the yield point of the material. A material stressed beyond its yield point will not fully recover the induced strain if the stress is relieved but will retain some permanent deformation. Typically, after yielding the stress in the material will continues to increase up to a maximum in the stress-strain curve, this is known as the ultimate strength of the material. Having reached this maximum, the stress will decrease and the material will exhibit necking until ultimate failure. Materials characterised by a well defined plastic region are said to be ductile (Fig 7). Some materials including cast metals and ceramics do not demonstrate a yield point and plastic region. In these materials fracture will occur without any permanent deformation. These materials are said to be brittle (Fig 7).

Fig 7

Some natural materials, such as ligaments, for example, are characterised by a stress-strain curve which does not demonstrate the normal elastic and plastic regions that are characteristic of metals. In this case the characteristics of the stress strain curve reflect the microstructure of the materials. Ligaments are composed of collagen fibres and organic matrix, the crimps within the bundles of collagen fibres allow a high level of deformation to occur at low stress levels. This is reflected in the characteristic toe-region exhibited by the stress-strain curve for such highly organised structures (Fig 9). Once the collagen fibres are stretched and become aligned in the direction of applied load the material exhibits much increased stiffness. This behaviour is highly adapted to the function of the tissue allowing some laxity in the early stages of motion but providing a high level of restraint at the extremes of movement.

6.2 Anisotropy
The mechanical properties of most synthetic materials are uniform along any direction, such materials are defined as being isotropic as they exhibit the same stress-strain characteristics irrespective of the direction of loading. This is not true for natural materials which are characterised by a high level of anisotropy, a very strong dependence of mechanical properties upon orientation. This characteristic is due to the structural make up of the constituent elements of natural materials. For example, in long bones the principal orientation of the collagen matrix is along the planes of maximum loading. This high level of structural hierarchy equips bone to withstand the typical physiological loading regimes that are associated with its functional requirements while minimising energy expenditure.

6.3 Viscoelasticity
Another characteristic of most natural materials is viscoelasticity, essentially this means that the material is strain-rate sensitive. A physical model used to characterise a viscoelastic material is a combination of springs and dash-pots, as the load is applied the spring takes up some load and the dash-pot gradually absorbs some of the load. At high rates of strain the material exhibits a stiffer characteristic as shown in Figure 8. This essentially is the way the body copes in nature with impact loads, the viscoelastic nature of the material allows the body to deal with high rates of loading without fracturing.

Fig 8

6.4 Fatigue
Fatigue is a phenomena whereby in the presence of cyclic loading, a material may fail at stress levels below the static yield stress. Fatigue testing of materials involves cyclically loading the material at different stress levels until failure occurs. A plot of stress versus number of cycles of loading, called the S-N curve, identifies the stress below which fatigue will not occur and this is called the *fatigue endurance limit*. In designing metal implants for use in the body it is important to ensure the stress levels encountered are below this endurance limit to avoid failure. Fatigue failures of metal implants occur in orthopaedics particularly if there are *stress risers* or local areas of stress concentrations in the implant. Stress risers are geometric features which raise the local stress in a structure they can be features of the design such as, screw holes in fracture plates, or defects, notches or scratches in the

implant. Even the presence of a screw hole in a bone after removal of a fracture plate can precipitate a stress concentration related fracture of the bone.

CONCLUSIONS:
The material presented sets out the basic concepts of biomechanics and biomaterials to provide an overview and some basic definitions and concepts. This will hopefully enable the reader to study in greater depth some of the textbooks referenced in the bibliography below.

BIBLIOGRAPHY:

Basic Biomechanics of the Musculoskeletal System
By Margareta Nordin, Victor Hirsch Frankel
Published by Lippincott Williams & Wilkins, 2001
ISBN 0683302477, 9780683302479
467 pages

Musculoskeletal Biomechanics
By Paul Brinckmann, W. Frobin, Gunnar Leivseth
Published by Thieme, 2002
ISBN 3131300515, 9783131300515
243 pages

Biomechanics and Biomaterials in Orthopedics
By Dominique G. Poitout, Reinat Kotz
Contributor Dominique G. Poitout, Reinat Kotz
Published by Springer, 2004
ISBN 1852334819, 9781852334819
654 pages

FIGURE CAPTIONS:

Fig 1: Given two vectors v_1 and v_2, the vector sum (v_1+v_2) can be graphically calculated by transposing the tail of the first vector (v_1) onto the arrowhead of the second (v_2). The resultant vector is obtained by joining the tail of the first vector (v_1) to the arrowhead of the second (v_2). To calculate the vector difference (v_1-v_2) it is necessary to invert the orientation of the second vector (v_2) and repeat the procedure described for the addition

Fig 2: Translation is a type of motion in which all the points in the body are subject to the same displacement vector (d); rotation refers to a type of motion characterised by the fact that different points in the body are subject to different displacement vectors (d and s).

Fig 3: Hip joint reaction diagram in single leg stance M = abductor muscle force, W = body weight component and J = hip joint reaction force

Fig 4: Load deformation curve (a) and equivalent stress strain diagram (b)

Fig 5: Axial stress, normal and shear stress components on plane, bending stress and torsional stress
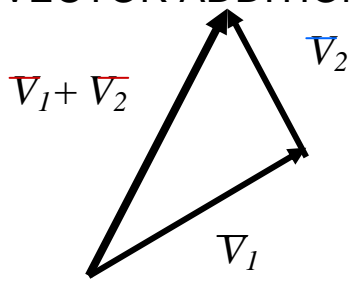
Fig 6: Second moment of area and polar moment of area for typical cross sections

Fig 7: Typical stress-strain curves, for brittle materials, ductile materials and ligament

Fig 8: Stress-strain curve at increasing strain rates for a typical viscoelastic material

V_1

V_2

$-V_2$
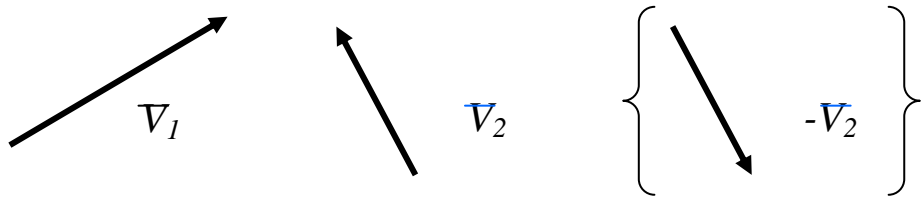
VECTOR ADDITION

VECTOR SUBTRACTION

V_2

$V_1 + V_2$

V_1
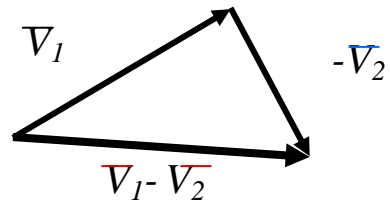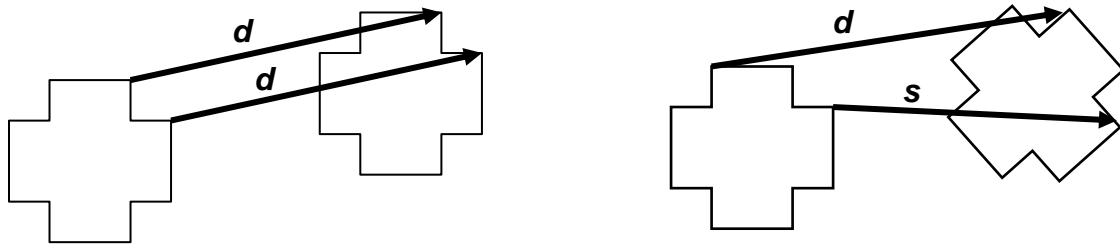
V_1

$-V_2$

$V_1 - V_2$

Figure 1

Translation (left) Translation and Rotation (right)

Figure 2

ONE LEGGED STANCE
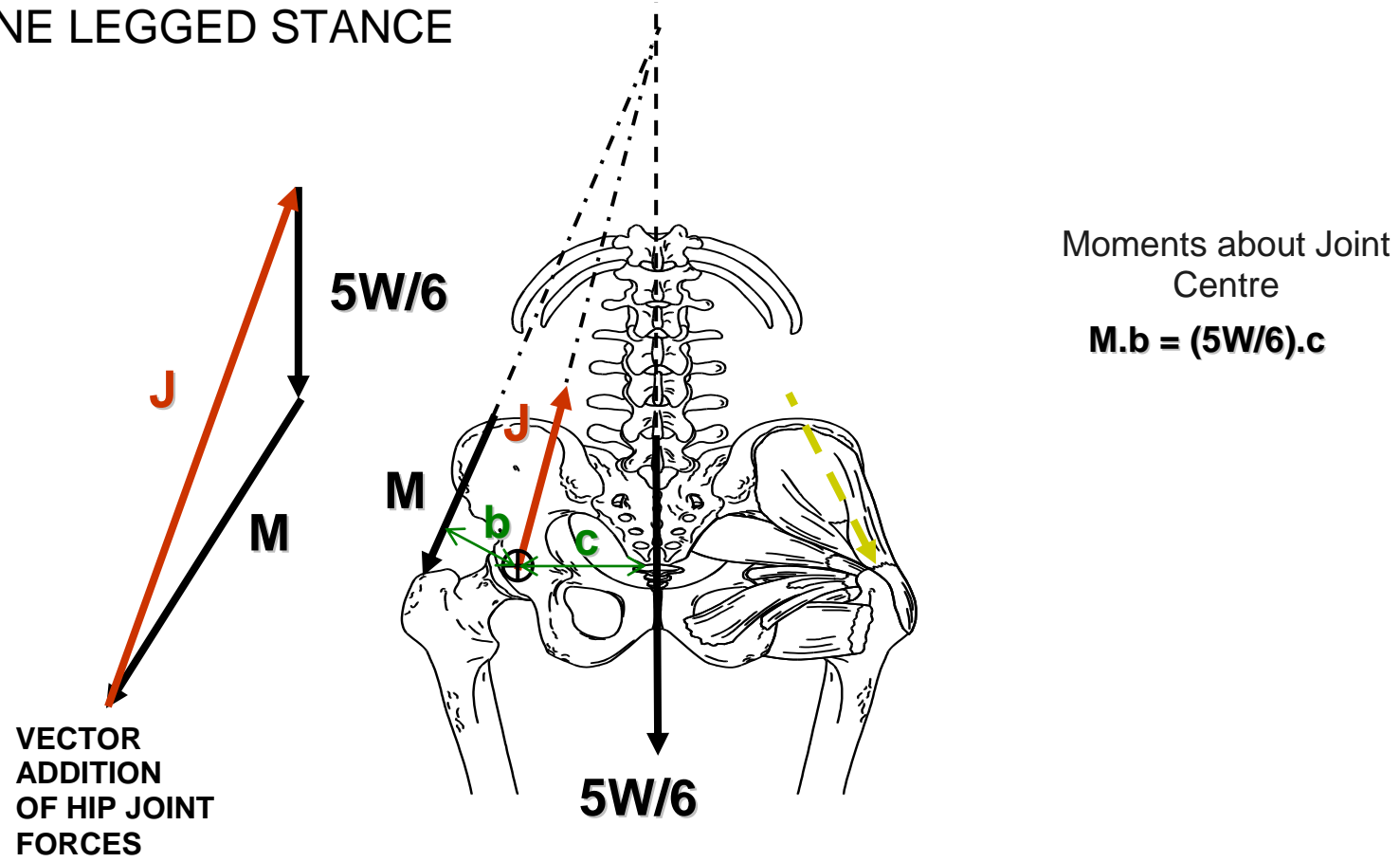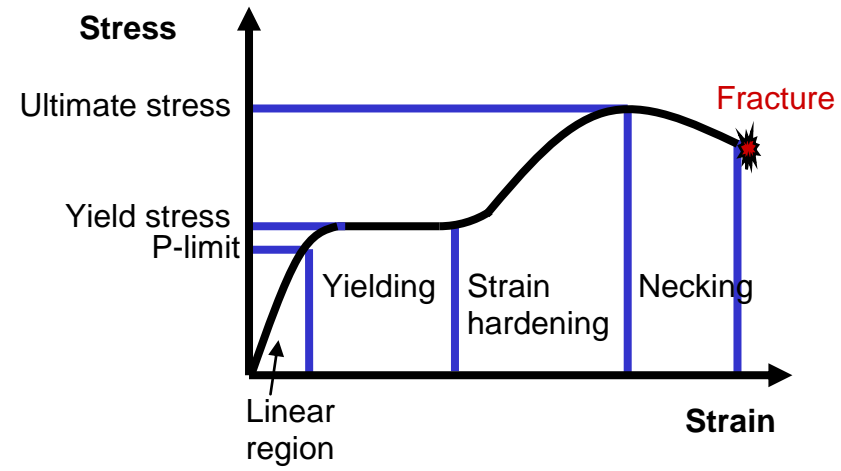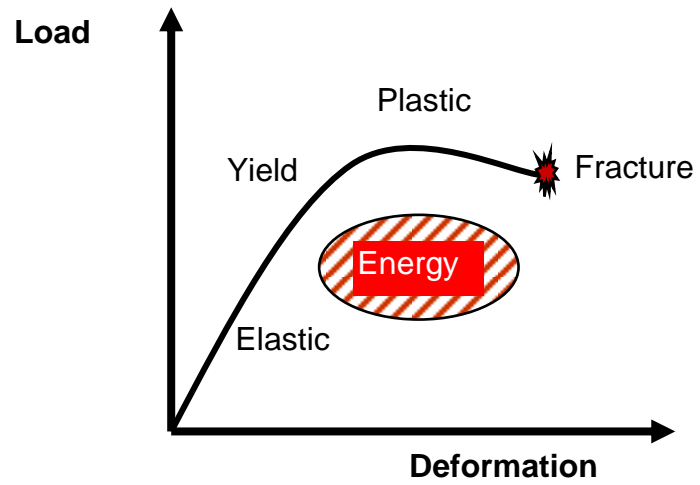
5W/6

J

M

**VECTOR
ADDITION
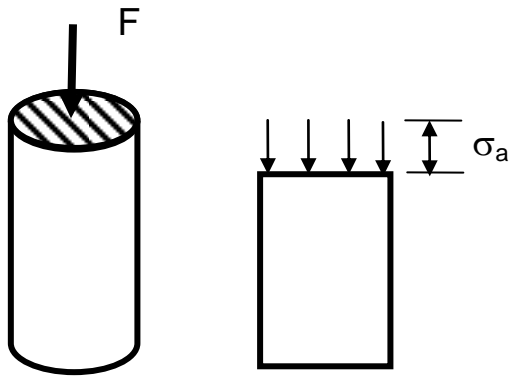OF HIP JOINT
FORCES**

J

M

b

c

5W/6

Moments about Joint Centre

M.b = (5W/6).c

Figure 3

Load

Plastic

Yield

Fracture

Energy

Elastic

Deformation

Stress

Ultimate stress

Fracture

Yield stress
P-limit

Yielding

Strain
hardening

Necking

Linear
region

Strain

Figure 4

AXIAL STRESS

σ_a

Normal Stress σ

Shear Stress τ

STRESSES ON PLANE

BENDING STRESS

F

M

$\sigma +$

$\sigma -$

TORSIONAL STRESS

T

γ

T

τ

Figure 5

h

neutral axis

b

$$I = \frac{b \cdot h^3}{12}$$

d

$J = \pi d^4/32$

d_i

d_o
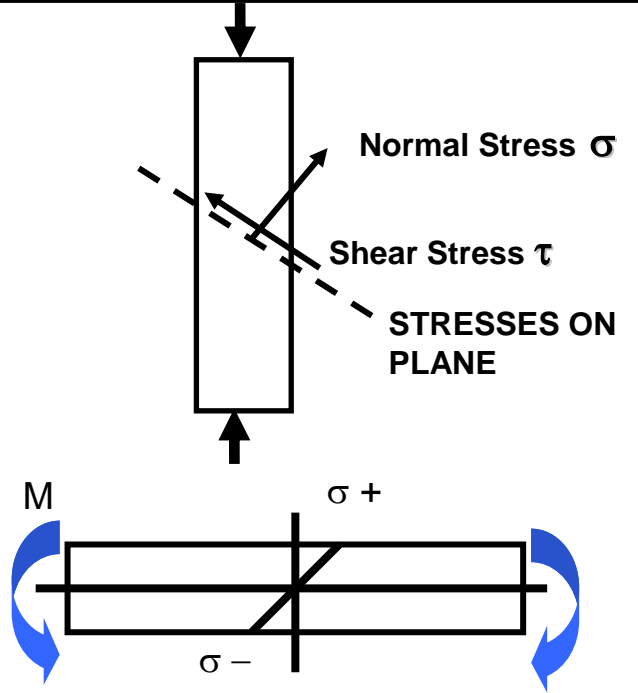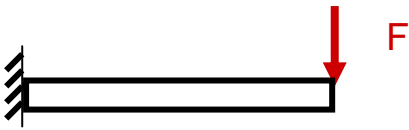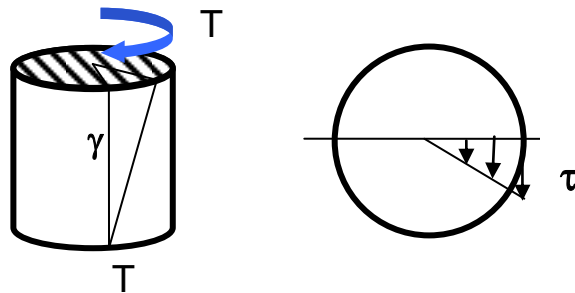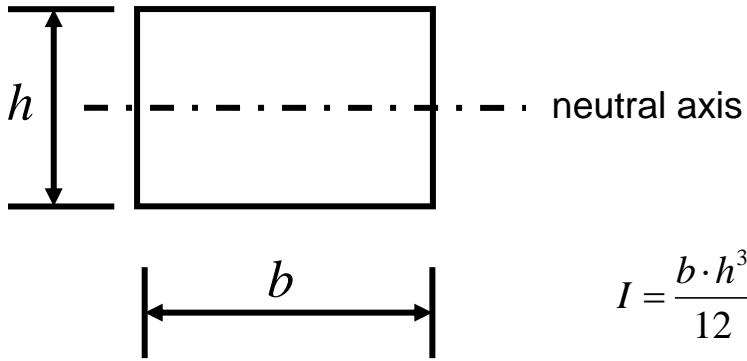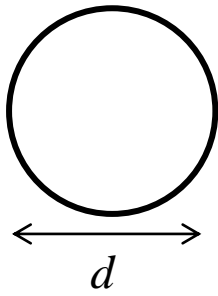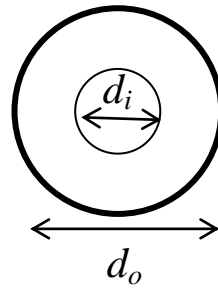
$J = \pi(d_o{}^4 - d_i{}^4)/32$

Figure 6

Figure 7

Figure 8