UNIT I

INTRODUCTION TO MECHANICS

- ✓ **Principles of mechanics**
- ✓ **Mechanics of Motion**
  - **Newton's laws of motion**
  - **Kinetics**
  - **Kinematics**
- ✓ Fluid Mechanics
  - **Euler and Navier-Stock's equation**
- ✓ **Viscoelasticity**
- ✓ **Constitutive Equation**
- ✓ Strain energy function

LIST OF IMPORTANT QUESTIONS

PART A

1. What is Mechanics?
2. Define kinesiology?
3. **What is Biomechanics? (N/D 2015)**
4. **Define vector (N/D 2015)**
5. **What is Viscoelasticity? (N/D 2015)**
6. Define Fluid mechanics.
7. **What is the difference between a uniaxial and triaxial accelerometer?(M/J 2016)**
8. **Define Newton's first law of inertia. (M/J 2016)**
9. Define Newton's second and third laws of motion.
10. Distinguish the difference between Kinetics and kinematics.
11. What is Constitution equation?
12. **What is yield point?(N/D 2016)**
13. **State Newton's 2nd law?(N/D 2016)**
14. **A brass specimen 12 mm in diameter and a length of 45 mm is loaded with a 35 KN force in tension. If the length increases by 0.15 mm, determine elastic modulus. If the diameter decreases by 0.0091mm calculate the Poisson's ratio.(M/J 2017)**
15. **What is Kinematics?(M/J 2017)**
16. **Draw the Maxwell and the Voigt model of Viscoelasticity.(M/J 2017)**

PART B

1. **Explain the principles of mechanics.(N/D 2015)**
2. **State and describe about the Newton's law of motion. (N/D 2015)**
3. Explain in detail fluid mechanism with Euler and Navier- Stoke's equation.
4. Explain Viscoelasticity?

5. Explain Constitution equation in detail.
6. **Describe the instrumentations used for measuring kinematic parameters in biomechanics?(M/J 2016)**
7. **Distinguish between linear, angular and general motion?(M/J 2016)**
8. **Explain the difference between biomechanics and kinesiology?(M/J 2016)**
9. **Elaborate on the applications of biomechanics?(N/D 2016)**
10. **Describe about the tools used for measuring kinetic variables in biomechanics?(N/D 2016)**
11. **What is biomechanics? Explain the different force that acts on the body?(M/J 2017)**
12. **Explain Newton's laws and give necessary examples for each law relating to the physiological system?(M/J 2017)**
13. **Describe the motion of viscous fluid by deriving the Navier-Stoke's equation?(M/J 2017)**

PART A

1. **What is Mechanics?**
    - Mechanics can be defined as that science which describes and predicts effect of force on bodies at rest or in motion.
    - It is divided into 3 parts: Mechanics of rigid bodies, mechanics of deformable bodies and mechanics of fluids.
    - Rigid Mechanics can be further divided in 2 main branches:

Static: It deals with the system of forces acting on bodies at rest. **Dynamic:** It deals with the effect of forces on bodies in motion.

2. Define kinesiology?

- Kinesiology is the scientific study of human or non-human body movement.
- Kinesiology addresses physiological, biomechanical, and psychological mechanisms of movement.
- Applications of kinesiology to human health (i.e. human kinesiology) include biomechanics and orthopedics; strength and conditioning; sport psychology; methods of rehabilitation, such as physical and occupational therapy; and sport and exercise.
- Studies of human and animal motion include measures from motion tracking systems, electrophysiology of muscle and brain activity, various methods for monitoring physiological function, and other behavioral and cognitive research techniques.

2. **What is Biomechanics?(N/D 2015)**
    - Biomechanics is defined as the study of the movement of living things using the science of mechanics. In humans, biomechanics often refers to the study of how the skeletal and musculature systems work under different conditions.
    - Biomechanics provides conceptual and mathematical tools that are necessary for understanding how living things move and how kinesiology (Scientific study of human movement) professionals might improve move mentor make movement safer.

Biomechanics Applications are;

- ✓ Improving human movement
- ✓ Treatment or prevention of injury
- ✓ Maintaining Physical Function
- ✓ Improving Musculoskeletal Health
- ✓ Product Design(athletic shoes, prosthetics)

3. Define vector (N/D 2015)

- Vectors are defined as mathematical expressions possessing magnitude and direction, which add according to the parallelogram law.
- Vectors are represented by arrows.
- Common examples are displacement and velocity.

4. What is Viscoelasticity?(N/D 2015)

- ➢ Viscoelasticity is the property of materials that exhibit both viscous and elastic characteristics when undergoing deformation.
- ➢ Viscous materials, like honey, resist shear flow and strain linearly with time when a stress is applied.

5. Define Fluid mechanics.

Fluid mechanics is the application of the laws of force and motion to fluids, i.e. liquids and gases. There are two branches of fluid mechanics:

- **Fluid Statics** or hydrostatics is the study of fluids at rest.

  The main equation required for this is Newton's second law for non-accelerating bodies, i.e., $\sum \vec{F} = 0$.

- **Fluid Dynamics** is the study of fluids in motion.

  The main equation required for this is Newton's second law for accelerating bodies.

6. What is the difference between a uniaxial and triaxial accelerometer?(M/J 2016)

An accelerometer detects bodily acceleration like Physical activity. Physical activity (PA) is essential in health promotion and disease prevention. PA protects against many diseases such as mental, nutritional, gastroenterological, cardiac, and respiratory diseases.

Uniaxial accelerometer	Triaxial accelerometer
• Measure vertical displacement by recording and storing acceleration in the vertical plane during a specified period of time. • Less precise • Eg: ActiGraph GT1M,Pensacola	• Measure acceleration in three planes. • More precise (Measure movement in 3D) • Eg: Tritrac-R3D, Monrovia, RT3

7. **Define Newton's first law of inertia. (M/J 2016)**

Newton's second law of motion is defined as everybody continues to be in its state of rest or of uniform motion in a straight line unless compelled by some external force to act on it. This law is often called "the law of inertia".

8. **Define Newton's second and third laws of motion.**

Newton's second law of motion is defined as the rate of change of momentum of a body is directly proportional to the applied force and takes place in the direction in which the force acts.

i.e.; $F=ma$

Newton's third law of motion is defined as to for every action, there is always an equal and opposite reaction.

9. **Distinguish the difference between Kinetics and kinematics.**

Kinematics:

- **Kinematics** is the branch of classical mechanics which describes the motion without considering the effect of force that causes of motion. (Ex. displacement, time, velocity, etc.)

- Kinematics would focus more on the jumping strategy, the way the jumper moves and how it contributes to jumping higher etc.
- Kinematics finds applications in study of movement of celestial bodies.
- Kinematics has more mathematical expressions

Kinetics:

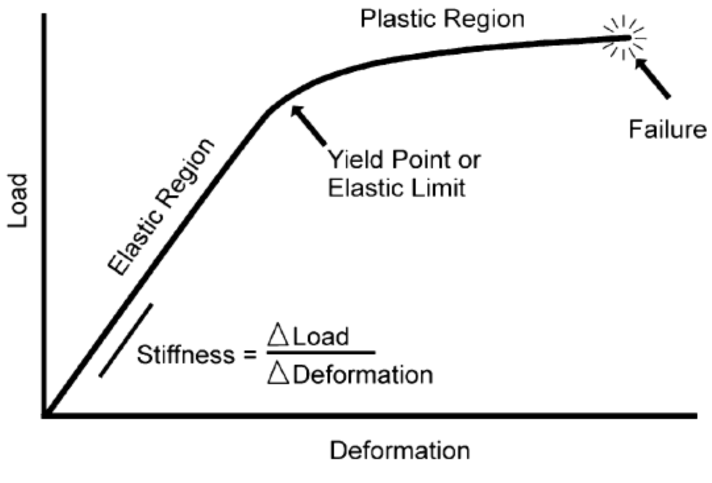
- **Kinetics** is a term for the branch of classical mechanics that is concerned with the relationship between the motion of bodies and its causes, namely forces and torques.
- Kinetics would focus more on the torque production at the joints, GRF (Ground Reaction Force), stiffness etc.
- It has practical applications in designing of automobiles
- Kinetics has more mathematical expressions

10. What is Constitutive equation?

The property of the material is described by **Constitutive equation**. It must be independent of any particular set of coordinates of various physical quantities. Hence a quantitative equation must be a tensor equation. Every term in the equation must be a tensor of the same rank.

11. What is yield point? (N/D 2016)

The yield point or elastic limit is the point on the graph separating the elastic and plastic regions. When the material is deformed beyond the yield point the material will not return to its initial dimensions.



Load–deformation graph of an elastic material.

12. State Newton's 2nd law?(N/D 2016)

Newton's Second Law:

If the resultant force acting on a particle is not zero, the particle will have acceleration proportional to the magnitude of the resultant force and in the direction of this resultant force:

$$F = ma$$

Where F is the resultant force, a is the acceleration and *m* is the mass of the particle. The units of the force are the Newton (N), the units of acceleration are meters per second squared (m/s2), and those of mass are the kilogram (kg).

13. A brass specimen 12 mm in diameter and a length of 45 mm is loaded with a 35 KN force in tension. If the length increases by 0.15 mm, determine elastic modulus. If the diameter decreases by 0.0091mm calculate the Poisson's ratio.(M/J 2017)

Solution:

Given: d = 12 mm, L=45 mm, F= 35KN, +ΔL = 0.15 mm and ΔD= 0.0091mm

Elastic modulus:

σ = F/A

$= F/\pi r^2$

$= 35 \times 10^3 / 3.14 \times 6 \times 10^{-3}$

=185.7Mpa

$E = \sigma/\varepsilon$; $\varepsilon = \frac{\Delta L/L}{\Delta D/D}$

$\frac{\Delta L}{L} = 0.15 \times 10^{-3}/45 \times 10^{-3}$ $\quad$ $\frac{\Delta D}{D} = 0.0091 \times 10^{-3}/12 \times 10^{-3}$

$= 0.003$ (or) 3.4×10^{-3} $\quad$ $= 0.00075$ (or) 0.75×10^{-3}

$\varepsilon = 3.4 \times 10^{-3} / 0.75 \times 10^{-3} = 4.54 \times 10^{-3}$

$E = 185.7 \times 10^6 / 4.54 \times 10^{-3}$

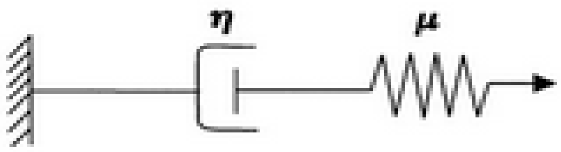
E = 40.90Gpa

14. What is Kinematics?(M/J 2017)

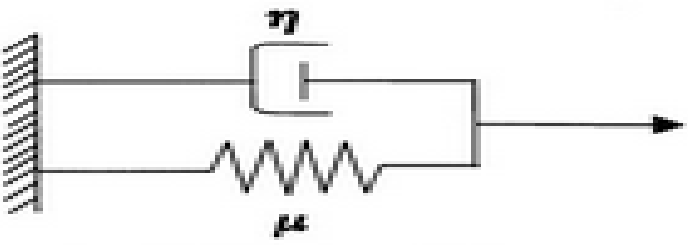
Kinematics:

- **Kinematics** is the branch of classical mechanics which describes the motion without considering the effect of force that causes of motion. (Ex. displacement, time, velocity, etc.)
- Kinematics would focus more on the jumping strategy, the way the jumper moves and how it contributes to jumping higher etc.
- Kinematics finds applications in study of movement of celestial bodies.
- Kinematics has more mathematical expressions

15. Draw the Maxwell and the Voigt model of Viscoelasticity (M/J 2017)



Maxwell model



Voigt model

16. Mention Principal stress equation and Mohr's circle.

Principal stress equation:

- Planes on which the shear stresses are zero are called the principal planes.
- Normal stress on a principal plane is called the principal stress.

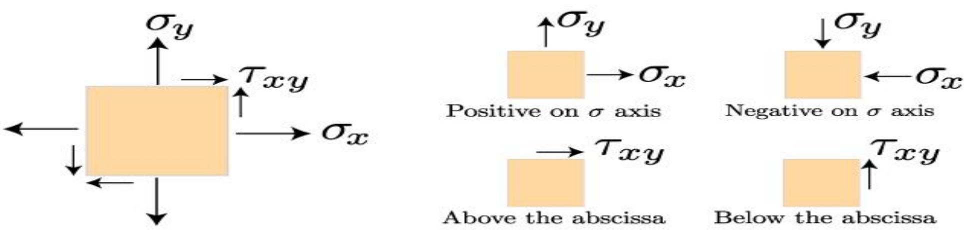
The principal normal stress will occur when the shear stress is zero, which means

$$\sigma_{1,2} = \frac{\sigma_x + \sigma_y}{2} \pm \sqrt{\left(\frac{\sigma_x - \sigma_y}{2}\right)^2 + \tau_{xy}^2}, \quad \tau_{\max} = \sqrt{\left(\frac{\sigma_x - \sigma_y}{2}\right)^2 + \tau_{xy}^2}$$

The principal shear stress is simply the square root term

Mohr's circle.

An alternative to using equations for the principal stresses is to use a graphical method known as Mohr's Circle. This involves creating a graph with sigma as abscissa and tau as ordinate, and plotting the given stress state.



Center is the average normal stress. The radius of that circle is the maximum shear stress.

17. Express fluid mechanics in Euler equation.

For inviscid flow (μ = 0), The Navier-Stokes equations reduce to

$$\rho\left(\frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y} + w\frac{\partial u}{\partial z}\right) = \rho g_x - \frac{\partial p}{\partial x}$$

$$\rho\left(\frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y} + w\frac{\partial v}{\partial z}\right) = \rho g_y - \frac{\partial p}{\partial y}$$

$$\rho\left(\frac{\partial w}{\partial t} + u\frac{\partial w}{\partial x} + v\frac{\partial w}{\partial y} + w\frac{\partial w}{\partial z}\right) = \rho g_z - \frac{\partial p}{\partial z}$$

Euler's equations in vector form as

$$\rho\frac{D\mathbf{V}}{Dt} = \rho\mathbf{g} - \nabla p$$

18. Express fluid mechanics in Navier- Stoke equation.

Navier-Stokes equations are

$$\rho\left(\frac{\partial u}{\partial t}+u\frac{\partial u}{\partial x}+v\frac{\partial u}{\partial y}+w\frac{\partial u}{\partial z}\right)=\rho g_x-\frac{\partial p}{\partial x}+\mu\left(\frac{\partial^2 u}{\partial x^2}+\frac{\partial^2 u}{\partial y^2}+\frac{\partial^2 u}{\partial z^2}\right)$$

$$\rho\left(\frac{\partial v}{\partial t}+u\frac{\partial v}{\partial x}+v\frac{\partial v}{\partial y}+w\frac{\partial v}{\partial z}\right)=\rho g_y-\frac{\partial p}{\partial y}+\mu\left(\frac{\partial^2 v}{\partial x^2}+\frac{\partial^2 v}{\partial y^2}+\frac{\partial^2 v}{\partial z^2}\right)$$

$$\rho\left(\frac{\partial w}{\partial t}+u\frac{\partial w}{\partial x}+v\frac{\partial w}{\partial y}+w\frac{\partial w}{\partial z}\right)=\rho g_z-\frac{\partial p}{\partial z}+\mu\left(\frac{\partial^2 w}{\partial x^2}+\frac{\partial^2 w}{\partial y^2}+\frac{\partial^2 w}{\partial z^2}\right)$$

In single vector form,

$$\rho\frac{D\mathbf{V}}{Dt}=\rho\mathbf{g}-\nabla p+\mu\nabla^2\mathbf{V}$$

19. What is stress transformation and why it is needed?

- Transforming stress components from one coordinate system to another at a given point.
- Stress transformation is to develop the ability to visualize planes passing through a point on which stresses are given or are being found, particularly the planes of maximum normal stress and maximum shear stress.

20. What is strain energy function?

- The elastic response of any solid in tension can be characterized by means of a stored-energy function.
- A hyper elastic material is an ideally elastic material for which the stress-strain relationship derives from a strain energy density function.
- strain energy function can be obtained by the equation ,

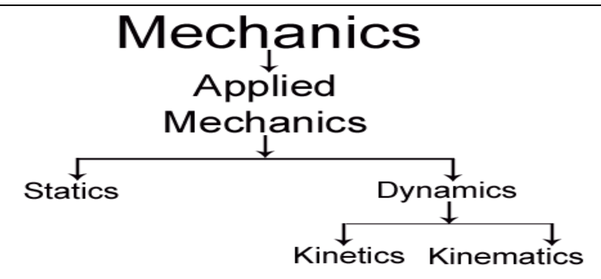
$$\mathbf{S_{ij}}=\frac{\partial(\rho_0 W)}{\partial E_{ij}}$$

S_{ij}- Stress component, ρ_0- Density, W-Strain energy per unit mass, and E_{ij} - strain components

PART B

1. Explain the principles of mechanics.(N/D 2015)

Mechanics is the science which studies effect of forces on the bodies at rest or in motion. The law and principle of mechanics which used or applied in engineering problems in various field, is called as **Applied Mechanics**.



Mechanics can be further divided in 2 main branches one is **'Static'** and other is **'Dynamic'**.

Static: It deals with the system of forces acting on bodies at rest.

Dynamic: It deals with the effect of forces on bodies in motion.

Dynamic is divided in branches Kinetics and kinematics. **Kinetics:** It is the branch of dynamics which studies motion considering mass of body and force causing motion.

Kinematics: It is the branch of dynamics which studies without considering the force of motion

Fundamental Concept of Mechanics:

- Space
- Mass
- Time
- Force

Space: It is concern with position of a body to fix the position of point.

Mass: To distinguish behaviour of the two bodies under the action of an identical force.

Time: This term mostly used in dynamics to relate the sequence of events.

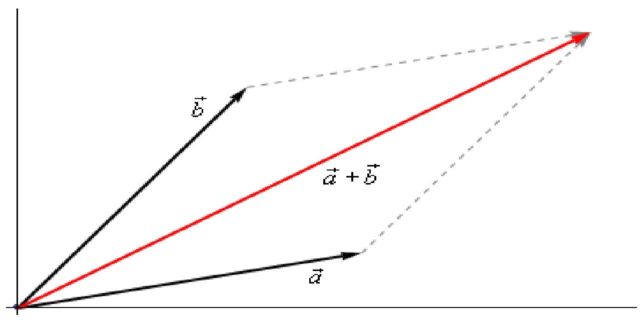
Force: It is an essential agency which changes or tends to change the state of rest or uniform motion of body.

FUNDAMENTAL PRINCIPLES OF APPLIED MECHANICS:

1. Law of parallelogram of force
2. Law of transmissibility
3. Newton's first law of motion
4. Newton's Second Law
5. Newton's Third Law
6. Newton's Law of Gravitation

1. Law of parallelogram of force

If two force vectors "a" and "b" acting on a body and replace these two vectors with a single vector that has the same effect on the body. Here we can use the parallelogram law for the addition of force vectors.



The resultant force (or the force that can replace the two vectors and still have the same effect of the body as the original two) is the diagonal of the parallelogram (vector "a+b"). Measure vector "a+b" to find its magnitude.

2. Law of transmissibility

"The condition of equilibrium or motion of a rigid body will not be changed if a force acting at a given point of the rigid body is replaced by another force of the same magnitude and same direction acting on a different point along the same line of action Or, "The action of a force may be transmitted along its line of action." Line of action of force: The line of action is an infinite straight line along which the force acts.

Here, force’s F and F' have the same magnitude and they act along the same line of action. Thus, the two forces are equivalent. "While the principle of transmissibility may be used freely to determine the conditions of motion of equilibrium of rigid bodies and to compute the external forces acting on these bodies, it should be avoided, or at least used with care, in determining internal forces and deformations."

3. Newton’s first law of motion

Every particle continues in a state of rest or uniform in a straight line unless it is compelled to change that state by forces impressed on it.

4. Newton’s Second Law

The rate change of linear momentum of an object is directly proportional to the external force on the object." The concept behind this is forces arise due to interaction between the bodies.

i.e; $F=ma$

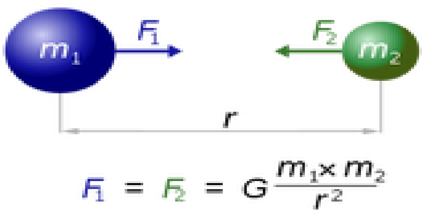
5. Newton'sThird Law

To every action there is an equal and opposite reaction. This law refers to a system of particles or bodies in contact with each other and interacting.

6. Newton's Law of Gravitation

"Two particles are attracted towards each other along the line connecting them with a force whose magnitude is directly proportional to the product of the masses and inversely proportional to the square of the distance between the particles."

$$F = G\frac{m_1 m_2}{r^2}$$



Newton's Law of Gravitation introduces the idea of an action exerted at a distance and extends the range of application of Newton's Third Law: the action F and the reaction -F are equal and opposite, and they have the same line of action.

2. State and describe about the Newton's law of motion. (N/D 2015)

Sir Isaac Newton's three laws of motion describe the motion of massive bodies and how they interact.

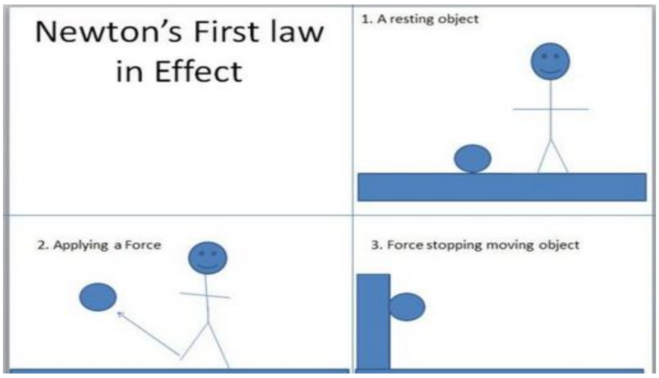
- **Newton's first law of motion**

Everybody continues to be in its state of rest or of uniform motion in a straight line unless compelled by some external force to act otherwise.

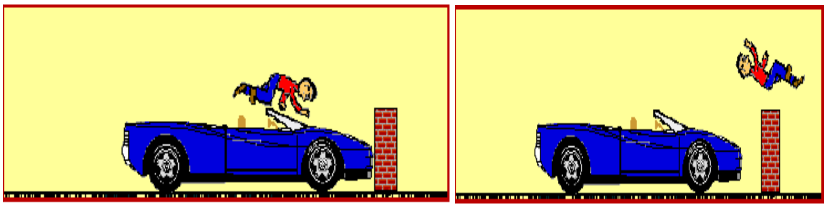
This means that there is a natural tendency of objects to keep on doing what they're doing. All objects resist changes in their state of motion. In the absence of an unbalanced force, an object in motion will maintain this state of motion.

Examples:

❖ If you kicked a ball in space, it would keep going forever, because there is no gravity. It will stop by an external force like hitting on the wall.



❖ If you are driving in your car at a very high speed and hit something, like a brick wall ora tree, the car will come to an instant stop, but you will keep moving forward. This is why cars have airbags, to protect you from smashing into the windscreen.

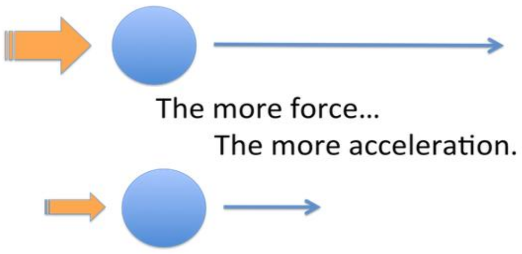


➢ **Newton's second law of motion**

The rate of change of momentum of a body is directly proportional to the applied force and takes place in the direction in which the force acts.

Everyone knows that heavier objects require more force to move the same distance as lighter objects.

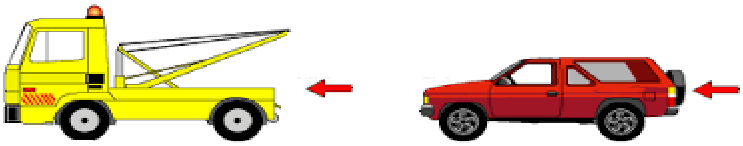
i.e.; **F=ma**



Examples:

❖ It is easier to push an empty shopping cart than a full one, because the full shopping cart has more mass than empty one. This means that more force is required to push the full shopping cart.

❖ If you use the same force to push a truck and push a car, and the car will have more acceleration than the truck, because the car has less mass.



➢ **Newton's third law of motion**

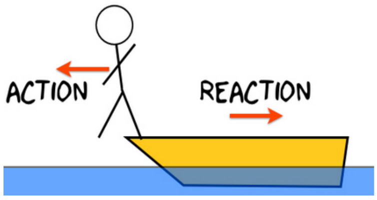
To every action, there is always an equal and opposite reaction.

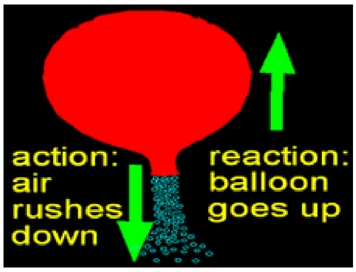
This means that for every force there is a reaction force that is equal in size, but opposite in direction. That is to say that whenever an object pushes another object it gets pushed back in the opposite direction equally hard.

In equation form, $F_{12}=-F_{21}$, where F_{12} is the force acting on object 1 by object 2 and F_{21} is the force on object 2 by object 1.

Examples:

❖ When you jump off a small rowing boat into water, you will push yourself forwards the water. The same force you used to push forward will make the boat move backwards.





❖ The air rushes out a balloon, the opposite reaction is that the balloon flies up.

3. Explain in detail fluid mechanism with Euler and Navier- Stoke equation.

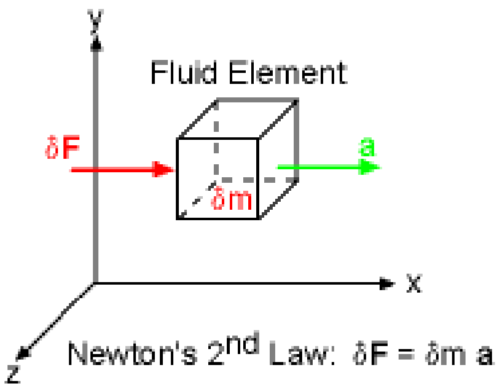
Fluid mechanics is the branch of physics that studies the mechanics of fluids (liquids, gases, and plasmas) and the forces on them. Fluid mechanics can be divided into **fluid statics**, the study of fluids at rest; and **fluid dynamics**, the study of the effect of forces on fluid motion.

Fluid mechanism is described by Navier-Stoke and Euler equation.

Navier-Stokes Equations:

The differential form of the linear momentum equation (also known as the Navier-Stokes equations).Newton's second law on a differential fluid element is

$$\delta \mathbf{F} = \delta m \, \mathbf{a}$$



where $\delta\mathbf{F}$ is the resultant force acting on the fluid element (mass = δm). $\mathbf{a}$ is the acceleration of the fluid element, and it is given by

$$\mathbf{a} = \frac{D\mathbf{V}}{Dt} = \frac{\partial \mathbf{V}}{\partial t} + (\mathbf{V} \cdot \nabla)\mathbf{V}$$

Expanding into its Cartesian components yields

$$a_x = \frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y} + w\frac{\partial u}{\partial z}$$
$$a_y = \frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y} + w\frac{\partial v}{\partial z}$$
$$a_z = \frac{\partial w}{\partial t} + u\frac{\partial w}{\partial x} + v\frac{\partial w}{\partial y} + w\frac{\partial w}{\partial z}$$

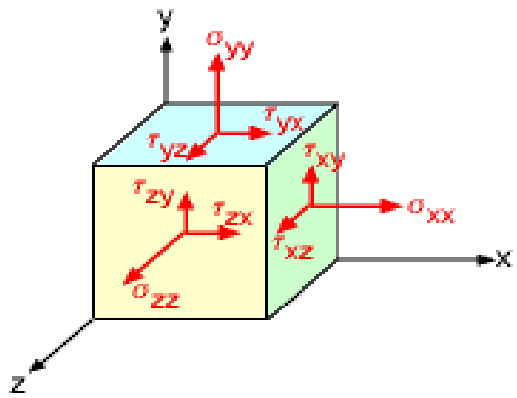
There are two types of forces acting on the fluid element: body force ($\delta\mathbf{F_B}$) and surface force ($\delta\mathbf{F_S}$).

$$\delta\mathbf{F} = \delta\mathbf{F_B} + \delta\mathbf{F_S}$$

The only body force considered here is the weight of the fluid element. That is,

$$\delta\mathbf{F_B} = \delta m\,\mathbf{g} = \delta m\,(g_x\,\mathbf{i} + g_y\,\mathbf{j} + g_z\,\mathbf{k})$$

The surface forces are due to the stresses exerted on the sides of the fluid element. There are two types of stresses applied on the surface: normal stress (σ_{ij}) and shear stress (τ_{ij}). Normal stress acts perpendicular to the surface while shear stress is tangential to the surface. The subscript i refers to the axis normal to the surface, and the subscript j represents the direction of the stress



A summation of the surface forces in the x-direction yields

$$\delta F_{Sx} = \left[\frac{\partial \sigma_{xx}}{\partial x} + \frac{\partial \tau_{yx}}{\partial y} + \frac{\partial \tau_{zx}}{\partial z}\right]\delta x\,\delta y\,\delta z$$

Note that the stresses are multiplied by the area to obtain the surface forces. Similarly, the total surface forces in the y- and z-directions are obtained as

$$\delta F_{Sz} = \left[\frac{\partial \sigma_{zz}}{\partial z} + \frac{\partial \tau_{xz}}{\partial x} + \frac{\partial \tau_{yz}}{\partial y}\right]\delta x\,\delta y\,\delta z$$

$$\delta F_{Sy} = \left[\frac{\partial \sigma_{yy}}{\partial y} + \frac{\partial \tau_{xy}}{\partial x} + \frac{\partial \tau_{zy}}{\partial z}\right]\delta x\,\delta y\,\delta z$$

The resultant surface force is then given as

$$\delta \mathbf{F_S} = \delta F_{Sx}\,\mathbf{i} + \delta F_{Sy}\,\mathbf{j} + \delta F_{Sz}\,\mathbf{k}$$

The mass of the fluid element can be expressed in terms of its volume and fluid density

($\delta m = \rho\,\delta x \delta y \delta z$), so that the linear momentum equation in Cartesian coordinates reduces to

$$\rho g_x + \frac{\partial \sigma_{xx}}{\partial x} + \frac{\partial \tau_{yx}}{\partial y} + \frac{\partial \tau_{zx}}{\partial z} = \rho\left(\frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y} + w\frac{\partial u}{\partial z}\right)$$

$$\rho g_y + \frac{\partial \sigma_{yy}}{\partial y} + \frac{\partial \tau_{xy}}{\partial x} + \frac{\partial \tau_{zy}}{\partial z} = \rho\left(\frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y} + w\frac{\partial v}{\partial z}\right)$$

$$\rho g_z + \frac{\partial \sigma_{zz}}{\partial z} + \frac{\partial \tau_{xz}}{\partial x} + \frac{\partial \tau_{yz}}{\partial y} = \rho\left(\frac{\partial w}{\partial t} + u\frac{\partial w}{\partial x} + v\frac{\partial w}{\partial y} + w\frac{\partial w}{\partial z}\right)$$

For Newtonian fluids shear stress field is symmetric, and it is related to the rate of shear strain in a linear fashion.

$$\tau_{xy} = \tau_{yx} = \mu\left(\frac{\partial v}{\partial x} + \frac{\partial u}{\partial y}\right)$$

$$\tau_{yz} = \tau_{zy} = \mu\left(\frac{\partial w}{\partial y} + \frac{\partial v}{\partial z}\right)$$

$$\sigma_{xx} = -p - \frac{2}{3}\mu \nabla \cdot \mathbf{V} + 2\mu\frac{\partial u}{\partial x}$$

$$\sigma_{yy} = -p - \frac{2}{3}\mu\nabla\cdot\mathbf{V} + 2\mu\frac{\partial v}{\partial y}$$

$$\sigma_{zz} = -p - \frac{2}{3}\mu\nabla\cdot\mathbf{V} + 2\mu\frac{\partial w}{\partial z}$$

where μ is the viscosity of the fluid. Pressure term, p, only acts normal to the surface for each element face. For incompressible flow, the term $\nabla\cdot\mathbf{V}$ is zero based on the continuity equation. The linear momentum equations thus become

$$\rho\left(\frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y} + w\frac{\partial u}{\partial z}\right) = \rho g_x - \frac{\partial p}{\partial x} + \mu\left(\frac{\partial^2 u}{\partial x^2} + \frac{\partial^2 u}{\partial y^2} + \frac{\partial^2 u}{\partial z^2}\right)$$
$$\rho\left(\frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y} + w\frac{\partial v}{\partial z}\right) = \rho g_y - \frac{\partial p}{\partial y} + \mu\left(\frac{\partial^2 v}{\partial x^2} + \frac{\partial^2 v}{\partial y^2} + \frac{\partial^2 v}{\partial z^2}\right)$$
$$\rho\left(\frac{\partial w}{\partial t} + u\frac{\partial w}{\partial x} + v\frac{\partial w}{\partial y} + w\frac{\partial w}{\partial z}\right) = \rho g_z - \frac{\partial p}{\partial z} + \mu\left(\frac{\partial^2 w}{\partial x^2} + \frac{\partial^2 w}{\partial y^2} + \frac{\partial^2 w}{\partial z^2}\right)$$

The above equations are generally referred to as the Navier-Stokes equations.

In single vector form:

$$\rho\frac{D\mathbf{V}}{Dt} = \rho\mathbf{g} - \nabla p + \mu\nabla^2\mathbf{V}$$

Euler's Equations

For inviscid flow (μ = 0), the Navier-Stokes equations reduce to

$$\rho\left(\frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y} + w\frac{\partial u}{\partial z}\right) = \rho g_x - \frac{\partial p}{\partial x}$$
$$\rho\left(\frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y} + w\frac{\partial v}{\partial z}\right) = \rho g_y - \frac{\partial p}{\partial y}$$
$$\rho\left(\frac{\partial w}{\partial t} + u\frac{\partial w}{\partial x} + v\frac{\partial w}{\partial y} + w\frac{\partial w}{\partial z}\right) = \rho g_z - \frac{\partial p}{\partial z}$$

The above equations are known as Euler's equations

Euler's equations can be written in vector form as

$$\rho \frac{D\mathbf{V}}{Dt} = \rho \mathbf{g} - \nabla p$$

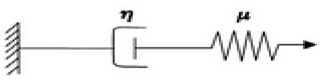
Euler's equations can be simplified further to obtain Bernoulli's equation, which is applicable to steady, incompressible, in viscid flow along a streamline.

4. Explain Viscoelasticity?

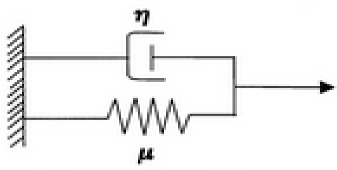
- Viscoelasticity is the property of material that exhibit both viscous and elastic characteristics when undergoing deformation.
- When a body is suddenly strained and then strain is maintained constant afterward, the corresponding stresses induced in the body decrease with time. This phenomenon is called **Stress relaxation** or relaxation for short.
- If the body is suddenly stressed and then the stress is maintained constant afterwards, the body continues to deform, and the phenomenon is called **creep.**
- If the body is subjected to cyclic loading, the stress- strain relationship in the loading process is usually somewhat different from that in the unloading process and the phenomenon is called **hysteresis.**
- The features of hysteresis, relaxation and creep are found in many materials. Collectivity, they are called features of viscoelasticity.
- Mechanical models are often used to discuss the viscoelastic behavior of materials.
- Three mechanical models of material behavior namely, the **Maxwell model, the voigt model and the Kelvin model (Standard linear solid)** all of which are composed of combination of linear spring with spring constant $\boldsymbol{\mu}$ and dashpots with coefficient of viscosity $\boldsymbol{\eta}$.
- A linear spring is supposed to produce instantaneously deformation proportional to the load.
- A dash pot is supposed to produce a velocity proportional to the load at any instant.
- Thus if F is the force acting in a spring and u is its extension, then

$$F=\mu u.$$

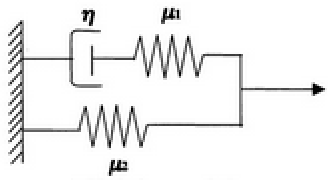
- If the force F acts on a dashpot, it will produce a velocity of deflection $\dot{u}$ · and F=$\eta\dot{u}$ ·



Maxwell model



Voigt model



Kelvin model

- In **Maxwell** model, the force F is acting and transmitted from the spring to the dashpot, and produces a displacement F/μ in the spring and a velocity F/ η in the dashpot. The velocity of the spring extension is$\dot{F}/\mu$. (Dot indicates the differentiation).
- The total velocity is the sum of these two.

$$\dot{u} = \dot{F}/\mu + F/\eta \text{ ---- } (1)$$

- For the **Voigt model**, the spring and dashpot have the same displacement. If the displacement is u, the velocity is $\dot{u}$ and the spring and the dashpot will produce forces μu and η $\dot{u}$ respectively.
- The total force ,

$$F = \mu u + \eta\dot{u} \text{ ---- } (2)$$

- For the **Kelvin model**, break down the displacement u into u1 of the dashpot and u1' of the spring.

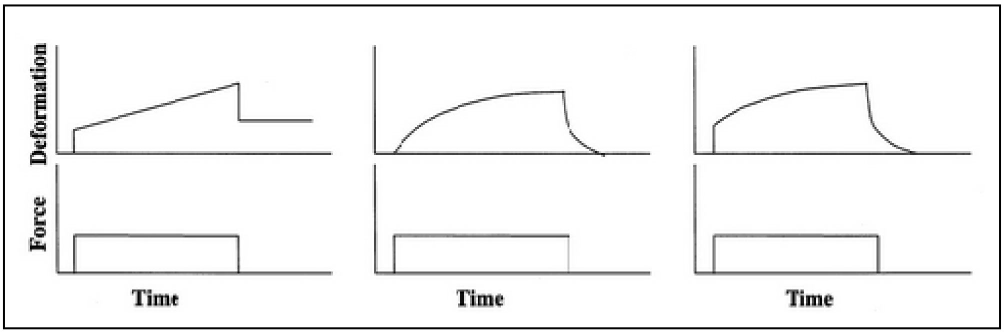
The total force, $$F + \tau_{\epsilon}\dot{F} = E_R\,(U + \tau_{\sigma}\dot{u}) \text{ ---- } (3)$$

τ_ϵ- Relaxation time for constant strain

τ_σ- Relaxation time for constant stress

If we solve equation (1), (2) and (3) for u (t), when F (t) is a unit step function, thus the results are called creep function.

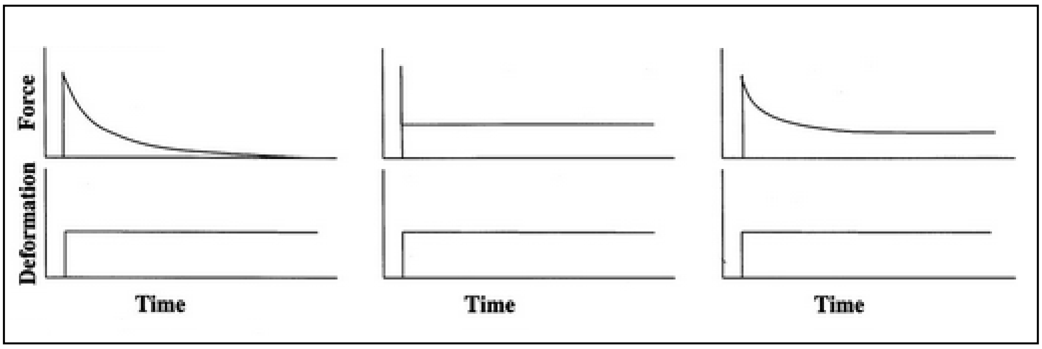
Creep function



Maxwell model **Voigt model** **Kelvin model**

- A body that obeys a load – deflection relation like Maxwell's model is said to Maxwell solid.
- Interchanging the roles of F and u, we obtain the relaxation function.

Relaxation function



Maxwell model **Voigt model** **Kelvin model**

5. Explain constitutive equation

A constitutive equation is described by the physical property of material. It must be independent of any particular set of coordinates of various physical quantities. Hence a quantitative equation must be a tensor equation. Every term in the equation must be a tensor of the same rank.

The constitutive equations are of;

-Linear elasticity

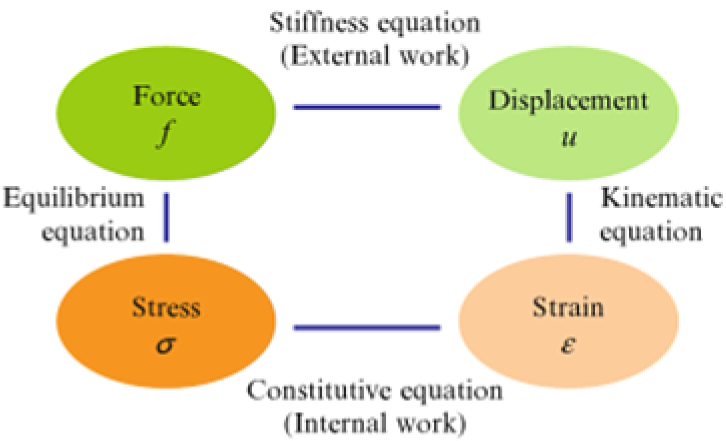
-Viscous fluids

-Linear Viscoelasticity

-Plasticity

1. Constitutive Equation: Linear Elasticity

The stress componentsσij are connected to force components by equilibrium equations and the strain components are related to the displacement components through kinematic equations. In order to explore the external work by force and displacement in a solid in terms of stress and strain, we need the third equations connecting stress and strain, that is, the constitutive equations. This equation gives the framework for the internal work.



In one dimensional case, the stress is described as a linear function of strain for some cases or as a nonlinear function of strain for other cases. Therefore the general expression will be

$$\sigma ij = \sigma ij\ (\varepsilon_{kl})$$

Connecting the stress componentsσij and the small strain componentsε_{kl}. When the stress is linearly related to strain, the constitutive equation is reduced to

$$\boldsymbol{\sigma ij} = \boldsymbol{C_{ijkl}\varepsilon_{kl}}$$

in tensor form or

$$\boldsymbol{\sigma} = \boldsymbol{C\ \varepsilon}$$

in matrix form

This is the three-dimensional expression of Hooke's law.

2. Constitutive Equation: Newtonian viscous fluids

$$T_{ij} = -p(\rho, \theta)\delta_{ij} + B_{ijkl}(\rho, \theta)D_{kl}.$$

- **For a fluid at rest,**

$$T_{ij} = -p(\rho, \theta)\delta_{ij}.$$

- **For an ideal fluid,**

$$T_{ij} = -p\delta_{ij}, \qquad \mathbf{T} = -p\mathbf{I}.$$

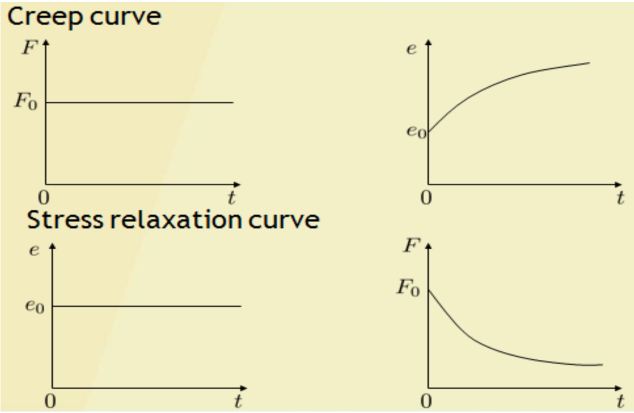
3. Constitutive Equation: Linear Viscoelasticity

- Assuming the superposition principle, then

  $T_{ij}(t) = \int_{-\infty}^{t} G_{ijkl}(t-\tau)\frac{\mathrm{d}E_{kl}(\tau)}{\mathrm{d}\tau}d\tau$, $G_{ijkl}(t-\tau)$ are the stress relaxation function.

- The inverse relation is

  $E_{ij}(\tau) = \int_{-\infty}^{t} J_{ijkl}(t-\tau)\frac{\mathrm{d}T_{kl}}{\mathrm{d}\tau}\mathrm{d}\tau.$, $J_{ijkl}(t-\tau)$ are the creep function.



4. Constitutive Equation: Plasticity

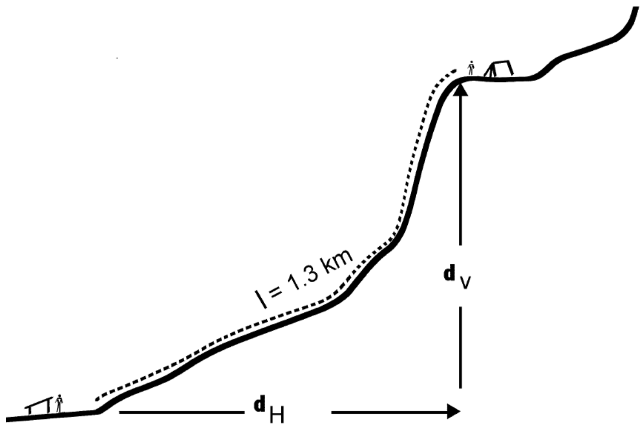
Plastic stress- strain relations are,

$$D_{ij} = D_{ij}^{(e)} + D_{ij}^{(p)}$$

6. **Describe the instrumentations used for measuring kinematic parameters in biomechanics?(M/J 2016)**

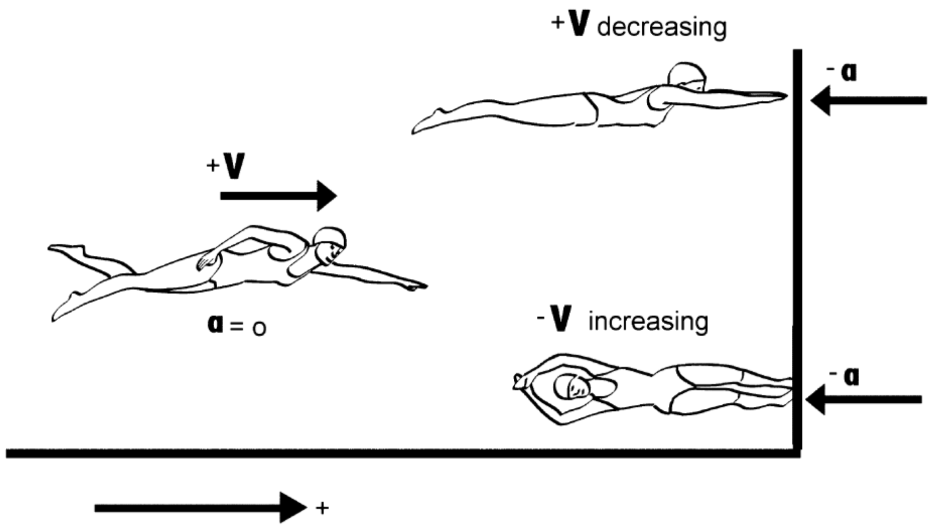
Linear motion

- Motion is change in position with respect to some frame of reference.
- In mathematical terms, linear motion is simple to define: final position minus initial position. The simplest linear motion variable is a scalar called distance (l).
- **Example:** an outdoor adventurer leaves base camp and climbs for 4 hours through rough terrain along the path.If her final position traced a 1.3-km climb measured relative to the base camp (0 km) with a pedometer, the distance she climbed was 1.3 km (final position – initial position).

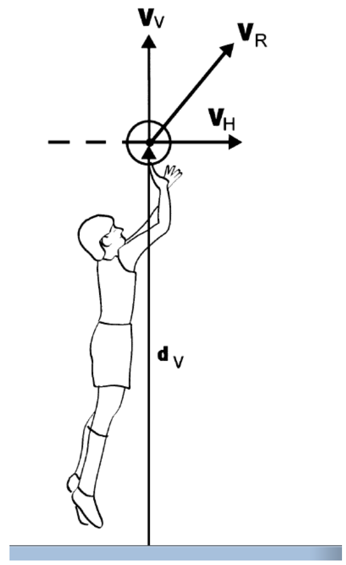


- The corresponding vector quantity to distance is displacement (d).
- Linear displacements are usually defined relative to right-angle directions, which are convenient for the purpose of the analysis.
- **Degrees of freedom** represent the kinematic complexity of a biomechanical model. The degrees of freedom (dof) correspond to the number of kinematic measurements needed to completely describe the position of an object.

- **Frame of reference:** It suggests from where we have to measure or observe the motion. Reference frames in biomechanics are either absolute or relative.
- **Speed:** Speed is how fast an object is moving without regard to direction. Speed is a scalar quantity like distance, and most people have an accurate intuitive understanding of speed.
- Speed(s) is defined as the rate of change of distance (s = *l*/t), so typical units are m/s, ft/s, km/hr, or miles/hr.
- **Example:** If you went jogging across town (5 miles) and arrived at the turn-around point in 30 minutes, your average speed would be (5 miles / 0.5 hours), or an average speed of 10 miles per hour.
- **Velocity:** Velocity is the vector corresponding to speed. Velocity is essentially the speed of an object, in a particular direction.
- Velocity is the rate of change of displacement (V = d/t), so its units are the same as speed, and are usually qualified by a directional adjective (i.e., horizontal, vertical, resultant).
- Let's assume that $\mathbf{d}_E$ = 8 m and $\mathbf{d}_W$= 2 m and that the time it took this student to change stations was 10 seconds. The average velocity along the water axis would be $\mathbf{V}_W = \mathbf{d}_W/t = 2/10 = 0.2$ m/s.
- **Acceleration:** The second derivative with respect to time, or the rate of change of velocity, is acceleration.
- **Example:** Imagine a person is swimming laps. Motion to the right is designated positive, and the swimmer has a relatively constant velocity (zero horizontal acceleration) in the middle of the pool and as she approaches the wall.
- As her hand touches the wall there is a negative acceleration that first slows her down and then speeds her up in the negative direction to begin swimming again. Thinking of the acceleration at the wall as a push in the negative direction is correct throughout the turn.
- As the swimmer touches the other wall there is a positive acceleration that decreases her negative velocity, and if she keeps pushing (hasn't had enough exercise) will increase her velocity in the positive direction back into the pool.

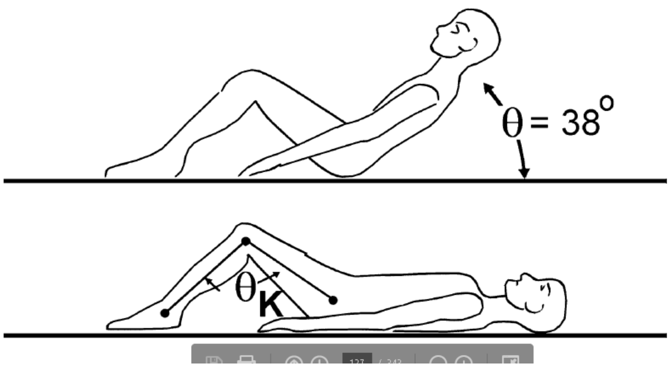


- **Uniformly accelerated motion:** In rare instances the forces acting on an object are constant and therefore create a constant acceleration in the direction of the resultant force.
- When an object is thrown or kicked without significant air resistance in the vertical direction, the path or **trajectory** will be some form of a parabola.
- **Optimal projection principle:** The Optimal Projection Principle refers to the angle(s) that an object is projected to achieve a particular goal.
- **Example:** If a ball was kicked and then landed at the same height, and the air resistance was negligible, the optimal angle of projection for producing maximum horizontal displacement would be 45º.



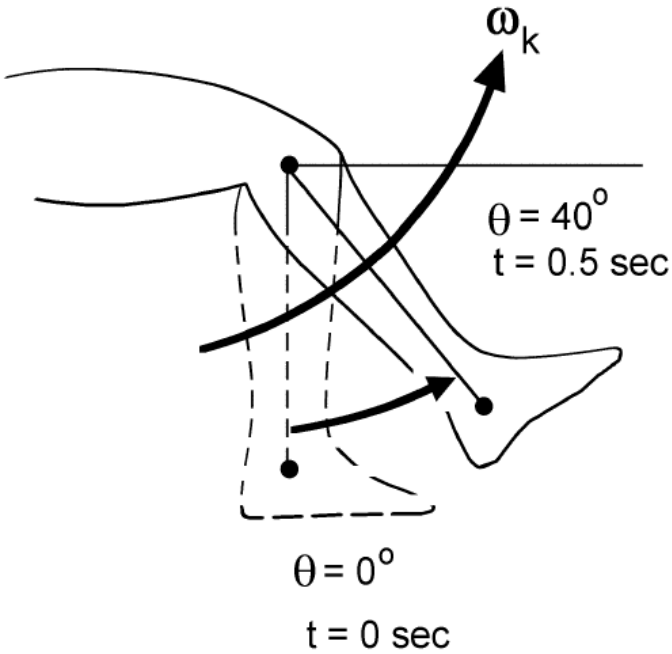
Angular motion

- **Angular kinematics** is the description of angular motion. Angular kinematics is particularly appropriate for the study of human movement because the motion of most human joints can be described using one, two, or three rotations.
- **Angular displacement** (**θ**: theta) is the vector quantity representing the
- change in angular position of an object. Angular displacements are measured in degrees, radians (dimensionless unit equal to 57.3º), and revolutions (360º).
- Angular displacement measured with a **goniometer** is one way to measure **static flexibility**.
- In analyzing the curl-up exercise, the angle between the thoracic spine and the floor is often used. This exercise is usually limited to the first 30 to 40º above the horizontal to limit the involvement of the hip flexors
- The angular displacement of the thoracic spine in the eccentric phase would be –38º (final angle minus initial angle: 0 – 38 = –38º).

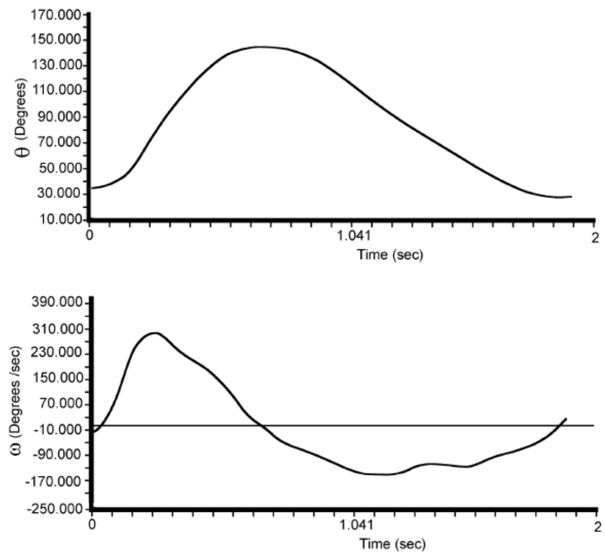


- **Absolute angle** also called trunk angle. It is measured relative to an "unmoving" earth frame of reference.
- **Relative angles** are defined between two segments that can both move. Examples of relative angles in biomechanics are joint angles.
- **Angular velocity(ω: omega)**is the rate of change of angular position and is usually expressed in degrees per second or radians per second. The formula for angular velocity is $\omega = \theta/t$.
- If we measure the angle of the lower leg from the vertical, the exerciser has moved their leg 40º in a 0.5-second period of time. The average knee extension angular velocity can be calculated as follows: $\omega_K = \theta/t = 40/0.5 = 80$

deg/s. The angular velocity is positive because the rotation is counterclockwise.



- **Angular Acceleration:** The rate of change of angular velocity is **angular acceleration** (α= ω/t). The typical units of angular acceleration are deg/s/s and radians/s/s.
- **Example:** the angular displacement and angular velocity of a simple elbow extension and flexion movement in the sagittal plane.



- **Coordination continuum principle:** Coordination is commonly defined as the sequence and timing of body actions used in a movement.
- A person lifting a heavy box simultaneously extends the hips, knees, and ankle.

7. **Distinguish between linear, angular and general motion?(M/J 2016)**

Linear motion

- Motion is change in position with respect to some frame of reference.
- In mathematical terms, linear motion is simple to define: final position minus initial position. The simplest linear motion variable is a scalar called distance (l).
- The corresponding vector quantity to distance is displacement (d).
- Linear displacements are usually defined relative to right-angle directions, which are convenient for the purpose of the analysis.
- **Speed:** Speed is how fast an object is moving without regard to direction. Speed is a scalar quantity like distance, and most people have an accurate intuitive understanding of speed.
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- **Uniformly accelerated motion:** In rare instances the forces acting on an object are constant and therefore create a constant acceleration in the direction of the resultant force.
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- **Angular velocity(ω: omega)**is the rate of change of angular position and is usually expressed in degrees per second or radians per second. The formula for angular velocity is $\omega = \theta/t$.
- **Angular Acceleration:** The rate of change of angular velocity is **angular acceleration** ($\alpha = \omega/t$). The typical units of angular acceleration are deg/s/s and radians/s/s.
- **Coordination continuum principle:** Coordination is commonly defined as the sequence and timing of body actions used in a movement.

General motion

- In physics, motion is a change in position of an object over time. Motion is described in terms of displacement, distance, velocity, acceleration, time and speed.
- Motion of a body is observed by attaching a frame of reference to an observer and measuring the change in position of the body relative to that frame.
- If the position of a body is not changing with respect to a given frame of reference, the body is said to be *at rest*, *motionless*, *immobile*, *stationary*, or to have constant (time-invariant) position.

- An object's motion cannot change unless it is acted upon by a force, as described.
- Motion applies to objects, bodies, and matter particles, to radiation, radiation fields and radiation particles, and to space, its curvature and space-time.

8. Explain the difference between biomechanics and kinesiology?(M/J 2016)

Biomechanics	Kinesiology
• Study of human movements. • It provides key information on the most effective and safest movement patterns, equipment, and relevant exercises to improve human movement. • The applications of biomechanics to human movement can be classified into two main areas: the improvement of performance and the reduction or treatment of injury	• Study of motion and its causes in living things. • It solve human movement problems every day, and one of their most important tools is biomechanics. • the typical scientific subdisciplines of kinesiology are biomechanics, exercise physiology, motor development, motor learning, pedagogy, psycho-social. • Within kinesiology, many biomechanistshave been interested in the application of biomechanics to sport and exercise.

9. Elaborate on the applications of biomechanics?(N/D 2016)

- **Biomechanics** has been defined as the study of the movement of living things using the science of mechanics. The applications of biomechanics to human movement can be classified into two main areas:

the improvement of performance and

the reduction or treatment of injury.

Improving Performance

- Human movement performance can be enhanced many ways. Effective movement involves anatomical factors, neuromuscular skills, physiological capacities, and psychological/ cognitive abilities.
- **Example:** Running
- Human performance can also be enhanced by improvements in the design of equipment. Many of these improvements are related to new materials and engineering designs.
- Another way biomechanics research improves performance is advances in exercise and conditioning programs.

Preventing and Treating Injury

- Sports medicine professionals have traditionally studied injury data to try to determine the potential causes of disease or injury (epidemiology).
- Engineers and occupational therapists use biomechanics to design work tasks and assistive equipment to prevent overuse injuries related to specific jobs.
- The biomechanical study of auto accidents has resulted in measures of the severity of head injuries, which has been applied in biomechanical testing, and in design of many kinds of helmets to prevent head injury.
- Biomechanics helps the physical therapist prescribe rehabilitative exercises, assistive devices, or orthotics.
- Qualitative analysis of gait (walking) also helps the therapist decide whether sufficient muscular strength and control have been regained in order to permit safe or cosmetically normal walking.

Medical applications

In medical applications, biomechanics play a major role and some of the few points are discussed below:

Clinical Problems in the Cardiovascular System

1. Prosthetic heart valve.
2. Heart assist devices, such as the left ventric1e assist pump, the aortic balloon pump, body acceleration synchronized with heartbeat, peripheral

cuffs, and diastolic counterpulsation.

3. Extracorporeal circulation. Heart-lung machine.Hemodialysis machine.

4. Heart replacement.

5. Postoperative trauma, pulmonary edema, and atelectasis.

6. Arterial pulse wave analysis.

7. Ultrasound applications. Phonoangiography. Turbulent noise analysis. Pseudo-sound generation at atherosclerotic constrictions in arteries.

Quantitative Physiology

1. Systems analysis of physiology.

2. Rheology of biological tissues, such as blood, muscles, bones, connective tissues, and artificial implantable materials.

3. Analysis of fluid transfer across biological membranes and blood vessels.

4. Diffusion analysis such as pulmonary function, and indicator dilution method.

5. Interfaces. Surfactants in the lung. Thrombogenic tendency of blood on artificial implantable materials. Recent work shows that platelet .action and thrombosis on interfaces are shear stress dependent, thus projecting mechanics to the foreground of this important problem.

6. Microcirculation. Biomechanics has contributed to every aspect of microcirculation research. Perhaps it is a historical accident, but physiologists and mechanics researchers have cooperated throughout the development of this branch of science.

Additional Applications

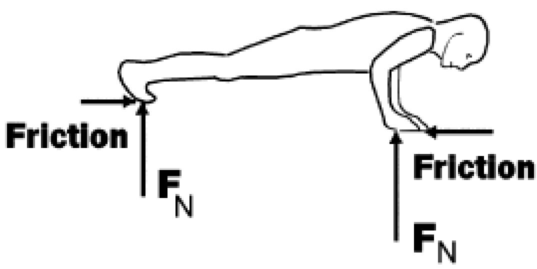
- Surgery
- Implantable Artificial Materials
- Orthopedics, Orthosis, Orthodontics.
- Artifical Limb.
- Artificial Internal organs.
- Wheelchairs and Beds.
- Occupational Safety and Health
- Flight Safety.
- Flying and Swimming in Nature.

10. Describe about the tools used for measuring kinetic variables in biomechanics?(N/D 2016)

Linear Kinetics

Contact forces

- The linear kinetics of the interaction of two objects in contact is also analyzed by resolving the forces into right-angle components.
- The forces between two objects in contact are resolved into the **normal reaction** and **friction**. The normal reaction is the force at right angles to the surfaces in contact, while friction is the force acting in parallel to the surfaces.
- Friction is the force resisting the sliding of the surfaces past each other.
- When the two surfaces are dry, the force of friction (**F**) is equal to the product of the coefficient of friction (μ) and the normal reaction (**F**N), or **F** = μ • **F**N.



Impulse momentum relationship

- Impulse is the effect of force acting over time. Impulse (**J**) is calculated as the product of force and time (**J** = **F** • t), so the typical units are N•s and lb•s. Impulse can be visualized as the area under a force–time graph.

- A person can increase the motion of an object by applying a greater impulse, and both the size of the force and duration of force application are equally important.

Force time principle

- The applied manifestation of Newton's Second Law of Motion as the Impulse–Momentum Relationship is the **Force–Time Principle.**
- If a person can apply force over a longer period of time (large impulse), they will be able to achieve a greater speed (change in momentum) than if they used similar forces in a shorter time interval.

Work energy relationship

Mechanical energy

- In mechanics, **energy** is the capacity to do work. In the movement of everyday objects, energy can be viewed as the mover of stuff (matter), even though at the atomic level matter and energy are more closely related.
- Energy is measured in Joules (J) and is a scalar quantity. One Joule of energy equals 0.74 ft·lbs.
- Energy is a scalar because it represents an ability to do work that can be transferred in any direction. Energy can take many forms (for example, heat, chemical, nuclear, motion, or position).
- There are three mechanical energies that are due to an object's motion or position. The energies of motion are linear and angular **kinetic energy**.
- Linear or translational kinetic energy can be calculated using the following formula: $KE_T = ½m\mathbf{v}^2$.
- The **Law of Conservation of Energy** states that energy cannot be created or destroyed; it is just transferred from one form to another.
- **Example:** A tumbler taking off from a mat has kinetic energy in the vertical direction that is converted into potential energy on the way up, and back into kinetic energy on the way down.
- **Mechanical work:** The mechanical work done on an object is defined as the product of the force and displacement in the direction of the force ($W = \mathbf{F} \cdot \mathbf{d}$). Joules are the units of work: one joule of work is equal to one Nm.

- **Mechanical power: Mechanical power** is an important kinetic variable for analyzing many human movements because it incorporates time. Power is defined as the rate of doing work, so mechanical power is the time derivative of mechanical work or work divided by time (P =W/t).

Angular kinetics

- **Torque:** The rotating effect of a force is called a **torque** or **moment of force**. Torque is calculated as the product of force (**F**) and the **moment arm**.
- The moment arm or leverage is the perpendicular displacement (**d**⊥) from the line of action of the force and the axis of rotation
- **Summing torques:** The state of an object's rotation depends on the balance of torques created by the forces acting on the object.
- Remember that summing or adding torques acting on an object must take into account the vector nature of torques.
- All the muscles of a muscle group sum together to create a joint torque in a particular direction.
- These muscle group torques must also be summed with torques from antagonist muscles, ligaments, and external forces to determine the net torque at a joint.
- **Angular inertia (moment of inertia):** In angular kinetics, inertia is measured by the moment of inertia, a term pretty easy to remember because it uses the terms inertia and moment from moment of force. Like the mass (linear inertia), moment of inertia is the resistance to angular acceleration.
- **Newton's angular analogues:** The angular analogue of Newton's third law says that for every torque there is an equal and opposite torque.
- The angular acceleration of an object is proportional to the resultant torque ,is in the same direction, and is inversely proportional to the moment of inertia. This is the angular expression of Newton's second law.

Equilibrium

- **Mechanical equilibrium** occurs when the forces and torques acting on an object sum to zero.

- Newton's second law accounts for both linear and angular conditions of **static equilibrium** ($\sum$**F** = 0, $\sum$**T** = 0), where an object is motionless or moving at a constant velocity.
- **Dynamic equilibrium** issued to refer to the kinetics of accelerated bodies using Newton's second law ($\sum$**F** = m• **a**, _ $\sum$**T** = **I** • α).

Centre of gravity

- The **center of gravity** is the location in space where the weight (gravitational force) of an object can be considered to act.

Principle of balance

- **Balance** is a person's ability to control their body position relative to some base of support.

11. What is biomechanics? Explain the different force that acts on the body?(M/J 2017)

- **Biomechanics** has been defined as the study of the movement of living things using the science of mechanics.
- **In simple words, biomechanics** is the study of motion and its causes in living things.

Force:

- Force may be defined as mechanical disturbance or load.
- A force acting on an object can deform the object, change its state of motion, or both.
- For example, a person sitting on a chair applies his/her weight on the chair, and yet the chair remains stationary.

Properties of Force as a Vector Quantity:

- A force vector can be illustrated graphically with an arrow such that the orientation of the arrow indicates the line of action of the force vector, the arrow head identifies the direction and sense along which the force is acting, and the base of the arrow represents the point of application of the force vector.

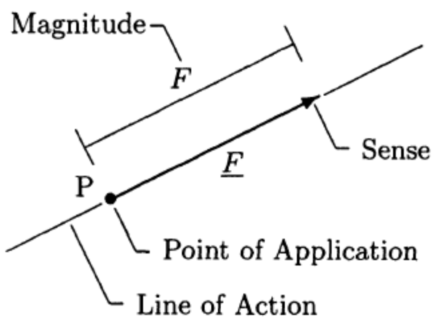


Figure 2.1 *Graphical representation of the force vector.*

Force Systems

- Any two or more forces acting on a single body form a force System.
- Forces may be classified according to their effect on the bodies upon which they are applied or according to their orientation as compared to one another.

1. External and Internal Forces

- A force may be broadly classified as external or internal.
- Almost all commonly known forces are external forces.
- For example, when you push a cart, hammer a nail, sit on a chair, kick a football,or shoot a basketball, you apply an external force on the cart, nail, chair, football, or basketball.
- Internal forces, on the otherhand, are the ones that hold a body together when the body is under the effect of externally applied forces.
- For example, a piece of string does not necessarily break when it is pulled from both ends.

2. Normal and Tangential Forces

- In mechanics, the word "normal" implies perpendicular.
- If a force acting on a surface is applied in a direction perpendicular to that surface, then the force is called a normal force
- For example, a book resting on a flat horizontal desk applies a normal force on the desk, the magnitude of which is equal to the weight of the book.
- A tangential force is that applied on a surface in the direction parallel to the surface.

- A good example of a tangential force is the frictional force, pushing or pulling a block will cause a frictional force to occur between the bottom surface of the block and the floor.
- The line of action of the frictional force is always tangential to the surfaces in contact.

3. Tensile and Compressive Forces

- A tensile force applied on a body will tend to stretch or elongate the body, whereas a compressive force will tend to shrink the body in the direction of the applied force.
- For example, a tensile force applied on a rubber band will stretch the band.
- Poking into an inflated balloon will produce a compressive force on the balloon. It must be noted that there are certain materials upon which only tensile forces can be applied.
- For example, a rope, a cable, or a string cannot withstand compressive forces.
- The shapes of these materials will be completely distorted under compressive forces. Similarly, muscles contract to produce tensile forces that pull together the bones to which they are attached.
- Muscles can neither produce compressive forces nor exert a push.

4. Coplanar Forces

- A system of forces is said to be coplanar if all the forces are acting on a two-dimensional (plane) surface.
- Forces forming a coplanar system have at most two non-zero components.
- Therefore, with respect to the Cartesian (rectangular) coordinate frame, it is sufficient to analyze coplanar force systems by considering the x and y components of the forces involved.

5. Collinear Forces

- A system of forces is collinear if all the forces have a common line of action.
- For example, the forces applied on a rope in a rope-pulling contest form a collinear force system.

6. **Concurrent Forces**

- A system of forces is concurrent if the lines of action of the forces have a common point of intersection.
- Examples of concurrent force systems can be seen in various traction devices.
- Owing to the weight in the weight pan, the cables stretch and forces are applied on the pulleys and the leg.
- The force applied on the leg holds the leg in place.

7. **Parallel Forces**

- A set of forces form a parallel force system if the lines of action of the forces are parallel to each other.
- An example of a parallel force system is illustrated by a human arm flexed at a right angle and holding an object.
- The forces on the forearm are the weight of the object, the weight of the arm itself, the tension in the biceps muscle, and the joint reaction force at the elbow.

8. **Gravitational Force or Weight**

- The force exerted by Earth on an object is called the gravitational force or weight of the object.
- The magnitude of weight of an object is equal to the mass of the object times the magnitude of gravitational acceleration.
- The magnitude of the gravitational acceleration can vary slightly with altitude.

9. **Frictional Forces**

- Frictional forces occur between two surfaces in contact when one surface slides or tends to slide over the other.
- When a body is in motion on a rough surface or when an object moves in a fluid(a viscous medium such as water), there is resistance to motion because of the interaction of the body with its surroundings.
- Insome applications friction may be desirable, while in others it may have to be reduced to a minimum.

- For example, it would be impossible to start walking in the absence of frictional forces.
- Automobile, bicycle, and wheelchair brakes utilize the principles of friction.
- There are several factors that influence frictional forces.
- Friction depends on the nature of the two sliding surfaces.
- For example, if all other conditions are the same, the friction between two metal surfaces would be different than the friction between twowood surfaces in contact.
- Friction is larger for materials that strongly interact. Friction depends on the surface quality and surface finish.
- A good surface finish can reduce frictional effects.
- The frictional force does not depend on the total surface area of contact.

12. Explain Newton's laws and give necessary examples for each law relating to the physiological system?(M/J 2017)

Newton's 3 Law's of Motion

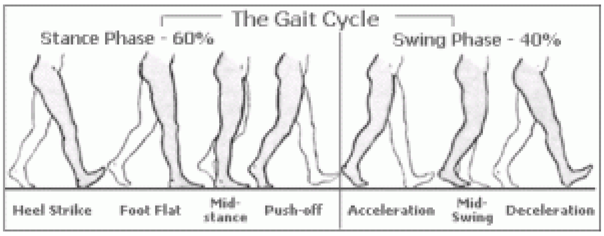
1) Law of Inertia

- An object in a state of constant velocity tends to remain in that state of motion unless an unbalanced force is applied to it.
- In other words, it is the resistance to motion changes. There are important considerations when conceptualizing inertia.
- One of these considerations is that rest is a constant velocity and can be considered to have inertia.
- Another consideration is that gravity is an unbalanced force acting on all objects.

Inertia applied to physiological systems

- Consider the late swing phase of gait and the forces going forward with the lower extremity.

- Just prior to heel-strike there are almost no muscles activated that bring the extremity forward, yet it is still proceeding to travel forward in space.
- This is inertia. To deal with this inertia the body deploys an eccentric contraction of the hamstrings to slow down the extremity to prepare for heel-strike and to reduce harsh reactionary forces.



Eccentric deceleration of inertia during late phase of gait

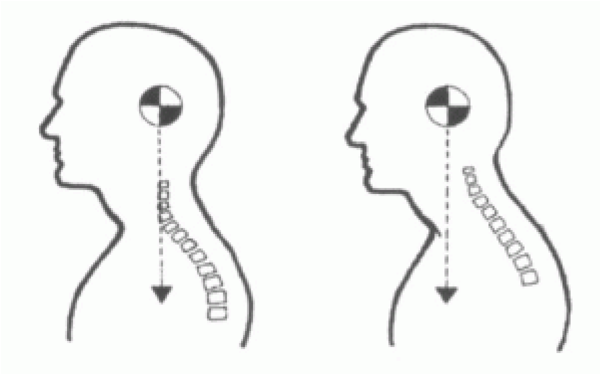
2) Force = Mass x Acceleration (F = ma)

- The net force applied to a body (mass) produces a proportional acceleration. This law describes the relationship between an object's mass, acceleration, and the applied force.
- Both acceleration and force must have the same vector direction. This can also be viewed in different terms:
- Momentum = mass x velocity. The change of momentum of a body is proportional to the impulse impressed on the body, and happens along the straight line on which that impulse is impressed.
- Momentum cannot be changed unless acted upon by an outside force; it can only be conserved.
- Acceleration is proportional to the unbalanced forces acting on it and inversely proportional to the mass of the object (a = F/m)

F=ma Applies to physiological system

- Muscles are the tissues that contract and create force on the body's levers (connective tissue, bones).
- With any human movement, F=ma can be used to create a simplified calculation of force. This equation can even be used with static positions.

- Consider the static forward head posture. Gravity and the mass of the head imposes an antero-inferior force.
- To counter this force and prevent your neck from snapping off at your desk, you have to constantly contract your levator scapulae, upper trapezius, and posterior cervical muscles to counter this force.
- By calculating the acceleration of gravity and mass of the head, you can begin to calculate the muscle forces necessary to prevent movement.



Mass of head and gravity combine to impose a downward force

3) Action Reaction Law

- For every action there is an equal and opposite reaction.
- This law describes how forces always come in pairs, meaning that anytime objects are contacting each other, they are exerting a force.
- An important consideration here is the concept that gravity is ALWAYS touching every object.

Action-Reaction Applies to physiological system

- Putting that ankle weight on a patients leg will create an increase in the force of the mass and downward pull with gravity, the reaction is that the opposing muscle will have to create a force to overcome this mass.
- Another example of this law is with ground reaction forces.
- Running on soft ground will result in much less impact forces than running on hard concrete.

Ground Reaction Forces During Gait

13. Describe the motion of viscous fluid by deriving the Navier Stoke's equation?(M/J 2017)

NAVIER-STOKES EQUATIONS

The integral form of the linear momentum equation was discussed in Linear Momentum Integral Equation. Recall, Newton's second law on a differential fluid element is

$$\delta \mathbf{F} = \delta m \mathbf{a}$$

where δ**F** is the resultant force acting on the fluid element (mass = δm). **a** is the acceleration of the fluid element, and it is given by

$$\mathbf{a} = \frac{D\mathbf{V}}{Dt} = \frac{\partial \mathbf{V}}{\partial t} + (\mathbf{V} \cdot \nabla)\mathbf{V}$$

Expanding into its Cartesian components yields

$$a_x = \frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y} + w\frac{\partial u}{\partial z}$$
$$a_y = \frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y} + w\frac{\partial v}{\partial z}$$
$$a_z = \frac{\partial w}{\partial t} + u\frac{\partial w}{\partial x} + v\frac{\partial w}{\partial y} + w\frac{\partial w}{\partial z}$$

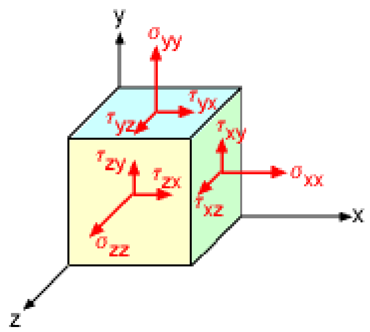
There are two types of forces acting on the fluid element: body force ($\delta\mathbf{F_B}$) and surface force ($\delta\mathbf{F_S}$).

$$\delta\mathbf{F} = \delta\mathbf{F_B} + \delta\mathbf{F_S}$$

The only body force considered here is the weight of the fluid element. That is,

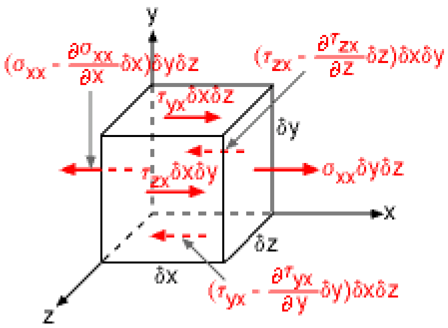
$$\delta\mathbf{F_B} = \delta m\mathbf{g} = \delta m\,(g_x\mathbf{i} + g_y\mathbf{j} + g_z\mathbf{k})$$

Generally, gravity only acts in one direction, but since the coordinate system is not set, all three terms are included for the general case.



Notations for the Stresses

- The surface forces are due to the stresses exerted on the sides of the fluid element. There are two types of stresses applied on the surface: normal stress (σ_{ij}) and shear stress (τ_{ij}).
- Normal stress acts perpendicular to the surface while shear stress is tangential to the surface.
- The subscript i refers to the axis normal to the surface, and the subscript j represents the direction of the stress.
- The notation of the stresses is illustrated further in the animation shown on the left.
- All surface forces acting in the x-direction on the fluid element are shown in the figure. A summation of the surface forces in the x-direction yields



Surface Forces in the x-direction

$$\delta F_{Sx} = \left[\frac{\partial \sigma_{xx}}{\partial x} + \frac{\partial \tau_{yx}}{\partial y} + \frac{\partial \tau_{zx}}{\partial z}\right]\delta x\, \delta y\, \delta z$$

- Note that the stresses are multiplied by the area to obtain the surface forces. Similarly, the total surface forces in the y- and z-directions (not shown in figure) are obtained as

$$\delta F_{Sz} = \left[\frac{\partial \sigma_{zz}}{\partial z} + \frac{\partial \tau_{xz}}{\partial x} + \frac{\partial \tau_{yz}}{\partial y}\right]\delta x\, \delta y\, \delta z$$

$$\delta F_{Sy} = \left[\frac{\partial \sigma_{yy}}{\partial y} + \frac{\partial \tau_{xy}}{\partial x} + \frac{\partial \tau_{zy}}{\partial z}\right]\delta x\, \delta y\, \delta z$$

- The resultant surface force is then given as

$$\delta \mathbf{F_S} = \delta F_{Sx}\mathbf{i} + \delta F_{Sy}\,\mathbf{j} + \delta F_{Sz}\,\mathbf{k}$$

- The mass of the fluid element can be expressed in terms of its volume and fluid density ($\delta m = \rho\, \delta x \delta y \delta z$), so that the linear momentum equation in Cartesian coordinates reduces to

$$\rho g_x + \frac{\partial \sigma_{xx}}{\partial x} + \frac{\partial \tau_{yx}}{\partial y} + \frac{\partial \tau_{zx}}{\partial z} = \rho\left(\frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y} + w\frac{\partial u}{\partial z}\right)$$

$$\rho g_y + \frac{\partial \sigma_{yy}}{\partial y} + \frac{\partial \tau_{xy}}{\partial x} + \frac{\partial \tau_{zy}}{\partial z} = \rho\left(\frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y} + w\frac{\partial v}{\partial z}\right)$$

$$\rho g_z + \frac{\partial \sigma_{zz}}{\partial z} + \frac{\partial \tau_{xz}}{\partial x} + \frac{\partial \tau_{yz}}{\partial y} = \rho\left(\frac{\partial w}{\partial t} + u\frac{\partial w}{\partial x} + v\frac{\partial w}{\partial y} + w\frac{\partial w}{\partial z}\right)$$

- For Newtonian fluids, such as water, oil and air, the shear stress field is symmetric, and it is related to the rate of shear strain in a linear fashion.

$$\tau_{xy} = \tau_{yx} = \mu\left(\frac{\partial v}{\partial x} + \frac{\partial u}{\partial y}\right)$$

$$\tau_{yz} = \tau_{zy} = \mu\left(\frac{\partial w}{\partial y} + \frac{\partial v}{\partial z}\right)$$

$$\tau_{zx} = \tau_{xz} = \mu\left(\frac{\partial u}{\partial z} + \frac{\partial w}{\partial x}\right)$$

$$\sigma_{xx} = -p - \frac{2}{3}\mu\nabla\cdot\mathbf{V} + 2\mu\frac{\partial u}{\partial x}$$

$$\sigma_{yy} = -p - \frac{2}{3}\mu\nabla\cdot\mathbf{V} + 2\mu\frac{\partial v}{\partial y}$$

$$\sigma_{zz} = -p - \frac{2}{3}\mu\nabla\cdot\mathbf{V} + 2\mu\frac{\partial w}{\partial z}$$

Where μ is the viscosity of the fluid. Notice, the pressure term, p, only acts normal to the surface for each element face. Also, the pressure is assumed to be the same on all three faces, i.e. hydrostatic pressure.

- For incompressible flow, the term $\nabla\cdot\mathbf{V}$ is zero based on the continuity equation. The linear momentum equations thus become

$$\rho\left(\frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y} + w\frac{\partial u}{\partial z}\right) = \rho g_x - \frac{\partial p}{\partial x} + \mu\left(\frac{\partial^2 u}{\partial x^2} + \frac{\partial^2 u}{\partial y^2} + \frac{\partial^2 u}{\partial z^2}\right)$$

$$\rho\left(\frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y} + w\frac{\partial v}{\partial z}\right) = \rho g_y - \frac{\partial p}{\partial y} + \mu\left(\frac{\partial^2 v}{\partial x^2} + \frac{\partial^2 v}{\partial y^2} + \frac{\partial^2 v}{\partial z^2}\right)$$

$$\rho\left(\frac{\partial w}{\partial t} + u\frac{\partial w}{\partial x} + v\frac{\partial w}{\partial y} + w\frac{\partial w}{\partial z}\right) = \rho g_z - \frac{\partial p}{\partial z} + \mu\left(\frac{\partial^2 w}{\partial x^2} + \frac{\partial^2 w}{\partial y^2} + \frac{\partial^2 w}{\partial z^2}\right)$$

- The above equations are generally referred to as the Navier-Stokes equations, and commonly written as a single vector form,

$$\rho \frac{D\mathbf{V}}{Dt} = \rho \mathbf{g} - \nabla p + \mu \nabla^2 \mathbf{V}$$

- Although the vector form looks simple, this equation is the core fluid mechanics equations and is an unsteady, nonlinear, 2nd order, partial differential equation.
- It is extremely hard to solve, and only simple 2D problems have been solved. Computational Fluid Dynamics (CFD) is most often used to solve the Navier-Stokes equations.

14. Write a brief note on Strain energy function.

- Elastic response of any solid in tension can be characterized by means of a stored energy function.
- The theory of elasticity is the construction of specific forms of strain- energy function from the result of experiments involving three- dimensional deformations.
- Let W be the strain energy per unit mass of the tissue.
- W= Strain energy per unit mass of the tissue and ρ_0 = Density in the zero stress state. Then,
- **$\rho_0 W$= Strain energy/ Unit volume of the tissue in the zero stress state.**
- Let W be expressed in terms of 9 strain components, $E_{11}, E_{22}, E_{33}, E_{12}, E_{21}, E_{23}, E_{32}, E_{13}, E_{31}$ and written in a form that is symmetric in the symmetric components E_{12} and E_{21}, E_{23} and E_{32} and E_{13} and E_{31}.
- The 9 components are treated as independent variables when partial derivatives of $\rho_0 W$ are formed.
- When such a strain energy function exists, the stress components S_{ij} can be obtained as derivatives of $\rho_0 W$.

$$S_{ij} = \frac{\partial(\rho_0 W)}{\partial E_{ij}}$$

- More explicitly, for rectangular elements we have

$$S_{11} = \frac{\partial(\rho_0 W)}{\partial E_{11}} \quad , \quad S_{22} = \frac{\partial(\rho_0 W)}{\partial E_{22}} \quad , \quad S_{33} = \frac{\partial(\rho_0 W)}{\partial E_{33}}$$

- Not all elastic materials have a strain- energy function. The strain-energy functions are called as **hyper elastic materials.**
- It can be shown that by the following,
- Let W be expressed in terms of gradient tensor $\frac{\partial xi}{\partial aj}$ not in terms of E_{ij}.

  Where (a_1, a_2, a_3) are the co-ordinates of a material particle in the zero stress state and (x_1, x_2, x_3) are the co-ordinates of the same particle in the deformed state of the body.

- Then the lagrangian stresses are ,
- $T_{ij} = \frac{\partial(\rho_0 W)}{\partial\,(\partial xi/\partial aj)}$ where $\partial xi/\partial aj = \lambda$ and $\frac{\partial xi}{\partial aj} \neq \frac{\partial xj}{\partial ai}$ and $T_{ij} \neq T_{ji}$.

- Referring to the rectangular element

  $T_{11} = \frac{\partial(\rho_0 W)}{\partial\lambda_1}$, $T_{22} = \frac{\partial(\rho_0 W)}{\partial\lambda_2}$, $T_{33} = \frac{\partial(\rho_0 W)}{\partial\lambda_3}$.

- If the material is perfectly elastic, then the existence of strain energy function can often be justified on the basis of thermodynamics.
- Living tissues are not perfectly elastic; therefore they cannot be having a strain energy function in the thermodynamic sense.

UNIT II

BIOFLUID MECHANICS

- ✓ **Viscosity** and capillary viscometer
- ✓ **Rheological properties of blood**
- ✓ **Laminar flow**
- ✓ **Couette flow**
- ✓ **Hagen-poiseuille equation**
- ✓ **Turbulent flow**
- ✓ **Cardiovascular s/m**

  - Development of artificial heart valve

  - Artificial heart valve

  - Testing of heart valve
- ✓ **Structure, functions, material properties and modeling of blood vessels.**

LIST OF IMPORTANT QUESTIONS

PART A

1. What is Biofluid Mechanics?
2. **What is viscosity? (M/J 2016)**
3. **State the four most commonly used basic types of heart valve prostheses. (M/J 2016)**
4. **What is Rheology? (N/D 2015)**
5. **State the abnormalities of the blood. (N/D 2015)**
6. **Explain Couette flow (or) what is couette flow? (A/M 2017)**
7. **Explain Laminar flow**
8. **Define Turbulent flow**
9. **What is Hagen-Poiseuille equation?**
10. **What is called a non-Newtonian fluid?(N/D 2016)**
11. **Calculate the Reynolds number for plasma flowing in a 1.5 mm artery at a velocity of 45 cm/s, and state the flow regime for a viscosity of 0.012 g/cms.(N/D 2016)**
12. **List the mechanical property of mitral valve. (A/M 2017)**
13. What is capillary viscometer?
14. What is Reynolds number?
15. Write a brief notes on Artificial Heart Valves.
16. What is mechanical heart valve?
17. What is biological heart valve?
18. What are the materials used in artificial heart valves?
19. What is Blood Rheology
20. What are the different types of blood vessels?
21. What are the functions of blood?
22. What are the elements of blood and its characteristics?
23. What are the testing methods of heart valves?

PART B

1. **Discuss about the mechanics of aortic and pulmonary valves. (M/J 2016)**
2. **Write a brief note on artificial heart valves or Discuss the mechanical properties of heart valves (N/D 2015)**
3. **What is Rheology? Discuss about Rheological properties of blood. Or Discuss about the elements of blood and its characteristics.(N/D 2015)(N/D 2016)**
4. **Distinguish between mechanical and biological valve?(N/D 2016)**
5. **Write short notes on current trends in heart valve design (N/D 2016)**
6. **When the blood is said to be Newtonian? Explain with necessary equations. (A/M 2017)**
7. **Prove that the velocity profile of blood in the blood vessel is parabola. (A/M 2017)**
8. Write a notes on testing of artificial heart valves
9. Describe the structure and functions of blood vessels.
10. Explain the material properties and modeling of blood vessels.
11. Explain the development of artificial heart valves.
12. Explain laminar flow, Couette flow and Hagen-poiseuille equation, turbulent flow.
13. **The aorta of a male patient had an inner radius of 13 mm and was 2.2 mm thick in the diastolic state. It was 50 cm long and expanded due to the pumping of the heart. When the heart valve opened in the systolic phase, 70 mL of blood was discharged. Half of this blood was initially stored in the aorta, expanding its wall to some inner radius. Assume the diastolic pressure and systolic pressure to be 80 mmHg and 130 mmHg, respectively, and the heart rate to be 72 beats per minute. Calculate the systolic radius and the wall thickness. What is the stress in the wall in the systolic state? How much energy is stored in the elastic wall? What is the average Reynolds number in the aorta? Is the flow laminar or turbulent? (M/J 2016)**

PART A

1. What is Biofluid Mechanics**?**

- Biological fluid mechanics, or biofluid mechanics, is the study of both gas and liquid fluid flows in or around biological organisms.
- An often studied liquid biofluid problem is that of blood flow in the human cardiovascular system.
- Under certain mathematical circumstances, blood flow can be modeled by the Navier–Stokes equations.
- Examples of biofluids are Amniotic fluid, CSF etc.

2. What is viscosity? (M/J 2016)

- Viscosity is a measure of a fluid's resistance to flow.
- It describes the internal friction of a moving fluid.
- Viscosity is a property arising from collisions between neighboring particles in a fluid that are moving at different velocities.
- For example, honey has a much higher viscosity than water.

3. State the four most commonly used basic types of heart valve prostheses. (M/J 2016)

- Caged ball
- Tilting disc
- Heterograft (Porcine or bovine)
- Homograft (Human cadavers)

4. What is Rheology? (N/D 2015)

- A branch of physics that deals with the deformation and flow of matter, especially the non-Newtonian flow of liquids and the plastic flow of solids.
- Rheological properties can be measured from bulk sample deformation using a mechanical rheometer or on a micro-scale by using a micro capillary viscometer or an optical technique such as Microrheology.

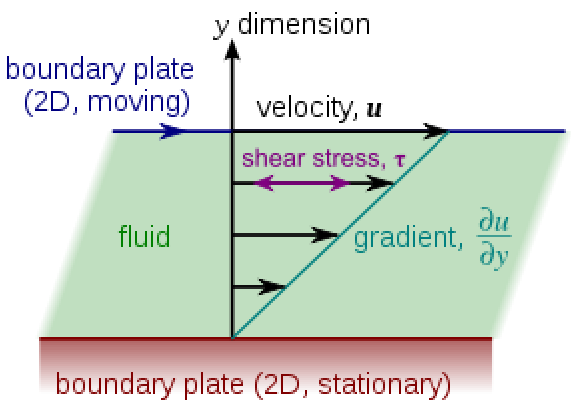
5. **State the abnormalities of the blood. (N/D 2015)**

Any abnormalities in tests conducted on blood. Some of the blood abnormalities are the following:

- Bone marrow disease
- Acute lymphocytic leukemia
- Anemia
- Thrombosis
- Deep vein thrombosis
- Thrombocytosis

6. **Explain Couette flow (or) what is couette flow? (A/M 2017)**

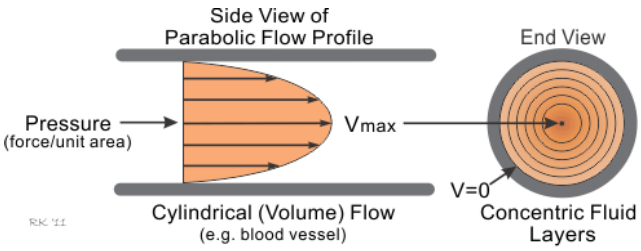
- Couette flow is the laminar flow of a viscous fluid in the space between two parallel plates, one of which is moving relative to the other.
- The flow is driven by virtue of viscous drag force acting on the fluid and the applied pressure gradient parallel to the plates.
- Couette flow is frequently used in undergraduate physics and engineering courses to illustrate shear-driven fluid motion.



7. **Explain Laminar flow**

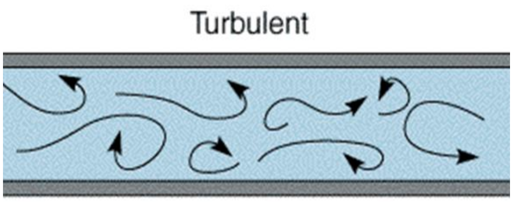
- It is the normal condition for blood flow throughout most of the circulatory system.
- It is characterized by concentric layers of blood moving in parallel down the length of a blood vessel.
- The highest velocity (V_{max}) is found in the center of the vessel. The lowest velocity (V=0) is found along the vessel wall.

- The flow profile is parabolic once laminar flow is fully developed. This occurs in long, straight blood vessels, under steady flow conditions.



8. Define Turbulent flow

- Movement of a fluid in which the individual particles of fluid move in irregular patterns even though the overall flow is in one direction.
- Turbulent flow is common in non viscous fluids moving at high velocities.



9. What is Hagen-Poiseuille equation?

- Hagen–Poiseuille equation also known as the **Hagen–Poiseuille** law.
- Physical law that gives the **pressure drop** in an incompressible and Newtonian fluid in laminar flow flowing through a long cylindrical pipe of constant cross section.

$$\Phi = \frac{\pi}{2\eta}\frac{|\Delta P|}{\Delta x}\int_0^R (rR^2 - r^3)\,dr = \frac{|\Delta P|\pi R^4}{8\eta\Delta x}$$

10. What is called a non-Newtonian fluid?(N/D 2016)

- In reality most fluids are non-Newtonian, which means that their viscosity is dependent on shear rate (Shear Thinning or Thickening) or the deformation history (Thixotropic fluids).
- In contrast to Newtonian fluids, non-Newtonian fluids display either a non-linear relation between shear stress and shear rate have a yield stress, or viscosity that is dependent on time or deformation history

- Non-Newtonian behavior of fluids can be caused by several factors, all of them related to structural reorganization of the fluid molecules due to flow. In polymer melts and solutions, it is the alignment of the highly anisotropic chains what results in a decreased viscosity. In colloids, it is the segregation of the different phases in the flow that causes a shear thinning behavior.

11. Calculate the Reynolds number for plasma flowing in a 1.5 mm artery at a velocity of 45 cm/s, and state the flow regime for a viscosity of 0.012 g/cms.(N/D 2016)

Solution:

$$R_e = \frac{\rho VL}{\mu}$$

Given: v= 45 cm/s, μ= 0.012 g/cms, L= 1.5 mm, ρ= 1.025g/ml or 1025 kg/m^3

$$\textbf{Re} = \frac{1.025 \times 45 \times 1.5 \times 10^{-3}}{0.012}$$

$$= \frac{0.0691875}{0.012}$$

$$= 5.76$$

The flow is laminar.

12. List the mechanical property of mitral valve. (A/M 2017)

- Biaxial testing,
- histological measurements and
- theoretical continuum mechanics modeling

13. What is capillary viscometer?

- A capillary viscometer is an instrument used to measure the viscosity, or thickness, of a liquid.
- The flow, or efflux, time is directly proportional to the liquid's kinematic viscosity, and may be converted directly to viscosity by use of a conversion factor unique to each instrument.

- Viscosity is generally temperature dependent, so the capillary viscometer is usually used in a controlled-temperature water bath set to a specific temperature.

14. What is Reynolds number?

- Reynolds number is used to check whether the flow is laminar or turbulent.
- It is denoted by **R_e.** This number got by comparing inertial force with viscous force.
- Reynolds number formula is used in the problems to find the Velocity (V), density (*ρ*), Viscosity (*μ*) and diameter (L) of the fluid. It is dimensionless.

$$\text{Reynolds number} = \frac{\text{Inertial force}}{\text{Viscous force}}$$

$$\text{i.e., } R_e = \frac{\rho \mathrm{VL}}{\mu}$$

Where, *ρ* is the density of the fluid, V is the velocity of the fluid, *ρ* is the density of fluid, *μ* is the viscosity of fluid, L is the length or diameter of the fluid.

15. Write a brief notes on Artificial Heart Valves.

An artificial heart valve is a device implanted in the heart of a patient with valvular heart disease. When one of the four heart valves malfunctions, the medical choice may be to replace the natural valve with an artificial valve. This requires open-heart surgery. There are two basic types of valves that can be used for valve replacement.

❖ **Mechanical heart valves**

- Caged-ball
- Tilting-disk
- Single leaflet
- Bileaflet.

❖ **Tissue heart valves**

- Homograft or allograft (from human)
- Xenograft (from animal)
  - Bovine (cow)cardiac tissue

- Porcine (pig) heart valves

16. What is mechanical heart valve?

- Mechanical valves are made completely from man-made materials.
- They have made up of very light weight and long lasting materials.
- The main advantage of mechanical valves is that they last a very long time in most cases 20 years or more.
- Ex: Caged ball, Tilting disk, Single leaflet, Bileaflet.

17. What is biological heart valve?

- Biological valves are made from human or animal tissue that has been specially treated so that your body does not reject the valves.
- After being treated, the valves are attached to man-made materials to give them support.
- They do not last as long as man-made valves, although they probably last longer in older people.
- Ex: Homograft or allograft, Xenograft, Bovine, Porcine.

18. What are the materials used in artificial heart valves?

- Metal alloys
- Pyrolytic carbon
- Dacron
- Teflon

19. What is Blood Rheology?

- Hemorheology, or blood rheology, is the study of flow properties of blood and its elements of plasma and cells.
- Proper tissue perfusion can occur only when blood's rheological properties are within certain levels.
- Alterations of these properties play significant roles in disease processes.

- Blood viscosity is determined by plasma viscosity, hematocrit (volume fraction of red blood cell, which constitutes 99.9% of the cellular elements) and mechanical properties of red blood cells.

20. What are the different types of blood vessels?

- Arteries
- Arterioles
- Capillaries
- Venules
- Veins

21. What are the functions of blood?

- Carries oxygen and nutrients to active tissues
- Delivers carbon dioxide to the lungs
- Brings metabolic end products to kidneys
- Blood is a buffering reservoir that controls pH of bio-fluids
- Plays a major role in the body's immune system
- In addition to mass, it also transports heat

22. What are the elements of blood and its characteristics?

- The main elements of blood include RBC,WBC, platelets, and plasma.
- **RBC** carries O_2 from the lungs to all other body tissues and pick up CO_2 back to lungs.
- **WBC** are one of the body's defenses against disease.
- **Platelets** are blood elements that lead to the formation of blood clots in response to injury.
- **Plasma** is a yellowish fluid composed of about 92 percent water and 7 percent vital proteins, such as albumin, gamma globulin, anti-hemophilic factor, and other clotting factors. (55% in blood)

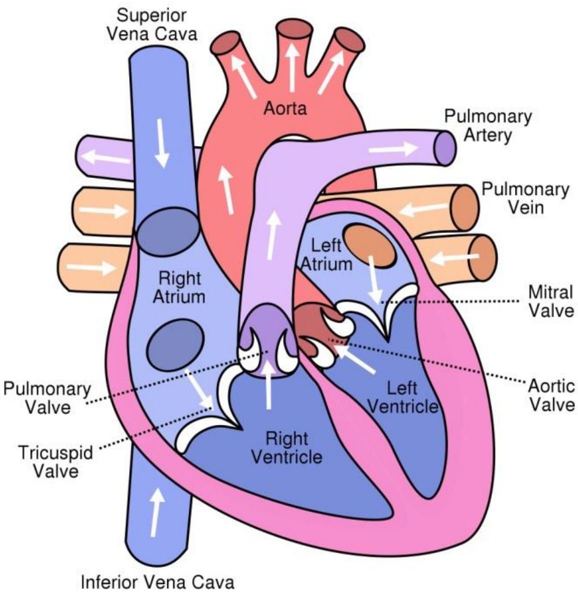
23. What are the testing methods of heart valves?

- In vitro testing of artificial heart valves
- Durability test
- Testing of polymer-based heart-valve prosthesis

PART B

1. Discuss about the mechanics of aortic and pulmonary valve. (M/J 2016)

- The heart has four separate compartments or chambers. The upper chamber on each side of the heart, which is called an atrium, receives and collects the blood coming to the heart.
- The atrium then delivers blood to the powerful lower chamber, called a ventricle, which pumps blood away from the heart through powerful, rhythmic contractions.
- The right side receives oxygen-poor blood from the various regions of the body and delivers it to the lungs. In the lungs, oxygen is absorbed in the blood.
- The left side of the heart receives the oxygen-rich blood from the lungs and delivers it to the rest of the body.
- The **aortic and pulmonic valves are known as the semilunar valves**, whereas the tricuspid and mitral valves are referred to as the atrioventricular valves.
- All 4 cardiac valves are surrounded by fibrous tissue forming partial or complete valvular rings, or annuli. These annuli join the fibrous skeleton of the heart to anchor and support the valvular structures.



- **The aortic valve** is located between the aorta and the heart's left ventricle. The pulmonary vein delivers oxygenated blood to the heart's left atrium. Then it passes through the mitral valve and into the left ventricle. With each of the heart muscle's contractions, oxygenated blood exits the left ventricle through the aortic valve.
- The main function of aortic valve is that it prevents back flow of blood from aorta to left ventricle. Aortic **regurgitation (back flow)** occurs if oxygenated blood flows in the wrong direction.
- The **pulmonic valve** is located in the right ventricle of the heart. It allows blood to leave the heart via the arteries.
- The pulmonic valve opens into the pulmonary artery. It is a one-way valve, meaning that blood cannot flow back into the heart through it.
- The valve is opened by the increased blood pressure of the ventricular systole (contraction of the muscular tissue), pushing blood out of the heart and into the artery. It closes when the pressure drops inside the heart.

2. Write a brief note on artificial heart valves or Discuss the mechanical properties of heart valves (N/D 2015)

An **artificial heart valve** is a device implanted in the heart of a patient with valvular heart disease. When one of the four heart valves malfunctions, the medical choice may be to replace the natural valve with an artificial valve. This requires open-heart surgery.

Valves are integral to the normal physiological functioning of the human heart. Natural heart valves are evolved to forms that perform the functional requirement of inducing unidirectional blood flow through the valve structure from one chamber of the heart to another.

There are two basic types of valves that can be used for valve replacement,

- ❖ Mechanical heart valves and
- ❖ Tissue heart valves

Mechanical heart valves

- Mechanical valves are made completely from man-made materials.
- They have made up of very light weight and long lasting materials.
- The main advantage of mechanical valves is that they last a very long time in most cases 20 years or more.
- Anticoagulant should be needed.
- Ex: Caged ball, Tilting disk, Single leaflet, Bileaflet.

Tissue heart Valves:

- Biological valves are made from human or animal tissue that has been specially treated so that your body does not reject the valves.
- After being treated, the valves are attached to man-made materials to give them support.
- They do not last as long as man-made valves, limited lifespan average 15 years.
- No need of anticoagulant.
- Ex: Homograft or allograft, Xenograft, Bovine, Porcine.
  Their main weakness however, is their

Types of mechanical heart valve:

The major types of mechanical valves are ,

- Caged-ball
- Tilting-disk
- Single leaflet
- Bileaflet.

Caged-ball

- The first artificial heart valve was the **caged-ball**, which utilizes a metal cage to house a silicone elastomer ball.
- When blood pressure in the chamber of the heart exceeds that of the pressure on the outside of the chamber the ball is pushed against the cage and allows blood to flow.
- At the completion of the heart's contraction, the ball moves back against the base of the valve forming a seal.

❖ The first human implant consisted of a silicone ball enclosed in a cage formed by wires originating from the valve housing.
❖ Caged ball valves have a high tendency to forming blood clots, so the patient must have a high degree of anti-coagulation.

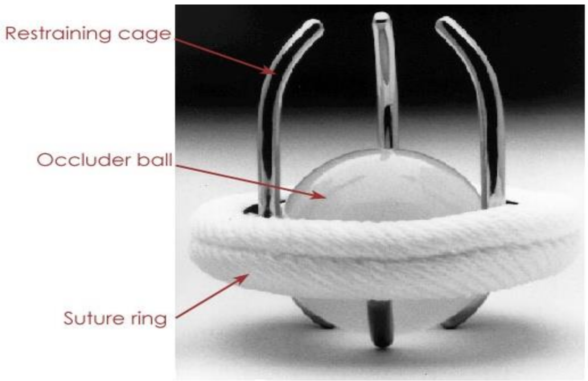


Figure Caged-ball valve.

Advantage:

- Very simple design, only moving part is the ball.
- Poor hemodynamic, blood must flow around the ball making the heart work harder
- Higher rate of thromboembolism.

Tilting Disc Valve:

❖ In the mid-1970s, a new valve was introduced: the tilting disc valve. The purpose in creating the titling-disc valve was to restore the central blood flow that was lost with the ball valve design.
❖ These valves consist of a single circular disc restrained by two metal struts and a metal ring. The struts are attached to the metal ring. The struts prevent the disc from escaping the device in either direction.
❖ The disc opens and closes based on the same principles used in the ball valve design, except a disc is used instead of a ball.
❖ Tilting disc valves can open at an angle of 60° and at a rate of 70 beats per minute. The angular opening of this valve reduces damage to blood cells.
❖ These are major improvements over the ball design but the struts of the tilting disc valves tend to fatigue and fracture over long periods of time.

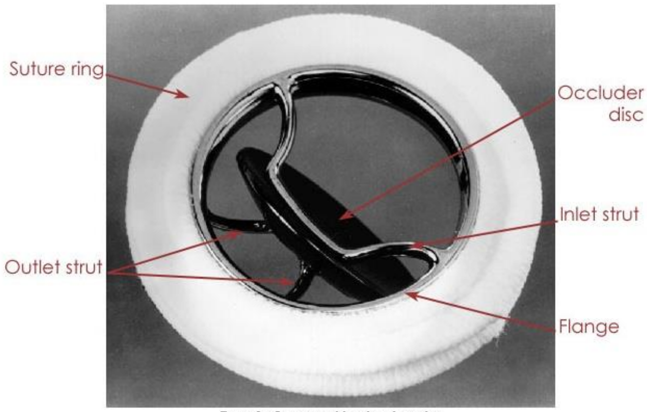


Figure ˆ: Over view of the titling-disc valve

Advantages:

- Improved hemodynamic.
- Lower rates of thromboembolism.

Single leaflet valve/kay Shiley mitral disc valve:

❖ Another type of valve is the single leaflet valve. These generally consist of two parallel metal struts that hold a disc in place over the orifice.
❖ It is very similar to the ball valve; however it has a flat disc instead of a ball.


The Kay-Sheiley single leaflet valve

Bileaflet valves

❖ The first bileaflet valves were introduced in 1978. Some bileaflet valves are shown in Figure.
❖ The bileaflet design consists of two semicircular leaflets which pivot on hinges. Bileaflet valves have the best central flow – the leaflets open completely, allowing very little resistance to blood flow.

- These valves correct the problem of central flow and blood cell damage; however, they allow some backflow.
- This is a serious design flaw: many natural heart valves are replaced with mechanical valves because the valve became stiff and allowed backflow. Nevertheless, the majority of mechanical heart valves used today are bileaflet valves because they allow the least resistance to flow and the least blood damage.

Advantages:

- Most implanted valve today
- Most improved hemodynamic
- Still a risk for thrombosis
- Allow some backflow

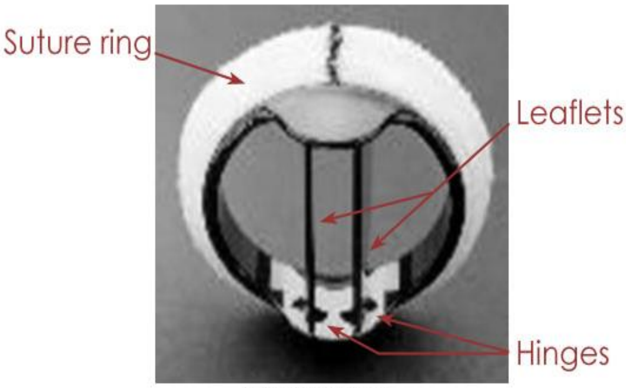


Figure . : Bileaflets valve

Material used:

One of the major design considerations of any object to be implanted in the body is the choice of materials.

Those used must be able to withstand the harsh and corrosive environment of the body, they must be inert, and they must be biocompatible so they do not elicit a rejection response.

Particularly in heart valves, factors to consider are:

- How will the material impact hemodynamic, will it cause platelet aggregation or thrombosis,
- How will the device damage blood cells, and

- How are the mechanical properties sufficient to withstand the repeated cycles the valve will encounter in its lifetime.

Many different materials are used in the creation of artificial heart valves.

Metal alloys

- Metal alloys consisting of stainless steel or titanium are often used to give mechanical strength and for their corrosion resistance properties. The struts on some leaflet valves and the cage on caged-ball models are commonly made of metal alloys due to their strength and durability requirements.

Pyrolytic carbon

- Pyrolytic carbon is another valuable material for its strength and its ability to prevent clotting. It is biocompatible, thromboresistant, resistant to wear, and has high strength and durability.
- It is able to stand up to the repeated opening and closing cycles it must endure when used in a mechanical heart valve.
- It is commonly used for the inner orifice and the leaflets of bileaflet valves. The ATS Bi-leaflet valve shown here has leaflets made of pyrolytic carbon.

Dacron

- A material often used for the suture ring (which is used to attach the valve to the body) is Dacron.
- Dacron is a long chain polyester made from ethylene glycol and terephthalic acid.
- It is a synthetic fiber that has many uses in industry, including thermal insulation and sails for boats.

Teflon

- Another material that is commonly used for the suture ring is Teflon.
- Teflon is used in many medical applications because of its signature low coefficient of friction.

- Teflon is relatively inert and highly biocompatible. As with Dacron it is often used for vascular grafts.

3. **What is Rheology? Discuss about Rheological properties of blood. Or Discuss about the elements of blood and its characteristics.(N/D 2015)(N/D 2016)**

Rheology

- A branch of physics that deals with the deformation and flow of matter, especially the non-Newtonian flow of liquids and the plastic flow of solids.
- Rheological properties can be measured from bulk sample deformation using a mechanical rheometer or on a micro scale by using a micro capillary viscometer or an optical technique such as Microrheology.
- Newtonian fluids can be characterized by a single coefficient of viscosity for a specific temperature. Although this viscosity will change with temperature, it does not change with the strain rate. Only a small group of fluids exhibit such constant viscosity.
- The large class of fluids whose viscosity changes with the strain rate (the relative flow velocity) are called non-Newtonian fluids.
- Rheology generally accounts for the behaviour of non-Newtonian fluids, by characterizing the minimum number of functions that are needed to relate stresses with rate of change of strain or strain rates.
- For example, ketchup can have its viscosity reduced by shaking (or other forms of mechanical agitation, where the relative movement of different layers in the material actually causes the reduction in viscosity) but water cannot.
- Some other non-Newtonian materials show the opposite behaviour. Viscosity going up with relative deformation, which is called, shear thickening or dilatant materials. Since Sir Isaac Newton originated the concept of viscosity, the study of liquids with strain rate dependent viscosity is also often called Non-Newtonian fluid mechanics.
- The experimental characterization of a material's rheological behaviour is known as rheometry, although the term rheology is frequently used synonymously with rheometry.
- Theoretical aspects of rheology are the relation of the flow/deformation behaviour of material and its internal structure and the flow/deformation

behaviour of materials that cannot be described by classical fluid mechanics or elasticity.

- One of the major tasks of rheology is to empirically establish the relationships between deformations (or rates of deformation) and stresses.
- These experimental techniques are known as rheometry and are concerned with the determination with well-defined rheological material functions.

Blood Rheology

- Hemorheology, or blood rheology, is the study of flow properties of blood and its elements of plasma and cells.
- Alterations of these properties play significant roles in disease processes.
- Blood viscosity is determined by plasma viscosity, hematocrit (volume fraction of red blood cell, which constitutes 99.9% of the cellular elements) and mechanical properties of red blood cells.
- Red blood cells have unique mechanical behaviour, which can be discussed under the terms erythrocyte deformability and erythrocyte aggregation. Because of that, blood behaves as a non-Newtonian fluid.
- The viscosity of blood varies with shear rate. Blood becomes less viscous at high shear rates like those experienced in peak-systole. Contrarily, during end-diastole, blood moves more slowly and becomes thicker and stickier. Therefore, blood is a shear-thinning fluid.

Blood Viscosity

- Blood viscosity is a measure of the resistance of blood to flow.
- It can also be described as the thickness and stickiness of blood.
- This biophysical property makes it a critical determinant of friction against the vessel walls, the rate of venous return, the work required for the heart to pump blood, and how much oxygen is transported to tissues and organs.
- These functions of the cardiovascular system are directly related to vascular resistance, preload, after load, and perfusion, respectively.
- The primary determinants of blood viscosity are hematocrit, red blood cell deformability, red blood cell aggregation, and plasma viscosity.
- Plasma's viscosity is determined by water-content and macromolecular components, so these factors that affect blood viscosity are the plasma protein concentration and types of proteins in the plasma.

- Nevertheless, hematocrit has the strongest impact on whole blood viscosity. One unit increase in hematocrit can cause up to a 4% increase in blood viscosity.
- This relationship becomes increasingly sensitive as hematocrit increases. When the hematocrit rises to 60 or 70%, which it often does in polycythemia, the blood viscosity can become as great as 10 times that of water, and its flow through blood vessels is greatly retarded because of increased resistance to flow.
- This will lead to decreased oxygen delivery. Other factors influencing blood viscosity include temperature, where an increase in temperature results in a decrease in viscosity.
- This is particularly important in hypothermia, where an increase in blood viscosity will cause problems with blood circulation.

Blood Viscoelasticity

- Viscoelasticity is a property of human blood that is primarily due to the elastic energy that is stored in the deformation of red blood cells as the heart pumps the blood through the body.
- The energy transferred to the blood by the heart is partially stored in the elastic structure, another part is dissipated by viscosity, and the remaining energy is stored in the kinetic motion of the blood. When the pulsation of the heart is taken into account, an elastic regime becomes clearly evident.
- It has been shown that the previous concept of blood as a purely viscous fluid was inadequate since blood is not an ordinary fluid.
- Blood can more accurately be described as a fluidized suspension of elastic cells (or a sol).
- The red blood cells occupy about half of the volume of blood and possess elastic properties.
- This elastic property is the largest contributing factor to the viscoelastic behaviour of blood.

- The large volume percentage of red blood cells at a normal hematocrit level leaves little room for cell motion and deformation without interacting with a neighbouring cell.
- Calculations have shown that the maximum volume percentage of red blood cells without deformation is 58% which is in the range of normally occurring levels.
- Due to the limited space between red blood cells, it is obvious that in order for blood to flow, significant cell to cell interaction will play a key role.
- This interaction and tendency for cells to aggregate is a major contributor to the viscoelastic behaviour of blood.
- Red blood cell deformation and aggregation is also coupled with flow induced changes in the arrangement and orientation as a third major factor in its viscoelastic behaviour.
- Other factors contributing to the viscoelastic properties of blood is the plasma viscosity, plasma composition, temperature, and the rate of flow or shear rate. Together, these factors make human blood viscoelastic, non-Newtonian, and thixotropic.
- When the red cells are at rest or at very small shear rates, they tend to aggregate and stack together in an energetically favourable manner.
- The attraction is attributed to charged groups on the surface of cells and to the presence of fibrinogen and globulins.
- This aggregated configuration is an arrangement of cells with the least amount of deformation. With very low shear rates, the viscoelastic property of blood is dominated by the aggregation and cell deformability is relatively insignificant.
- As the shear rate increases the size of the aggregates begins to decrease. With a further increase in shear rate, the cells will rearrange and orient to provide channels for the plasma to pass through and for the cells to slide.
- In this low to medium shear rate range, the cells wiggle with respect to the neighbouring cells allowing flow.
- The influence of aggregation properties on the viscoelasticity diminish and the influence of red cell deformability begin to increase.

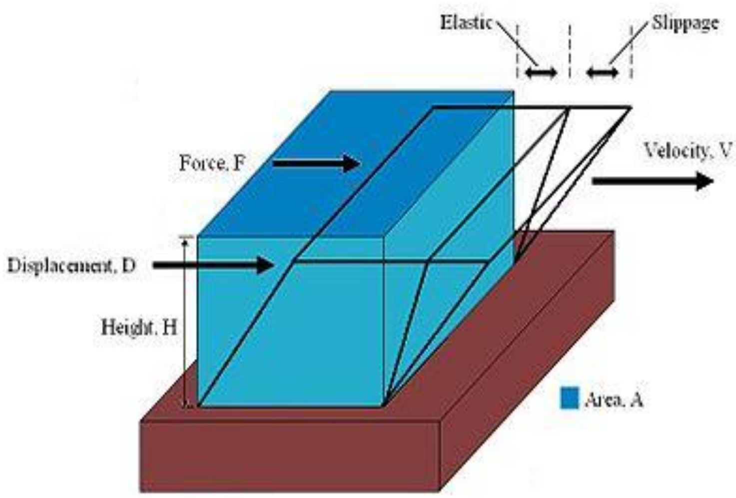
- As shear rates become large, red blood cells will stretch or deform and align with the flow. Cell layers are formed, separated by plasma, and flow is now attributed to layers of cells sliding on layers of plasma.
- The cell layer allows for easier flow of blood and as such there is a reduced viscosity and reduced elasticity.
- The viscoelasticity of the blood is dominated by the deformability of the red blood cells.

Maxwell Model

If a small cubical volume of blood is considered, with forces being acted upon it by the heart pumping and shear forces from boundaries. The change in shape of the cube will have 2 components:

- Elastic deformation which is recoverable and is stored in the structure of the blood.
- Slippage which is associated with a continuous input of viscous energy.

When the force is removed, the cube would recover partially. The elastic deformation is reversed but the slippage is not. This explains why the elastic portion is only noticeable in unsteady flow. In steady flow, the slippage will continue to increase and the measurements of non time varying force will neglect the contributions of the elasticity.



Shear Stress: $\tau = \frac{F}{A}$

Shear Strain: $\gamma = \frac{D}{H}$

Shear Rate: $\dot{\gamma} = \frac{V}{H}$

A sinusoidal time varying flow is used to simulate the pulsation of a heart. A viscoelastic material subjected to a time varying flow will result in a phase variation between τ and γ represented by ϕ. If $\phi = 0$, the material is a purely elastic because the stress and strain are in phase, so that the response of one caused by the other is immediate. If ϕ = 90°, the material is a purely viscous because strain lags behind stress by 90 degrees. A viscoelastic material will be somewhere in between 0 and 90 degrees.

Relating the equations to common viscoelastic terms we get the storage modulus,G', and the loss modulus,G".

$$G = G' + iG''$$

A viscoelastic Maxwell material model is commonly used to represent the viscoelastic properties of blood. It uses purely viscous damper and a purely elastic spring connected in series. Analysis of this model gives the complex viscosity in terms of the dashpot constant and the spring constant.

4. Distinguish between mechanical and biological valve?(N/D 2016)

Mechanical valve	**Biological valve**
• Mechanical valves are made completely from man-made	• Biological valves are made from human or animal tissue that has

materials. • They have made up of very light weight and long lasting materials. • The main advantage of mechanical valves is that they last a very long time in most cases 20 years or more. • Ex: Caged ball, Tilting disk, Single leaflet, Bileaflet.	been specially treated so that your body does not reject the valves. • After being treated, the valves are attached to man-made materials to give them support. • They do not last as long as man-made valves, although they probably last longer in older people. • Ex: Homograft or allograft, Xenograft, Bovine, Porcine.

5. Write short notes on current trends in heart valve design (N/D 2016)

Tissue-engineered heart valves

Currently, the following approaches are used for developing tissue-engineered valves :

Decellularized xenogenic tissues for cell seeding or direct implantation

- It has been assumed that the antigenicity of xenogenic tissues, such as bovine or porcine pericardium, porcine small intestinal submucosa, and porcine or ovine heart valves is caused by cell debris.
- In addition, porcine aortic valves need to be cross-linked by glutaraldehyde before implantation, which can cause cytotoxic and other adverse reactions.
- Glutaraldehyde-fixed pericardium was resistant to collagenase degradation and cell infiltration, but significantly prone to increased calcification.
- Homografts (i.e. allografts), which do not require glutaraldehyde cross-linking, can last 15-20 years and have better mechanical properties than a cellular glutaraldehyde-fixed xenografts.
- The decellularization technique has a different impact on tissue preservation and valve efficiency.
- Aggressive decellularization by enzymes (DNAase, RNAse), detergents, spit freezing and radiation, used in SynerGraft technology, resulted in catastrophic failure of decellularized xenogenic heart valves.

- On the other hand, deoxycholic acid completely decellularized heart valves, while structural proteins were retained and appeared to be intact.
- Implantation of decellularized allograft and xenograft valves induces tissue regeneration *in vivo* with efficient repopulation of the matrix by interstitial cells, but graft endothelialization is not sufficient.
- When MSC were seeded onto a decellularized porcine pulmonary valve pre-coated with fetal bovine serum and fibronectin, they colonized the scaffold, and endothelium lining, and the presence of fibroblasts and myofibroblasts.
- Another approach in engineering heart valves is the use of modified decellularized bovine.

Constructs containing polymerized extracellular matrix and entrapped cells:

- Type I collagen is an important component of the vessel wall, and a major matrix protein of the bone and other connective tissues.
- Collagen as a biomaterial has a number of useful properties.
- It is biodegradable, biocompatible, non-antigenic, haemostatic, easily modifiable, acts synergetically with other bioactive components, is biologically plastic due to its high tensile strength and minimal expressibility, and is compatible with synthetic polymers.
- Pure collagen, due to its fast degradation, often needs crosslinking or needs to be used in a blend with other polymers.
- Heart valve-shaped collagen scaffolds, prepared from decellularized porcine pericardium treated with penta-galloyl glucose.
- Collagen scaffolds composed of 1-5 % collagen prepared by rapid prototyping allowed adhesion, vascular grafts, heart valves, skin substitutes, or elastic cartilage.
- Elastin and also collagen may serve as nucleation sites for mineralization, independent of cellular components of the prostheses.
- Calcification can be reduced by treatment with aluminium chloride, ethanol/EDTA, the presence of glycosaminoglycans, or basic fibroblast growth factor.

- Scaffolds prepared from insoluble collagen and solubilized elastin induced angiogenesis, and increased elastic fiber synthesis without inducing calcification
- Disadvantages of fibrin include its poor mechanical properties, relatively quick degradation, and shrinkage due to structural changes and contraction of the newly synthesized collagen bundles.
- Pure collagen had the highest linear modulus and pure fibrin the lowest, while collagen-fibrin mixtures underwent the highest compaction.
- Dynamic conditioning of molded fibrin-based heart valve seeded with a mixture of smooth muscle cells and fibroblasts from ovine carotid artery in a bioreactor increased the cell attach
- In comparison with nature-derived materials, synthetic polymers are of more defined and controllable properties, and can be more easily reproduced.

- The aortic heart valve has a specific shape and undergoes a specific load during the cardiac cycle.
- Systole and diastole lead to different changes in shear stress and pressure load that bend the valve and subject the valve to tensile and compressive forces.
- Dynamic cultivation is also a part of a novel approach to valve tissue engineering, known as “in-body tissue architecture technology.

6. When the blood is said to be Newtonian? Explain with necessary equations. (A/M 2017)

The Newtonian Viscous Fluid

A Newtonian viscous fluid is a fluid for which the shear stress is linearly proportional to the strain rate. For a Newtonian fluid the stress-strain relationship is specified by the equation

$$\sigma_{ij} = -p\delta_{ij} + D_{ijkl} V_{kl} \qquad (1)$$

where σ_{ij} is the stress tensor,

V_{kl} is the strain rate tensor,

D_{ijkl} is a tensor of viscosity coefficients of the fluid, and

p is the static pressure.

- The term $-p\delta_{ij}$ represents the state of stress possible in a fluid at rest (when $V_{kl}= 0$).
- The static pressure p is assumed to depend on the density and temperature of the fluid according to an equation of state.For a Newtonian fluid we assume that the elements of the tensor Dijkl may depend on the temperature but not on the stress or the rate of deformation.
- If the fluid is *isotropic,* i.e., if the tensor *Dijkl* has the same array of components in any system of rectangular cartesian coordinates, then *Dijkl* can be expressed in terms of two independent constants λ and μ,

$$Dijkl = \lambda\delta_{ij}\delta_{kl} + \mu(\delta_{ik}\delta_{jl} + \delta_{il}\delta_{jk}) \qquad (2)$$

And we obtain

$$\sigma_{ij} = -p\delta_{ij} + \lambda V_{kk}\delta_{ij} + 2\mu V_{ij} \qquad (3)$$

A contraction of Eq. (3) gives

$$\sigma_{kk} = -3p + (3\lambda + 2\mu) V_{kk} \qquad (4)$$

If it is assumed that the mean normal stress $1/3\sigma_{kk}$ is independent of the rate of dilation V_{kk} then we must set

$$3\lambda + 2\mu = 0; \qquad (5)$$

and the constitutive equation becomes

$$\sigma_{ij} = -p\delta_{ij} + 2\mu V_{ij} - 2/3\,\mu V_{kk}\delta_{ij} \qquad (6)$$

This is called stokes fluid for which one material constant μ , the *coefficient of viscosity,* suffices to define its property.

If a fluid is *incompressible,* then $V_{kk} = 0$, and we have the constitutive equation for an incompressible viscous fluid:

$$\sigma_{ij} = -p\delta_{ij} + 2\mu V_{ij} \qquad (7)$$

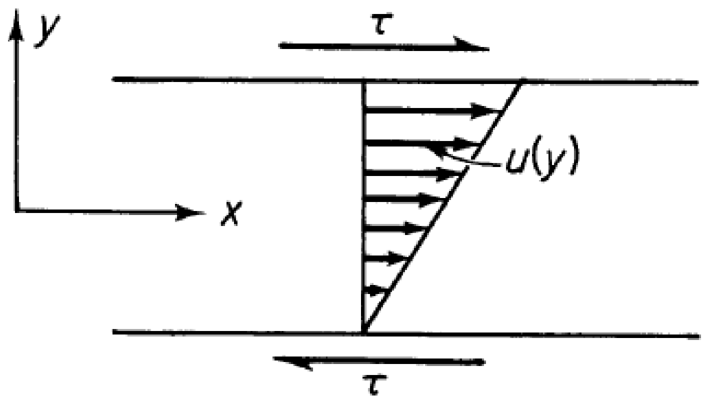
If μ = 0, we obtain the constitutive equation of the nonviscous fluid:

$$\sigma_{ij} = -p\delta_{ij} \quad (8)$$

Newton's concept of viscosity may be explained in the simplest case of a shear flow with a uniform velocity gradient. Newton proposed the relationship

$$\tau = \mu du/dy \quad (9)$$

for the shear stress τ, where μ is the coefficient ofviscosity. **In** the centimetergram-second system of units, in which the unit of force is the dyne, the unit ofμ is called a *poise,* in honor of Poiseuille. **In** the **SI** system, the unit of viscosity is newton-second per square meter *(Ns/m^2).*



Newtonian concept of viscosity

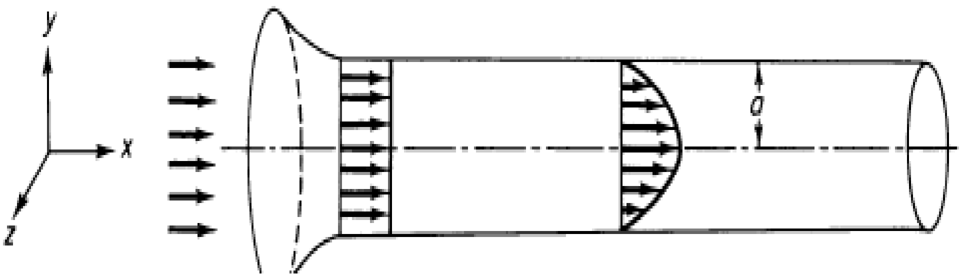
The viscosities of air and water are small, being 1.8 x 10^{-4} poise for air and 0.01 poise for water at atmospheric pressure and 20°e. At the same temperature the viscosity of glycerin is about 8.7 poise. The viscosity of liquids decreases as temperature increases. That of gases increases with increasing temperature.

7. Prove that the velocity profile of blood in the blood vessel is parabola. (A/M 2017)

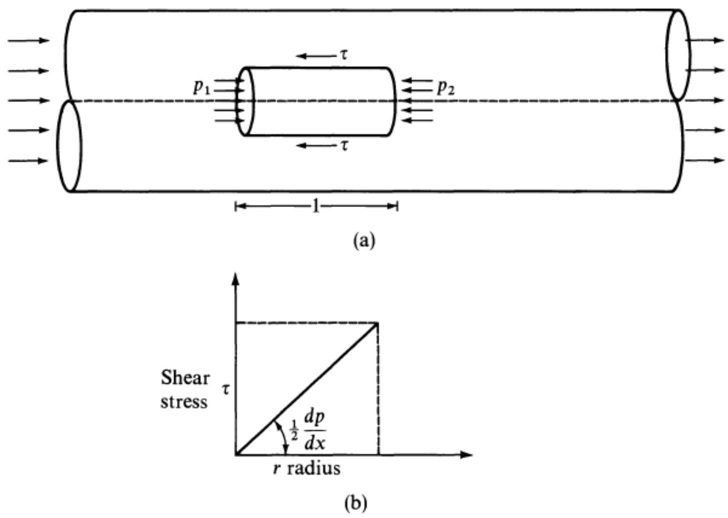
Laminar Flow of Blood in a Tube

- Let us consider the flow of blood in a circular cylindrical tube. Assume the flow to be *laminar,* that is, *not turbulent.*

- Also assume that the tube is long and the flow is steady, so that the conditions of flow change neither with the distance along the tube, nor with the time.
- The polar axis coincides with the axis of the cylinder. The flow obeys Navier-Stokes equations of motion of an incompressible fluid.
- The boundary condition is that blood adheres to the tube wall (the so-called *no-slip* condition).
- Since the boundary condition is axisymmetric, the flow is also axisymmetric and the only nonvanishing component ofvelocity is *u(r)* in the axial direction; *u(r)* is a function of *r* alone, and not of *x.*



Velocity profiles in a steady laminar flow into a circular cylindrical tube.



Steady flow in a long circular cylindrical pipe. (a) A free-body diagram of a centrally located element on which pressure and shear stresses act. (b) Relationship between the shear stress rand the radial distance *r* from the axis of symmetry.

- Isolate a cylindrical body of fluid of radius *r* and unit length in the axial direction. This body is subjected to a pressure P_1 on the left-hand end, P_2 on the right-hand end, and shear stress τ on the circumferential surface. SinceP_1 – P_2 = -1 . *(dp/dx)* acts on an area πr^2 , and τ acts on an area 1 . *2nr,* we have, for equilibrium, the balance of forces

$$\tau \cdot 2\pi r = -\pi r^2 \frac{dp}{dx},$$

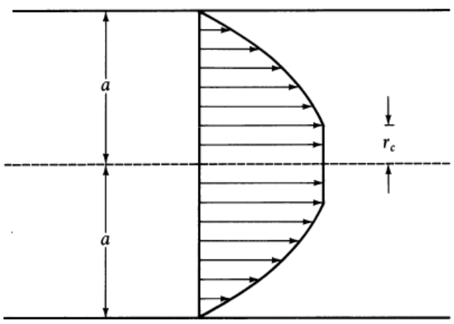
or

$$\tau = -\frac{r}{2}\frac{dp}{dx} \qquad \text{(Stokes, 1851).} \qquad (1)$$

Now we must introduce a constitutive equation that relates the shear stress τ to the velocity gradient. Let us first consider a Newtonian fluid.

A Newtonian Fluid

By the definition of Newtonian viscosity, we have



The velocity profile of laminar flow of blood in a long circular cylindrical pipe.

$$\tau = \mu \frac{du}{dr}. \qquad (2)$$

A substitution of Eq. (2) into Eq. (1) yields

$$\frac{du}{dr} = -\frac{r}{2\mu}\frac{dp}{dx}. \qquad (3)$$

Since the left-hand side is a function of *r,* the right-hand side must be also. Hence *dp/dx* cannot be a function of x. But since the fluid does not move in the radial direction, the pressures in the radial direction must be balanced, and *p* cannot vary with *r.* Hence the pressure gradient *dp/dx* must be a constant. Therefore, we can integrate Eq. (3) to obtain

$$u = -\frac{r^2}{4\mu}\frac{dp}{dx} + B, \tag{4}$$

where *B* is an integration constant. *B* can be determined by the boundary condition of no-slip:

$u=0$ when $r = a$. (5)

Combining Eqs. (4) and (5) yields the solution

$$u = \frac{1}{4\mu}(a^2 - r^2)\frac{dp}{dx}, \tag{6}$$

which shows that the velocity profile is a parabola

8. Write notes on testing of artificial heart valves.

The testing of the artificial heart valves can be done by the following methods:

- ✓ In vitro testing of artificial heart valves
- ✓ Durability test
  - Durability test system CVE FT-2
  - Durability test system CVE-FT3
- ✓ Testing of polymer-based heart-valve prosthesis

1) In vitro testing of artificial heart valves

- In vitro testing of artificial heart valves is often performed with simple fluids like glycerol solutions. Blood, however, is a non-Newtonian fluid with a complex viscoelastic behavior, and different flow fields in comparable geometries may result. Therefore, we used different polymer solutions (Polyacrylamid, Xanthan gum) with blood-like rheological properties as well as various Newtonian fluids (water, glycerol solutions) in our heart valve test device.

- Hydrodynamic parameters of Björk-Shiley heart valves with a tissue annulus diameter (TAD) of 21-29 mm can be investigated under aortic flow conditions. Systolic pressure difference, closing time and closing volume depend on TAD and the pressure differences across the valve. In contrast, rheological behavior has a pronounced influence upon leakage flow and leakage volume, respectively. Results show furthermore that the apparent viscosity data as a function of shear rate are not sufficient to characterize the rheological fluid behavior relevant to hydrodynamic parameters of the heart valves investigated.

2) Durability test

The assessment of the durability of heart valve prostheses according to the valid standards ISO 5840 and FDA is one focus in heart valve testing at the department of Cardiovascular Engineering (CVE).

- **Durability test system CVE FT-2**

In the totally revised and improved durability test system CVE FT-2 12 heart valve prostheses can be assessed simultaneously. The prostheses are mounted into separate test compartments, which are arranged circular on a mounting plate.

Separate test compartments permit an individual adjustment of each test prostheses. A motor driven swash plate generates a sinusoidal flow through the test valve by compression and extension of a metallic bellow mounted on the bottom side of the mounting plate. The valve is closed and fluid from the upper chamber flows to the lower chamber via an adjustable bypass. Pressures are measured directly upstream and downstream the test prostheses. The differential pressure across the closed prosthesis is calculated and can be controlled by adjustment of the bypass throttle. Stroke volume and test frequency are constant for all test compartments. The test prostheses can be observed by means of a triggered stroboscope. By means of a high speed video system a more detailed documentation of the prostheses' motion can be performed.

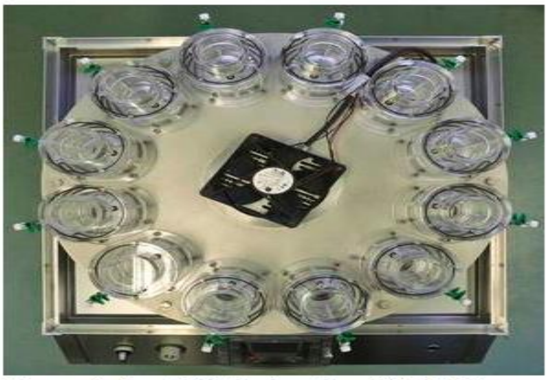


Figure 2: Durability test system CVE-FT2

- **Durability test system CVE-FT3**

CVE-FT3 is a third durability test system. The main drive mechanism is equal to CVE-FT2. Also in CVE-FT3 12 heart valve prostheses can be tested simultaneously in separate compartments. The difference between the two test systems is, that in the CVE-FT3 the movement of the swash plate is transferred directly to the heart valve prostheses.

The prostheses are moved through the fluid inside the test compartments, whereby an sinusoidal opening and closing is generated. The pressure measurement, the adjustment of the differential pressure across the closed valve and the documentation of the valve behavior is equal to the CVE-FT2.

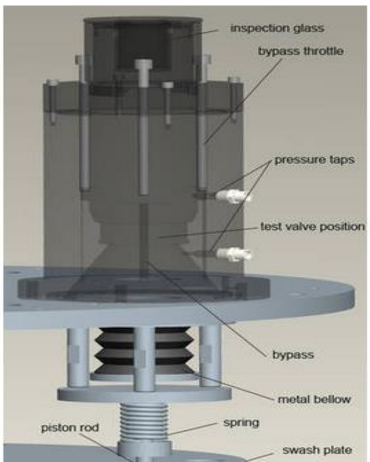


Figure 3: Testing unit CVE-FT2

3) Testing of polymer-based heart-valve prosthesis

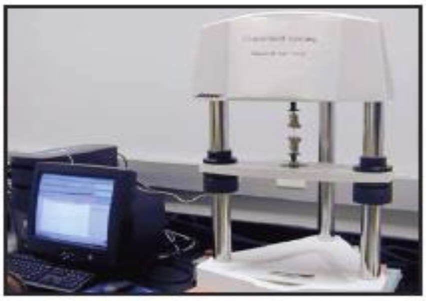
Heart valve prostheses have been used successfully in heart procedures since 1960 resulting in an overall improvement in the quality of life of the patients. Currently, there are two kinds of valves used: mechanical and bioprosthetic. Generally, mechanical valves are more durable than bioprosthetic valves. However, they sometimes involve side effects with irregular blood flow and clotting of blood around them. Bioprosthetic valves have better hemodynamic (blood flow) properties, but are more susceptible to wear as a result of material fatigue. Polymer trileaflet (PT) valves offer natural hemodynamic with the potential for better durability.

In order to identify the better material from which to manufacture valves, a certain proprietary polymer is compared to an existing implant-approved polymer (IAP). Static and dynamic properties of the polymers are being determined in order to establish the right polymer composite for the heart valve prosthesis.

Tensile and tension fatigue properties for each material are performed according to ASTM (American Society for Testing and Materials) standards. The tensile test is displacement controlled, and the specimen is stretched at a constant rate until failure. The tension fatigue test is load controlled; that is, the specimen is being cycled between two tensile loads. The loading frequency is 100 Hz.

Cycling continues until failure for each specimen. Since there can be significant fatigue damage without actual fracture, failure is defined as 50% loss in residual strength of the material.

Stretched Specimen



The ElectroForce® 3200 Test Setup

9. Describe structure and functions of blood vessels.

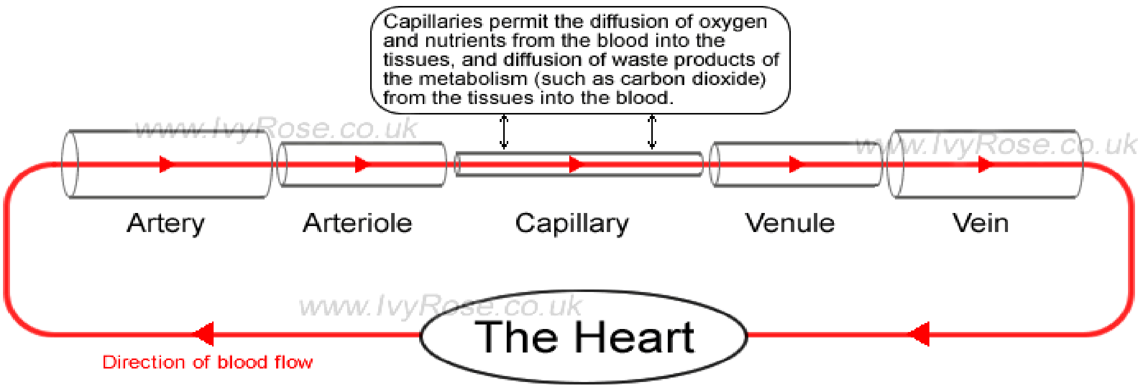
The major roles of the blood include:

- . Carries oxygen and nutrients to active tissues.
- Delivers carbon dioxide to the lungs
- Brings metabolic end products to kidneys
- Blood is a buffering reservoir that controls pH of bio-fluids
- Plays a major role in the body's immune system

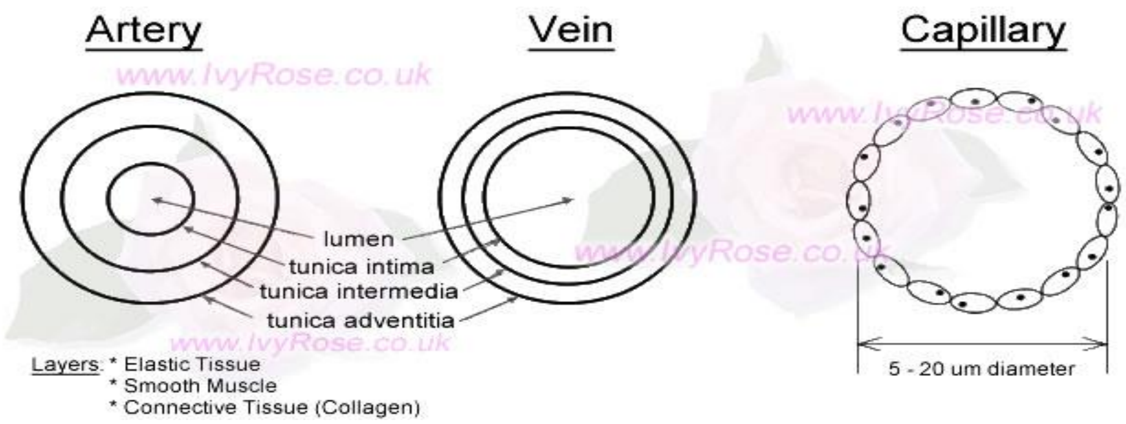
The main types of blood vessels are:

- Arteries
- Arterioles
- Capillaries
- Venules
- Veins

The following diagram summarises the sequence of blood flow through the heart, arteries, arterioles, capillaries, venules, veins, then back to the heart:



The following diagram summarizes the structural differences between different types of blood vessels.



Structure and Functions of Blood Vessels

	Structure	Functions
Arteries	The walls (outer structure) of arteries contain smooth muscle fibre that contract and relax under the instructions of the sympathetic nervous system.	• Transport blood away from the heart • Transport oxygenated blood only (except in the case of the pulmonary artery).
Arterioles	Arterioles are tiny branches of arteries that lead to capillaries. These are also under the control of the sympathetic nervous system, and constrict and dilate, to regulate blood flow.	• Transport blood from arteries to capillaries • Arterioles are the main regulators of blood flow and pressure.

Capillaries	Capillaries are tiny (extremely narrow) blood vessels, of approximately 5-20 micrometers (one micrometer = 0.000001metre) diameter. There are networks of capillaries in most of the organs and tissues of the body. These capillaries are supplied with blood by arterioles and drained by venules. Capillary walls are only one cell thick (see diagram), which permits exchanges of material between the contents of the capillary and the surrounding tissue.	• Function is to supply the tissues of the body with the components of blood, and (carried by the blood), and also to remove waste from the surrounding cells ... as opposed to simply moving the blood around the body (in the case of other blood vessels) • Exchange of oxygen, carbon dioxide, water, salts, etc., between the blood and the surrounding body tissues.
Venules	Venules are minute vessels that drain blood from capillaries and into veins. Many venules unite to form a vein.	• Drains blood from capillaries into veins, for return to the heart.
Veins	The walls (outer structure) of veins consist of three layers of tissues that are thinner and less elastic than the corresponding layers of arteries. Veins include **valves** that aid the return of blood to the heart by preventing blood from flowing in the reverse direction.	• Transport blood towards the heart. • Transport deoxygenated blood only (except in the case of the pulmonary vein).

Comparison between Arteries and Veins

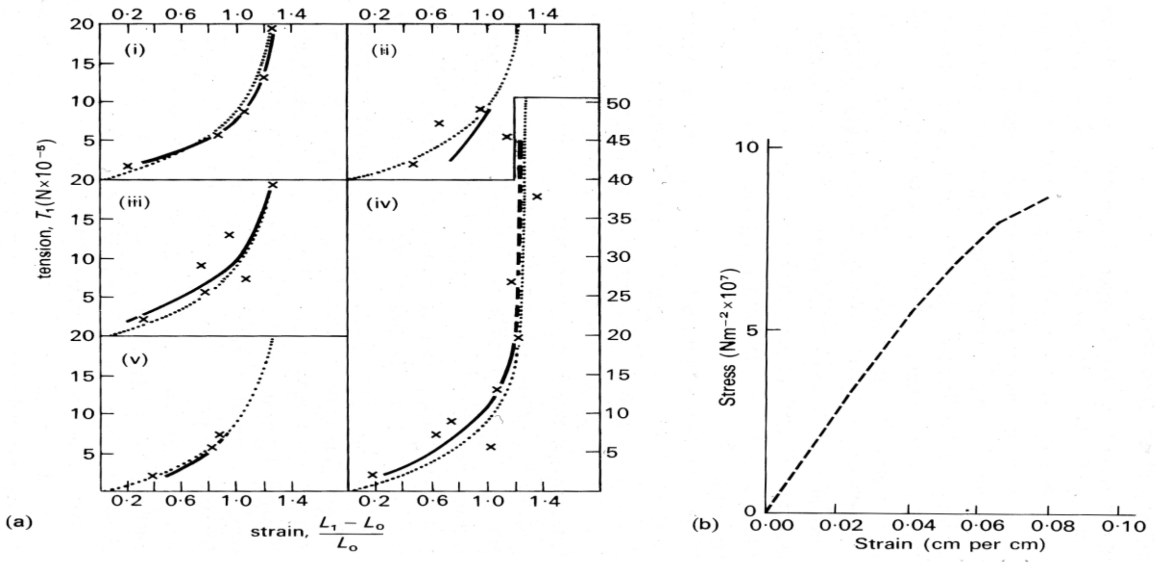
Arteries	Veins
• Transport blood away from the heart	**7.** Transport blood towards the heart
• Carry Oxygenated Blood (except in the case of the Pulmonary Artery)	**8.** Carry De-oxygenated Blood (except in the case of the Pulmonary Vein)
• Have relatively narrow lumens (see diagram above)	**9.** Have relatively wide lumens (see diagram above)
• Have relatively more muscle/elastic tissue	**10.** Have relatively less muscle/elastic tissue
• Transports blood under higher pressure (than veins)	**11.** Transports blood under lower pressure (than arteries)
• Do not have valves (except for the semi-lunar valves of the pulmonary artery and the aorta).	**12.** Have valves throughout the main veins of the body. These are to prevent blood flowing in the wrong direction, as this could (in theory) return waste materials to the tissues.

10**.** Explain material properties & modeling of blood vessel system

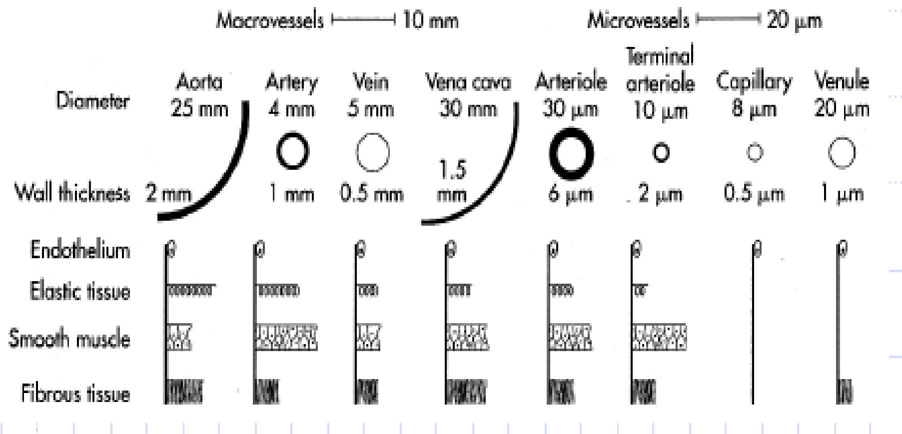
PROPERTIES OF BLOOD VESSEL:

- Homogenous and isotropic assumptions.
  - do not vary within the material.
  - Independent of direction in which stress is applied.

- Blood vessel walls
    - 70% of walls consist of water (incompressible & inelastic).
    - Rest of material: elastic fibers
    - Three types of fibers:
        - Collagen
        - Elastin
        - Smooth muscle.



- **Non-linear behavior.**
    - Measure effective Young's modulus (E_{eff}) by measuring local slope.



Endothelium: Single cell layer separating wall and blood

Elastic tissue: Elastin

Fibrous tissue: Collagen

- **Viscoelastic behavior**
    - Purely elastic material:
        - No phase lag between strain and stress

- **Viscoelastic material**
    - Strain lags stress.
    - Possible to represent major features of viscoelastic behavior via a Dynamic Young's modulus (E_{dyn}) and effective viscosity (h).

MODELING OF BLOOD VESSEL SYSTEM

The essentiality of experimental and speculative investigate on is on blood vessels of the cardiovascular system. The modeling evaluate numerically the blood vessels of the cardiovascular system using Inlet section, Reinhold's number, rate of flow and area of cross-section. Here the modeling of blood vessel by using fractals.

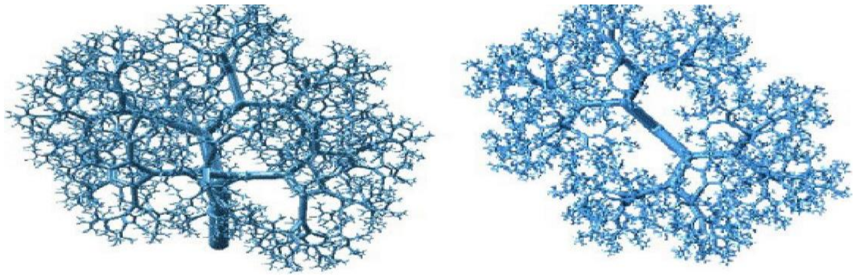
Fractals

A fractal is an object or quantity that displays self-similarity, in a somewhat technical sense, on all scales. The object need not exhibit exactly the same structure at all scales, but the same "type" of structures must appear on all scales.

Modeling of the blood vessels system through Fractals

In real blood vessels system, vessel walls are elastic and change its diameters. In this way resistance of blood vessel system is regulated. This process is known as auto regulation and corrects nutrition of all cells in human body. Blood flow estimation assumes laminar flow for the entire fractal vascular tree.

The blood vessel of the human cardiovascular system is modeled through the fractal vascular tree like structure is shown below. The basic idea of fractal is that breaking down an object into smaller parts and each of which resembles the whole has been invoked for the blood vessels.



Structure of the Fractal vascular tree

Methods

Blood flow in the cardiovascular system has been precisely calculated by the following methods. These results are very useful to investigate the defects in the above system. The following methods are the most important parameters in transport vessel tree.

Rate of flow (Q)

It is defined as the quantity of the fluid flowing per second through a section of a pipe or a channel. For an incompressible fluid the rate of flow is expressed as the volume of fluid flowing across the section per second. Mathematically the rate of flow is defined as

$$Q = AxV \qquad \text{.....(1)}$$

where Q is the Rate of blood flow *(mi/sec), A* is the cross-sectional area (cm^2), V is the velocity of the blood.

Reynold's number

Using the Reynold's equation we can see that a large diameter, with rapid flow, where the density of the blood is high tends towards turbulence. Rapid changes in vessel diameter may lead to turbulent flow, for instance when a narrower vessel widens to a larger one. When normal laminar blood flow becomes turbulent flow

- The rate of blood flow (i.e.,) the velocity of flow, is high
- It passes by an obstruction in a vessel (as in case of compression by cuff of Sphygmomanometer).
- It makes a sharp turn.
- It passes over a rough surface

Turbulence is also related to the diameter of the vessel and the viscosity of the blood which can be expressed by ratio of inertia to viscous forces.

$$R_e = \frac{\rho VL}{\mu} \qquad \text{.....(2)}$$

Where R_e is the Reynold's number, L is the length of the blood vessel, V is the Average velocity of the blood. This is the most important dimensionless number; it describes the fluid flow regime. If R_e is higher, there is a greater probability of the turbulence.

Inlet Section

The blood flow in the blood vessel is mainly concerned with Inlet section. The length of the section is proportional to vessel diameter, Reinhold's number and coefficient $l_e = AdR_e$ …..(3)

Where l_e is the inlet section, X = 0.056 a value from Navier-Stokes equation, d diameter of the blood vessel and R_e Reynold's number.

Note: The Kind of flow depends on value of R_e.

1. If Re < 2000 the flow is Laminar
2. If Re > 4000 the flow is turbulent
3. If 2000 < Re < 4000 it is called transition flow.

A synthetic view of human cardiovascular system has been exhibited with a remarkable variety of components is given in the following table for the normal Human being.

Data Report of Normal Human Being

Vessel	Radius (r)(cm)	Total Cross Section (A) (cm^2)	Wall thickness (h) (cm)	Average Velocity (V) (cm/sec)	Length (L) (cm)
Aorta	1.25	4.9	0.2	21.25	50
Arteries	0.2	20	0.1	60xl0~2	50
Arterioles	1.5xl0~3	400	2xl0~3	90x10^	1
Capillaries	3X10"4	4500	lxlO-4	160xl0~3	0.1
Venules	lxlO-3	4000	2X10"4	65xl0~3	0.2
Veins	0.25	40	0.05	50xl0~2	2.5
Venae ava	1.5	18	0.15	2.36	50

Data Report of Ischemic heart disease

Vessel	Radius (r) (cm)	Total Cross Section (A) (cm^2)	Wall thickness (h) (cm)	Average Velocity (V) (cm/sec)	Length (L) (cm)
Aorta	0.03	0.03	0.95	43.7	43
Arteries	0.29	128	0.76	42.5	43
Arterioles	1.4xl0~^{3}	350	2xl0~^{3}	90x10^	1
Capillaries	3X10"^{4}	4500	lxlO^{-4}	160xl0~^{8}	0.1
Venules	lxlO^{-3}	4000	2X10"^{4}	65xl0~^{8}	0.2
Veins	0.77	400	0.11	20.4	1.5
Venacava	0.7	3	0.15	11	43

11. Explain the development of artificial heart valves.

Introduction

The heart is a vital muscular organ which pumps blood around the body. Valves are tissue flap components of the heart which react to inertial forces exerted by surrounding blood by opening and closing. When functioning correctly, valves ensure unidirectional blood flow during the cardiac cycle, with maximized flow rate and minimized flow resistance. The four valves of the human heart are the aortic, mitral, pulmonary and tricuspid valves.

Heart valves can become defective through congenital disease, where abnormalities in the valve develop before birth, or acquired disease, where problems develop within valves that once functioned normally. Incorrect function of heart valves affects hemodynamic (blood flow) performance, primarily via valvular stenosis in forward flow and valvular incompetence (regurgitation) in reverse flow. Valvular stenosis is the narrowing of the valve which increases resistance to blood flow, while valvular incompetence refers to the failure of valve leaflets to close fully generating reverse blood flow. Both conditions result in decreased efficiency of the heart, and it is often necessary to repair or replace the native valve to reestablish correct valve function. Artificial valves have been developed to replace defective native valves, and fall into two main categories: mechanical and bioprosthetic.

Since the introduction of artificial heart valve replacements there have been significant improvements in their function and design. Artificial valves, however, remain inferior in both aspects to native valves, warranting further development.

Computational models can be used to predict and validate function of valve designs and highlight possible problems, aiding the development of the prostheses.

Mechanical Heart Valves

Mechanical heart valves are based on the occluder principle. Each valve has an occluder which determines whether the valve is open or closed. The position of the occluder depends on the blood pressure on either side of it and helps to maintain unidirectional flow. The three main subtypes of mechanical heart valves are the ball-and-cage, tilting disc and bileaflet.

Challenges of Mechanical Heart Valves

The main advantage of mechanical over other types of artificial valves is their durability. Due to the properties of materials used in their manufacture, mechanical valves can often last the lifetime of the patient. They are therefore favored over other types of valves in pediatric and adolescent patients, although the inability of the valve to develop and repair is a major drawback for these patient groups. Issues which arise with implantation of mechanical valves involve both engineering and biological problems. Since the system that the artificial valve is implanted into is composed of living cells, the interaction between the biology of the system and the mechanics of the valve is an important issue. The implanting of a prostheses into the heart can initiate cascades of biological events, such as formation of blood clots, or result in a change of function or failure of the heart or valve.

Cavitation is the rapid formation of vaporous bubbles in a fluid, caused by a reduction in local pressure below the fluid vapour pressure. When the pressure recovers the formed bubbles collapse producing thermal and pressure shock waves which damage solid material near the cavitation site. This damage has been identified as erosion and pitting (the formation of cavities in the material) on the housings and leaflets of implanted mechanical valves.

Thrombosis and Embolisms Thrombosis is another adverse effect of mechanical valves, and refers to the formation of a stationary thrombus, or blood clot, along the wall of a blood vessel. The detachment of such a clot from the vessel wall, referred to as an embolism, can lead to an obstruction of a blood vessel elsewhere in the circulatory system, which can lead to conditions such as myocardial infarction or ischemic stroke.

Hemorrhage due to thromboembolic risks associated with mechanical valves, patients are prescribed oral anticoagulation (OAC) therapy. While this minimizes thrombus formation, the therapy increases the risk of hemorrhage. Although improvements in mechanical valve design have resulted in a decrease in the amount of anticoagulant required, OAC therapy is still an integral part of mechanical heart valve implantation.

An important test of valve design is hemodynamic performance, affected by the geometry of the valve, such as hinge design, leaflet shape and opening angle, and the manufacturing material. Materials considered must satisfy several requirements including biocompatibility, and biological and chemical stability. Biocompatibility ensures that the biological system in question can endure implantation of the prosthesis without destruction of tissue. Biological and chemical stability ensure that the material is resistant to degradation due to biological interactions with the body and environmental agents encountered during implantation, respectively. If the material used does not satisfy these requirements, inflammatory diseases, such as endocarditic, can develop and potentially lead to valve or heart failure. Mechanical properties of the material must also be considered, to ensure durability and correct function of the valve.

Initial valve designs used metals such as stainless steel. However, due to the polycrystalline nature of metals, issues such as material fatigue led to the failure of some replacement valves. Material fatigue is structural damage caused by cyclic loading of the material, which accumulates and may lead to the formation of cracks and failure.

The fatigue of metals was partially addressed by using alloys and strong metals such as titanium, however the introduction of pyrolytic carbon as a material for artificial valves largely overcame this problem.

Pyrolytic carbon is a biocompatible material with thromboresistant properties, which makes it suitable for use in mechanical valve production. The material also presents good durability, wear resistance and strength, all of which are important to minimize possible mechanical failure of the valve. Pyrolytic carbon was initially used only for the manufacture of occluder discs, with the struts and housing made of metal alloys, with the first bileaflet valves comprising an all-pyrolytic carbon design.

Bioprosthetic and Tissue Engineered Heart Valves

While the development of mechanical valves is an important issue for certain patient, other types of valves have also been developed. Bioprosthetic valves are a major group of artificial valves, while tissue engineered heart valve are at the forefront of current research.

- **Bioprosthetic Heart Valves**

Bioprosthetic heart valves use tissue from animal or human sources to replace defective native valves. Natural heart valves work on the principle of maximal central flow, with blood flowing through the centre of the valve in order to maximize flow rate and minimize flow resistance. Bioprosthetic artificial valves attempt to mimic this design and are generally made up of three tissue leaflets, homologous to native valves. The sourced leaflets are fixed in gluteraldehyde solution, which also decreases biodegradation, and attached to an annular housing which functions much like the native aortic annulus.

Bioprosthetic valve replacements are grouped into **xenografts** and **allografts**. Xenografts are replacements using tissue valves sourced from animals, usually porcine aortic valve leaflets or either bovine or equine pericardium. Porcine aortic valves are chosen due to anatomical similarities with native valves, while bovine tissue presents better durability .The use of already formed porcine valves simplifies the replacement valve production process, since the need to produce individual cusps is avoided.

Allografts are replacements using valves sourced from human cadavers. In this case, the valve is geometrically similar to the native valve, but the leaflets stiffen and annulus shrinks slightly due to cryopreservation. Allograft valves often fail due to structural damage which occurs as a result of the functioning of the valve. There are also issues with availability of cadaveric valves, especially of the correct size and geometry.

In order to implant bioprosthetic valves, support is required for the leaflets, achieved by both stented and non-stented methods. Stented valves are generally xenografts, and both porcine and pericardial tissue is used. A metal or polymer stent is used to support the leaflets at the commissures (joins). Stented valves have been reported to be relatively stenotic compared to native valves, due to construction of the leaflets, the stent restricting full opening of the leaflets and the stiffness of the fixed tissue. Non-stented valve implants, the necessary support is provided by attaching the valve to the aorta.

Developments are also being made on valves which can be implanted via percutaneous methods, whereby a collapsed valved stent is delivered to the implantation site by catheter and balloon-expanded to open out the valve. The development of these valves depends largely on the availability of suitable biomaterials for use in collapsable stents. Collapsable valves that can be compressed into a catheter, yet still be able to unfold into the precise geometry required for them to function correctly, are also necessary.

- **Tissue Engineered Heart Valves**

A limitation of mechanical and bioprosthetic valves is the lack of living tissue, deeming the valves incapable of repair, growth or remodeling. Heart valve tissue engineering (HVTE) is an area of regenerative medicine concerned with the use of autologous cells and bioengineered materials to develop replacement valves which have the structure and function of native valves.

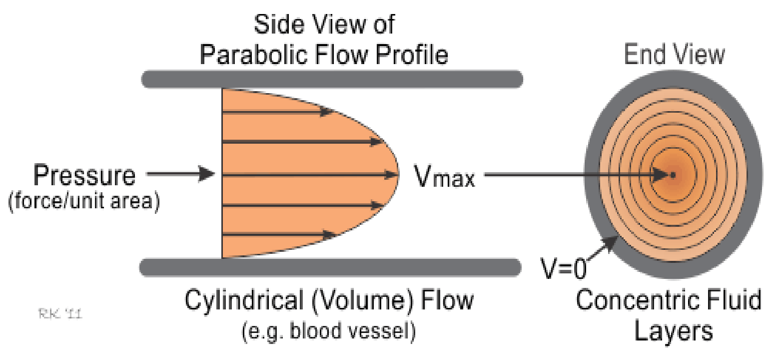
Native valves have a “layered architectural” structure constituting of the extracellular matrix (ECM) and specialized cells.

The ECM is the ‘frame’ on which cells form tissue, and gives the tissue the necessary structural and mechanical properties. The cells involved include valvular endothelial cells (VECs) and valvular interstitial cells (VICs). VECs line the valve surface to provide the necessary thromboresistance and regulate immune and inflammatory reactions, while VICs act to remodel the ECM and repair functional damage, thus determining valve durability.

Tissue engineering attempts to form functional tissue capable of remodeling and growth based on this structure. One approach is to use scaffolds seeded with cells in a bioreactor to allow tissue formation. Bioabsorbable scaffolds are used as temporary matrices for cells to attach to in order to proliferate and organize, developing the ECM. As ECM development occurs, the scaffold is fully degraded. Scaffolds are therefore required to have characteristics resembling the native ECM. The structures are grown in vitro in bioreactors, which provide mechanical and metabolic support in an environment resembling in vivo conditions. The tissue is then implanted to allow in vivo growth and remodeling.

12. Explain laminar flow, Couette flow and Hagen-poiseuille equation, turbulent flow

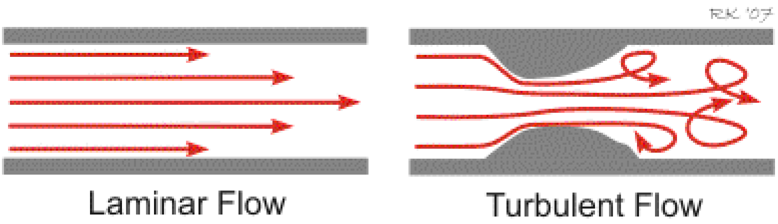
Laminar flow is the normal condition for blood flow throughout most of the circulatory system. It is characterized by concentric layers of blood moving in parallel down the length of a blood vessel. The highest velocity (V_{max}) is found in the center of the vessel. The lowest velocity (V=0) is found along the vessel wall. The flow profile is parabolic once laminar flow is fully developed. This occurs in long, straight blood vessels, under steady flow conditions.



One practical implication of parabolic, laminar flow is that when flow velocity is measured using a Doppler flow meter, the velocity value represents the average velocity of a cross-section of the vessel, not the maximal velocity found in the center of the flow stream.

The orderly movement of adjacent layers of blood flow through a vessel helps to reduce energy losses in the flowing blood by minimizing viscous interactions between the adjacent layers of blood and the wall of the blood vessel. Disruption of laminar flow leads to turbulence and increased energy losses.

Generally in the body, blood flow is laminar. However, under conditions of high flow, particularly in the ascending aorta, laminar flow can be disrupted and become turbulent. When this occurs, blood does not flow linearly and smoothly in adjacent layers, but instead the flow can be described as being chaotic. Turbulent flow also occurs in large arteries at branch points, in diseased and narrowed (stenotic) arteries (see figure below), and across stenotic heart valves.



Turbulence does not begin to occur until the velocity of flow becomes high enough that the flow lamina break apart. Therefore, as blood flow velocity increases in a blood vessel or across a heart valve, there is not a gradual increase in turbulence. Instead, turbulence occurs when a critical Reynolds number (Re) is exceeded. Reynolds number is a way to predict under ideal conditions when turbulence will occur. The equation for Reynolds number is:

$$\mathrm{Re} = \frac{(\bar{v} \cdot D \cdot \rho)}{\eta}$$

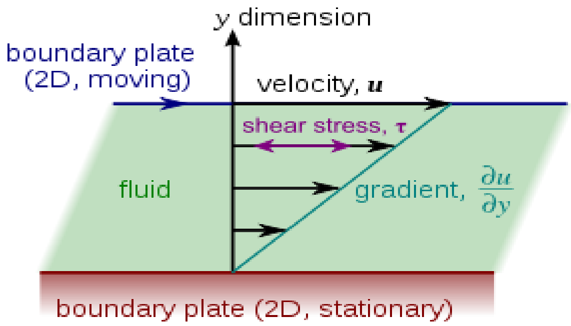
Where v = mean velocity, D = vessel diameter, ρ = blood density, and η = blood viscosity. As can be seen in this equation, Re increases as velocity increases, and decreases as viscosity increases. Therefore, high velocities and low blood viscosity (as occurs with anemia due to reduced hematocrit) are more likely to cause

turbulence. An increase in diameter without a change in velocity also increases Re and the likelihood of turbulence; however, the velocity in vessels ordinarily decreases disproportionately as diameter increases. The reason for this is that flow (F) equals the product of mean velocity (V) times cross-sectional area (A), and area is proportionate to radius squared; therefore, the velocity at constant flow is inversely related to radius (or diameter) squared. For example, if radius (or diameter) is doubled, the velocity decreases to one-fourth its normal value, and Re decreases by one-half.

Under ideal conditions (e.g., long, straight, smooth blood vessels), the critical Re is relatively high. However, in branching vessels, or in vessels with atherosclerotic plaques protruding into the lumen, the critical Re is much lower so that there can be turbulence even at normal physiological flow velocities.

Turbulence generates sound waves (e.g., ejection murmurs, carotid bruits) that can be heard with a stethoscope. Because higher velocities enhance turbulence, murmurs intensify as flow increases. Elevated cardiac outputs, even across anatomically normal aortic valves, can cause physiological murmurs because of turbulence. This sometimes occurs in pregnant women who have elevated cardiac output and who may also have anemia, which decreases blood viscosity. Both factors increase the Reynolds number, which increases the likelihood of turbulence.

Couette flow is the laminar flow of a viscous fluid in the space between two parallel plates, one of which is moving relative to the other. The flow is driven by virtue of viscous drag force acting on the fluid and the applied pressure gradient parallel to the plates.



Couette flow is frequently used in undergraduate physics and engineering courses to illustrate shear-driven fluid motion. The simplest conceptual configuration finds two infinite, parallel plates separated by a distance h. One plate, say the top one, translates with a constant velocity u_0 in its own plane. Neglecting pressure gradients, the Navier–Stokes equations simplify to

$$\frac{d^2u}{dy^2} = 0,$$

where y is a spatial coordinate normal to the plates and u(y) is the velocity distribution. This equation reflects the assumption that the flow is uni-directional. That is, only one of the three velocity components (u, v, w) is non-trivial. If y originates at the lower plate, the boundary conditions are u(0) = 0 and u(h) = u_0. The exact solution

$$u(y) = u_0 \frac{y}{h}$$

can be found by integrating twice and solving for the constants using the boundary conditions.

13. The aorta of a male patient had an inner radius of 13 mm and was 2.2 mm thick in the diastolic state. It was 50 cm long and expanded due to the pumping of the heart. When the heart valve opened in the systolic phase, 70 mL of blood was discharged. Half of this blood was initially stored in the aorta, expanding its wall to some inner radius. Assume the diastolic pressure and systolic pressure to be 80 mmHg and 130 mmHg, respectively, and the heart rate to be 72 beats per minute. Calculate the systolic radius and the wall thickness. What is the stress in the wall in the systolic state? How much energy is stored in the elastic wall? What is the average Reynolds number in the aorta? Is the flow laminar or turbulent? (M/J 2016)

Solution:

i. Volume of the lumen in diastole = π* (13 *10^{-3}) ^2 = 0.5 = 0.265$10^{-3}$ m^3. Increase in volume due to storage of blood = 70/2 = 35 mL = 0.035 m^3 Volume of the lumen in systole= (0.265 +0.035)* 10^{-3}= 0.3 * 10^{-3}m3= π *(r_5)^2* 0.5

ii. Systolic radius = 13.8 mm

Cross-sectional area in diastole = π *[((13 + 2.2) *10^{-3})^2— (13 * 10^{-3})^2] = 194.9* 10^{-3} m^2

Cross-sectional area in systole = 194.9*10^{-3} m^2 = π * [((13.8 +t_s)* 10^{-3})^2— (13.8 *10^{-3}) ^2] t_s= 2 mm

Hoop stress, σ_s = $P_s R_s$/ t_s = 130 mmHg * (133 Pa/mmHg) * 0.0138m/0.002 = 0.113 MPa

iii. Energy stored in the elastic wall = $\mathrm{P_{mean}}$ ΔV

P_{mean} = ($P_{Systolic}$+ 2$P_{Diastolic}$)/3 = (130 + 2 * 80)/3 = 96.67 mmHg
= 96.96 *133 Pa = 12.9 ΔV= 35 mL = 35 *10^{-6} m^3 ,
P_{mean}ΔV= 12.9 * 10^3* 35*10^{-6}= 0.45J

iv. Volumetric flow rate = 70*10^{-6} (m^3 /beat)/(60 sec/72 beats) = 84.1*10^{-6} (m^3/s)

Velocity = volumetric flow rate/average cross-sectional area
= 84.1 * 10^{-6} /(π * ((13 + 13.8)/2 * 10^{-3})^2)= 0.149 m/s
Density of blood (assume), p = 1.1 gm/cc = 1,100 kg/m3
Viscosity (assume), μ = 3 cP = 3*10^{-3} kg/m.s, N_{Re}= ρΔV/ μ = 1,100 * (2* 0.0134) * 0.149/3* 10^{-3}= 1,465. Hence, the flow is **laminar.**

UNIT III

BIOSOLID MECHANICS

- ✓ **Hard tissues**
  - **Bone structure & composition mechanical properties of bone.**
  - **Cortical & cancellous bones**
  - Viscoelastic properties

    Maxwell model & Voight model - Anisotropy
- ✓ **Soft tissues**

  Structure, functions, material properties & modeling of
  - **Cartilage**
  - **Tendon**
  - **Ligament**
  - **Muscle**

LIST OF IMPORTANT QUESTIONS

PART A

1. **What is cortical (or compact) bone?**
2. **What is cancellous (or trabecular) bone?**
3. **Mention composition of bone?**
4. **Explain bone elasticity.**
5. **Explain remodeling or formation of bone.**
6. **What are three types of cells present in the bone?**
7. **What are the mechanical properties of bone?**
8. **What are the metabolic functions of the skeletal s/m? (M/J 2016)**
9. **Identify the various types of soft tissues. (M/J 2016)**
10. **Identify the unique ability of the muscle tissue. Which is defined as the development of tension in the muscle? (N/D 2015)**
11. **What are the functions of skeletal muscle? (N/D 2016)**
12. **Differentiate the mechanical property of tendons and ligaments. (N/D 2016)**
13. **Draw the Maxwell and Voigt model of viscoelasticity(N/D 2016)(A/M 2017)**
14. Draw the structure of tendon & ligament.
15. What is Cartilage? Mention its types & function?
16. What is muscle? Mention its function
17. What are the anisotropic Characteristics of Bone tissue
18. What is Lamellae

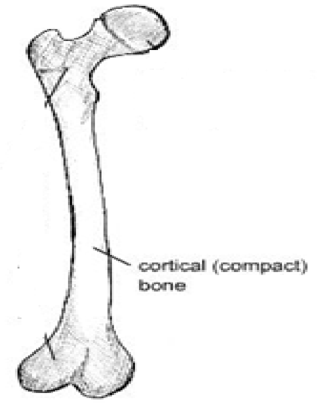
PART B

1. **Explain Structure and composition of bone. (N/D 2015)**
2. **Explain mechanical properties of bone. (N/D 2015)**
3. **Explain the structure, function, material properties and modeling of cartilage,(N/D 2015)**
4. **Explain the structure, function, material properties and modeling of tendon and ligament or Describe the structure and function of ligaments and tendons (M/J 2016) , (N/D 2015)**
5. **Explain the structure, function, material properties and modeling of muscle or identify the three classes of muscle models involved in predicting muscle force. Describe about them(M/J 2016), (N/D 2015)**
6. **Describe the three types of muscle tissue: smooth, skeletal and cardiac. (N/D 2016)**
7. **Describe how the force developed between the actin and myosin filaments of a sarcomere is transferred to the tendon of the muscle. (N/D 2016).**
8. **Explain stresses in bone.**
9. **Discuss about the viscoelastic properties of bone with Maxwell and Voight model.**
10. **Describe the functions of articular cartilage.(N/D 2016)**
11. **Discuss the factors that influence the force developed during muscular activity.(N/D 2016)**
12. **With a neat diagram explain the composition and mechanical property of cortical bone.(A/M 2017)**
13. **How the cartilages are classified? Explain its composition and mechanical property.(A/M 2017)**

PART A

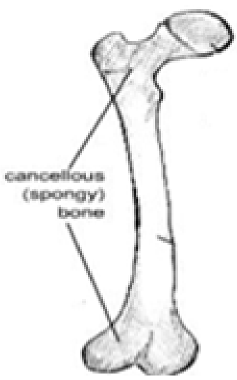
1. What is cortical (or compact) bone?

- Cortical bone forms a dense cylinder down the shaft of the bone surrounding the central marrow cavity.
- While cortical bone accounts for 80% of the mass of bone in the human body, it has a much lower surface area than cancellous bone due to its lower porosity.



2. What is cancellous (or trabecular) bone?

- Cancellous (or trabecular) bone is located at the ends of long bones, accounts for roughly 20% of the total mass of the skeleton, and has an open, honeycomb structure.
- It has a much lower Young's modulus than cortical bone, and this graded modulus gradually matches the properties of the cortical bone to the cartilage that forms the articulating surface on the femoral head.



3. **Mention composition of bone?**

- Bone itself consists mainly of collagen fibers and an inorganic bone mineral in the form of small crystals. *In vivo* bone (living bone in the body) contains between 10% and 20% water. Of its dry mass, approximately 60-70% is bone mineral.
- The composition of the mineral component can be approximated as hydroxyapatite (HA), with the chemical formula $Ca_{10}(PO_4)_6(OH)_2$.
- However, whereas HA as has a Ca:P ratio of 5:3 (1.67), bone mineral itself has Ca:P ratios ranging from 1.37 - 1.87.
- This is because the composition of bone mineral is much more complex and contains additional ions such as silicon, carbonate and zinc.

4. **Explain bone elasticity.**

    - Bone mineral is a ceramic material and exhibits normal Hookean elastic behaviour, i.e. a linear stress-strain relationship. In contrast, collagen is a polymer that exhibits a J-shaped stress-strain curve.

    - Typical stress-strain curves for compact bone, tested in tension or compression in the wet condition, are approximately a straight line. Bone generally has a maximum total elongation of only 0.5 - 3%, and therefore is classified as a brittle rather than a ductile solid.

5. **Explain remodeling or formation of bone.**

- Bone formation is an essential process in the development of the human body. It starts during the development of the fetus, and continues throughout childhood and adolescence as the skeleton grows.

- Bone remodeling meanwhile is a life-long process, consisting of r**esorption** (the breaking down of old bone) and **ossification** (formation of new bone), and is key to shaping the skeleton and to the repair of bone fractures.

6. **What are three types of cells present in the bone?**
    - Osteoblasts
    - osteocytes
    - osteoclasts

    These are respectively responsible for the production, maintenance and resorption of bone.

7. **What are the mechanical properties of bone?**
    - Young's modulus
    - Elasticity

Mechanical properties must be considered in two orthogonal directions:

- Longitudinal, i.e. parallel to osteon alignment. This is the usual direction of loading.
- Transverse, i.e. at right-angles to the long axis of the bone.

8. **What are the metabolic functions of the skeletal s/m? (M/J 2016)**
    - Support
    - Protects
    - Assists in movement
    - Blood cell production

9. **Identify the various types of soft tissues. (M/J 2016)**
    - Tendon
    - Ligament
    - Muscle
    - Fibrous tissues
    - Nerves
    - Blood vessels

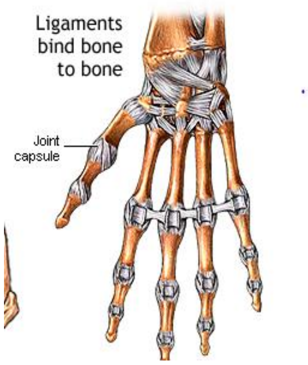
10. Identify the unique ability of the muscle tissue. Which is defined as the development of tension in the muscle? (N/D 2015)

- The mechanical factors determining tension in contraction of the single cross striated muscle fibre are investigated by a continuous registration of stiffness and tension at rest and during contraction.

11. What are the functions of skeletal muscle? (N/D 2016)

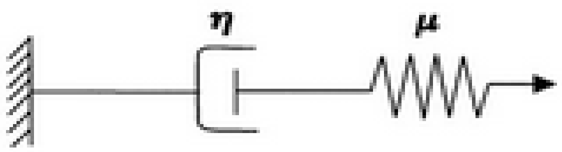
- Through contraction and relaxation, skeletal muscles help to support and move our body.
- Skeletal muscles help with nutrition and temperature regulation.
- It supports movement and homeostasis.
- Skeletal muscle contraction maintains body posture and position.
- A special type of skeletal muscle called a sphincter regulates opening of our digestive and urinary systems.

12. Differentiate the mechanical property of tendons and ligaments. (N/D 2016)

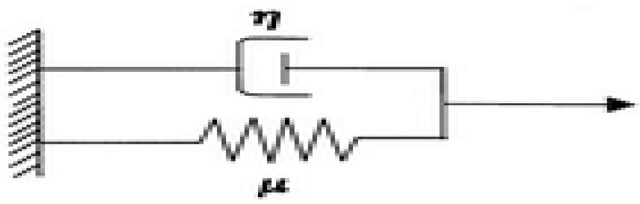
Tendons	Ligaments
• A tendon is a fibrous connective tissue which attaches muscle to bone. • Tendons may also attach muscles to structures such as the eyeball. • A tendon serves to move the bone or structure.	• A ligament is a fibrous connective tissue which attaches bone to bone, and usually serves to hold structures together and keep them stable. 

13. Draw the Maxwell and Voigt model of viscoelasticity(N/D 2016)(A/M 2017)

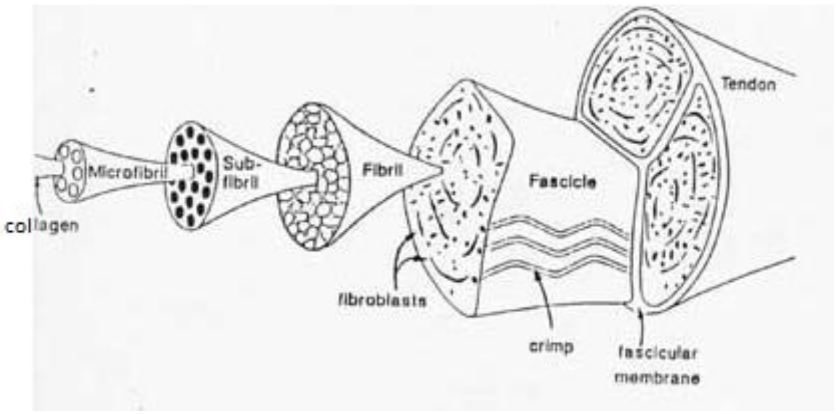


Maxwell model



Voigt model

14. Draw the structure of tendon & ligament.



15. What is Cartilage? Mention its types & function?

Cartilage is a connective tissue consisting of a **dense matrix of collagen fibers and elastic fibers** embedded in a rubbery ground substance. The matrix is produced by cells called chondroblasts, which become embedded in the matrix as chondrocytes.

Types of cartilage:

1) Hyaline cartilage
2) Fibro cartilage
3) Elastic cartilage

Functions:

1) Hyaline cartilage reduces friction at joints, movement, support, and growth.
2) White Fibro cartilage acts as a shock absorber, provides sturdiness without impending movement, and deepens sockets.
3) Elastic cartilage maintains shape and support. Elastic cartilage also protects the ends of bones to prevent them from rubbing together.

16. What is muscle? Mention its function

A band or bundle of fibrous tissue in a human or animal body that has the ability to contract, producing movement in or maintaining the position of parts of the body.

Types of muscle

- Skeletal muscle tissue
- Cardiac muscle tissue
- Smooth muscle tissue

FUNCTION:

- Cardiac muscle tissue -Pumping of blood through the heart**.**

17. What are the anisotropic Characteristics of Bone tissue

Bone is an anisotropic material, indicating that the bone behavior will change depending on the direction of the load application. In general, the bone tissue may lead to higher loads in the longitudinal direction and a lesser quantity of load when applied over the bone surface. The bone is strong to support loads in the longitudinal direction because it is used to receive loads in this direction.

18. What is Lamellae

- The lamellae are concentric rings of a strong matrix formed from mineral salts including calcium and phosphates and collagen fibers.
- The word lamellae literally means "little plates"
- The mineral salts result in the hardness of the bone structure, while the collagen fibers contribute its strength**.**

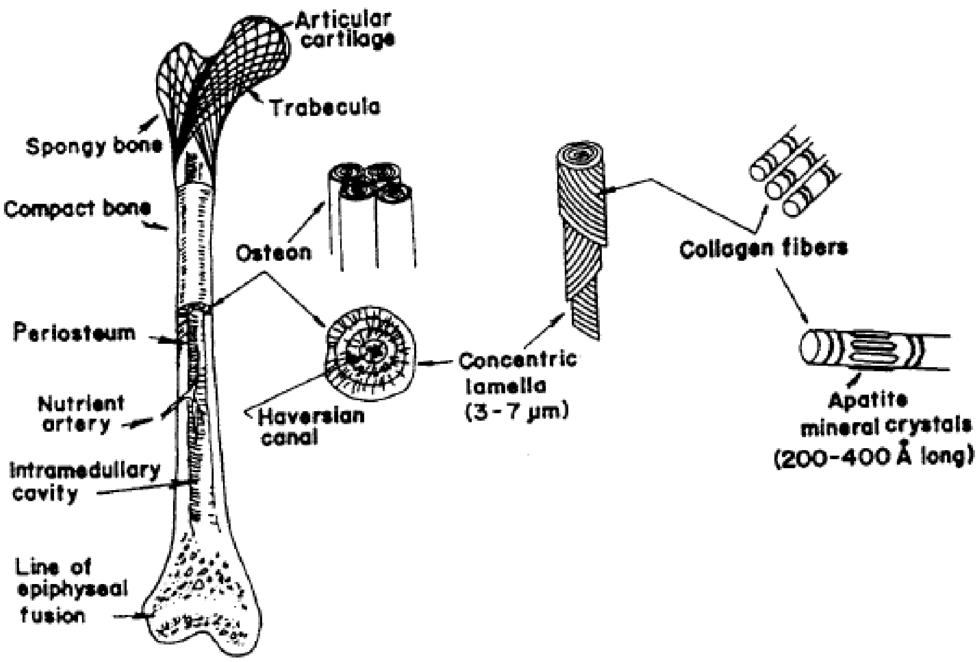
PART B

1. Explain Structure and composition of bone (N/D 2015)

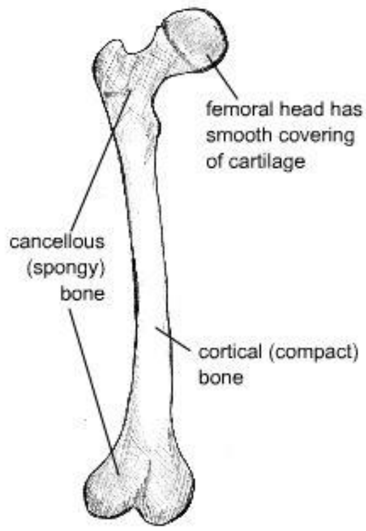
- Hard tissue, mineralized tissue, and calcified tissue are often used as synonyms for bone when describing the structure and properties of bone or tooth.
- Use of the terms mineralized and calcified arises from the fact that the major constituent of bone is a calcium phosphate (thus the term calcified) in the form of a crystalline carbonate apatite (similar to naturally occurring minerals, thus the term mineralized).
- Bone is an anisotropic, heterogeneous, inhomogeneous, nonlinear, thermorheologically complex viscoelastic material.
- In the dry state, bone exhibits piezoelectric properties. Because of the complexity of the structure–property relationships in bone, the most important feature of bone material is its stiffness."

Structure of Bone

- The complexity of bone's properties arises from the complexity in its structure. The structure of a human femur is the following.



- At the smallest unit of structure we have the **tropocollagen** molecule and the associated apatite crystallites.
- Apatite crystallites have been found to be carbonate-substituted hydroxyapatite.The crystallites appear to be about 4 X 20X60 nm in size. This level is denoted the molecular.
- The next level is the ultrastructural ,Here, the collagen and apatite are intimately associated and assembled into a microfibrilar composite(3 to 5μm)
- At the next level, the microstructural, these fibers are either randomly arranged (woven bone) or organized into concentric lamellar groups (osteons) or linear lamellar groups (plexiform bone).
- In addition to the differences in lamellar organization at this level, there are also two different types of architectural structure.
    1. The **dense** type of bone found, for example, in the shafts of long bone is known as **compact or cortical bone**.
    2. A more **porous or spongy** type of bone is found, for example, at the articulating ends of long bones. This is called **cancellous bone**.



- It is important to note that the material and structural organization of collagen- apatite making up osteonic and plexiform bone are the same as the material comprising cancellous bone.

- Finally, we have the whole bone itself constructed of osteons and portions of older, partially destroyed osteons (called interstitial lamellae) in the case of humans or of osteons and/or plexiform bone in the case of mammals. This we denote the Macro structural level.

Composition of Bone

- The composition of bone depends on a large number of factors: the species, which bone, the location from which the sample is taken, and the age, sex, and type of bone tissue, e.g., cancellous, cortical.
- Bone itself consists mainly of collagen fibers and an inorganic bone mineral in the form of small crystals.
- In vivo bone (living bone in the body) contains between 10% and 20% water. Of its dry mass, approximately 60-70% is bone mineral. Most of the rest is collagen, but bone also contains a small amount of other substances such as proteins and inorganic salts.
- Collagen is the main fibrous protein in the body. It has a triple helical structure, and specific points along the collagen fibers serve as nucleation sites for the bone mineral crystals.
- The composition of the mineral component can be approximated as hydroxyapatite (HA), with the chemical formula $Ca_{10}(PO_4)_6(OH)_2$. However, whereas HA as has a Ca:P ratio of 5:3 (1.67), bone mineral itself has Ca:P ratios ranging from 1.37 - 1.87. This is because the composition of bone mineral is much more complex and contains additional ions such as silicon, carbonate and zinc.
- Cartilage is a collagen-based tissue containing very large protein-polysaccharide molecules that form a gel in which the collagen fibers are entangled.
- Articular, or hyaline cartilage forms the bearing surfaces of the movable joints of the body.
- Mechanically, articular cartilage behaves as a linear viscoelastic solid. It also has a very low coefficient of friction (< 0.01), largely attributed to the presence of synovial fluid that can be squeezed out upon compressive loading.

Composition of Adult Human and Bovine Cortical Bone

Species	% H2 O	Apatite	% Dry Weight Collagen	Glycosaminoglycan
Bovine	9.1	76.4	21.5	Not Determined
Human	7.3	67.2	21.2	0.34

2. What are the Mechanical properties of bone? (N/D 2015)

- Although an organic material, bone can often be considered in the same way as man-made engineering materials.
- However, due to the nature of its synthesis it is likely to show more variation in measured properties than typical engineering materials. Factors include:

- Age
- Gender
- Location in the body
- Temperature
- Mineral content
- Amount of water present
- Disease, e.g. osteoporosis

- These variables dependent on each other. For example, the mineral content will vary according to the bone's location in the body, and with the age of the patient.
- As humans age, their bones typically become less dense and the strength of these bones decreases, meaning they are more susceptible to fracture.
- Osteoporosis is a disease involving a marked decrease in bone mass, and it is most often found in post-menopausal women.
- These variables mean that there is a range of measured properties for bone, and so values given in tables will always be an average, with quite a considerable spread possible in the data.

- In addition, the anisotropic structure of bone means that its mechanical properties must be considered in two orthogonal directions:
  - Longitudinal, i.e. parallel to osteon alignment. This is the usual direction of loading
  - Transverse, i.e. at right-angles to the long axis of the bone.

Young's Modulus

- Bone can be considered to consist primarily of collagen fibers and an inorganic matrix, and a simple level it can be analyzed as a fiber composite.
- Composites are materials that are composed of two or more different components.
- The Young's Modulus of aligned fiber composites can be calculated using the Rule of Mixtures and the Inverse Rule of Mixtures for loading parallel and perpendicular to the fibers respectively.

RULE OF MIXTURES

$E_{ax} = fE_f + (1 - f)\, E_m$

INVERSE RULE OF MIXTURES

$$E_{ax} = \left[\frac{f}{E_f} + \frac{(1-f)}{E_m}\right]^{-1}$$

Where

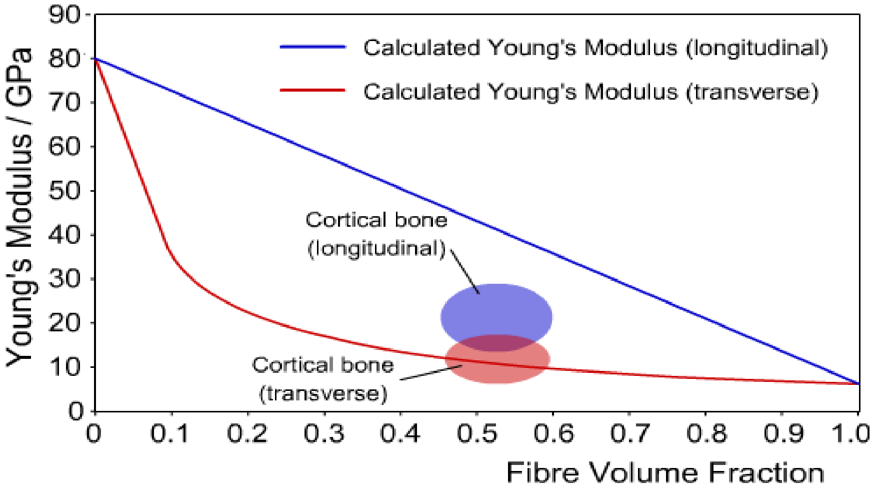
E_f = Young's Modulus of fibres

E_m = Young's modulus of matrix

E_{ax}, E_{trans} = Young's Modulus of composite in axial and transverse directions

f= volume fraction of fibres

- These formulae predict that the composite will be stiffer in the axial direction than the transverse, so cortical bone will be stiffer in the direction parallel to the osteons (i.e. parallel to the long axis of the bone).
- The chart below shows calculated values for the Young's Modulus of bone in both the longitudinal and transverse directions, for a range of fiber volume fractions, as well as the actual values.



Calculated and experimental values of Young's Modulus for cortical bone

- We can see that for the transverse direction, the composite model closely agrees with experimental values.
- However, in the longitudinal direction the difference is large, indicating the model does not give an accurate picture of the behavior of bone.
- A better approximation would be to model bone as a two level composite. One level is provided by hydroxyapatite-reinforced collagen in a single osteon, and the second level is obtained by the approximately hexagonal packing of osteons in a matrix of interstitial bone.
- The actual values for the Young's modulus of bone, compared to collagen and hydroxyapatite, are shown in the table below.

Material	**Young's Modulus, E (GPa)**
Collagen (dry)	6
Bone mineral (Hydroxyapatite)	80
Cortical bone, longitudinal	11-21
Cortical bone, transverse	5-13

- The measured value of Young's Modulus also depends on temperature, decreasing with an increase in temperature, and the strain rate, increasing in value with an increase in strain rate.

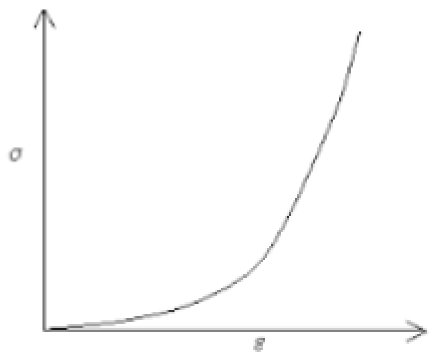
Tensile and Compressive Strength

- Bones such as the femur are subjected to bending moments during normal loading. These create both tensile and compressive stresses in different regions of the bone.
- There is a large variation in measured values of both the tensile and compressive strength of bone.
- Different bones in the body need to support different forces, so there is a large variation in strength between them. Additionally, age is an important factor, with strength often decreasing as a person gets older.

	Longitudinal direction	**Transverse direction**
Tensile strength (MPa)	60-70	~50
Compressive strength (MPa)	70-280	~50

Elasticity

- Bone mineral is a ceramic material and exhibits normal Hookean elastic behavior, i.e. a linear stress-strain relationship. In contrast, collagen is a polymer that exhibits a J-shaped stress-strain curve.

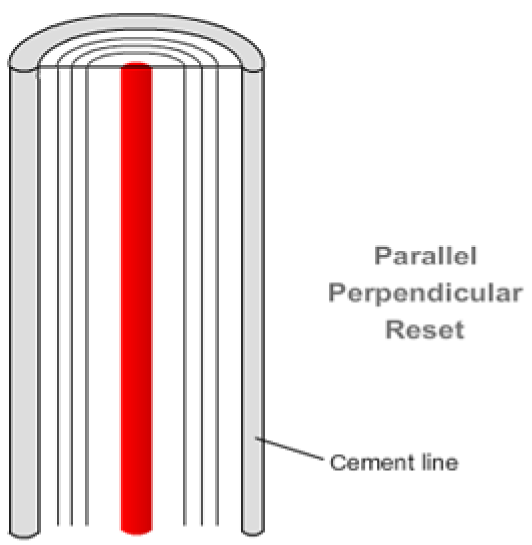


- Typical stress-strain curves for compact bone, tested in tension or compression in the wet condition, are approximately a straight line. Bone generally has a maximum total elongation of only 0.5 - 3%, and therefore is classified as a brittle rather than a ductile solid.

Fracture Toughness

- In contrast to the findings for tensile and compressive strength and modulus, the values of toughness in the transverse direction are generally higher than those in the longitudinal direction.
- This is due to the presence of the cement lines in the microstructure. These are narrow regions around the outermost lamellae in the osteons, and they form the weakest constituent of bone.
- Crack propagation parallel to the osteons can occur much more easily through these regions and this significantly decreases the fracture toughness of cortical bone in the longitudinal direction.
- If a crack is propagating perpendicular to an osteon it will change direction when it reaches a cement line, thus blunting the crack.

Crack propagation in cortical bone



The cement line forms the weakest part of bone.

- As a result, although bone is classified as a brittle material (with the major component being mineral), its toughness is excellent.

- This is much tougher than man-made ceramics due to the presence of the collagen fibers in bone.
- Since the stress-strain curves for loading and unloading are different the elasticity is therefore time-dependent, a common feature of fibrous proteins.

3. Explain the Structure, function, material properties and modeling of cartilage (N/D 2015)

STRUCTURE AND FUNCTIONS OF CARTILAGE:

- Cartilage is usually found in close association with bone in the body.
- It is a type of connective tissue which is tough, semi-transparent, elastic and flexible.
- The matrix or ground substance of cartilage consists mainly of glycoprotein material, chondroitin.
- The cartilage cells (chondrocytes) lie scattered in the matrix.
- Cartilage is covered by a dense fibrous membrane, the perichondrium. No nerves or blood vessels occur in cartilage.
- In some vertebrates, such as sharks, the entire skeleton is made up of cartilage. In mammal embryos, the skeleton first forms as cartilage tissue.
- Cartilage acts as a model and is gradually replaced by bone as the embryo grows. Such cartilage is known as temporary cartilage. The process by which bone tissue follows the cartilage model and slowly replaces it is known as ossification. Permanent cartilage (cartilage which does not become ossified) is found in the tip of the nose, in the external ear and in the walls of the trachea (windpipe) and the larynx (voice-box).
- 3 types
- **Hyaline cartilage**
- **Fibrocartilage**
- **Elastic cartilage**

Hyaline cartilage.

- Hyaline cartilage is semi-transparent and appears bluish-white in colour.

- It is extremely strong, but very flexible and elastic.
- There is an extensive amount of rubbery matrix between the cells and the matrix contains a number of collagenous fibers.
- Hyaline cartilage occurs in trachea, the larynx, the tip of the nose, in the connection between the ribs and the breastbone and also the ends of bone where they form joints.
- Temporary cartilage in mammalian embryos also consists of hyaline cartilage.

Functions

- **Reduces friction at joints.**

  By virtue of the smooth surface of hyaline cartilage, it provides a sliding area which reduces friction, thus facilitating bone movement.

- **Movement**

  Hyaline cartilage joins bones firmly together in such a way that a certain amount of movement is still possible between them.

- **Support**

  The c-shaped cartilagenous rings in the windpipes (trachea and bronchi) assist in keeping those tubes open.

- **Growth**

  Hyaline cartilage is responsible for the longitudinal growth of bone in the neck regions of the long bones.

Fibrocartilage.

- White fibrocartilage is an extremely tough tissue. The orientation of the bundles depends upon the stresses acting on the cartilage.
- The collagenous bundles take up a direction parallel to the cartilage.
- Fibrocartilage is found as discs between the vertebrae between the pubic bones in front of the pelvic girdle and around the edges of the articular cavities such as the glenoid cavity in the shoulder joint.

Functions

- Shock absorbers.

The cartilage between the adjacent vertebrae absorbs the shocks that will otherwise damage and jar the bones while we run or walk.

- Provides sturdiness without impeding movement.

  The white fibrocartilage forms a firm joint between bones but still allows for a reasonable degree of movement.

- Deepens sockets.

  In articular cavities (such as the ball-and-socket joints in the hip and shoulder regions) white fibrocartilage deepens the sockets to make dislocation less possible.

Elastic cartilage.

- Basically elastic cartilage is similar to hyaline cartilage, but in addition to the collagenous fibers.
- The matrix of the elastic also contains an abundant network of branched yellow elastic fibers.
- They run through the matrix in all directions.
- This type of cartilage is found in the lobe of the ear, the epiglottis and in parts of the larynx.

Functions

- **Maintain shape.**

  In the ear, for example, elastic cartilage helps to maintain the shape and flexibility of the organ.

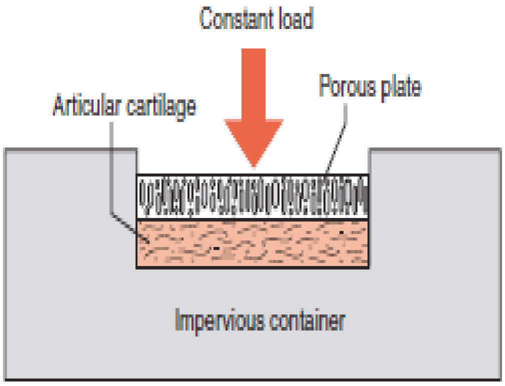
- **Support**

  Elastic cartilage also strengthens and supports these structures.

MATERIAL PROPERTIES AND MODELING OF CARTILAGE

- A confined compression test is one of the commonly used methods for determining material properties of cartilage.
- Confined compression is used in either a “creep” mode or a “relaxation” mode.

- In the creep mode, a constant load is applied to the cartilage through a porous plate, and the displacement of the tissue is measured as a function of time.
- In relaxation mode, a constant displacement is applied to the tissue, and the force needed to maintain the displacement is measured.



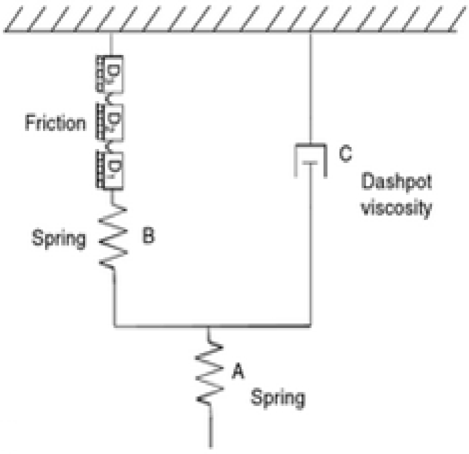
- In creep mode, the cartilage deforms under a constant load, but the deformation is not instantaneous, as it would be in a single-phase elastic material such as a spring.
- The displacement of the cartilage is a function of time. This corresponds to a relatively large flow of fluid out of the cartilage.
- As the rate of displacement slows and the displacement approaches a constant value, the flow of fluid likewise slows.
- At equilibrium, the displacement is constant and fluid flow has stopped. In general, it takes several thousand seconds to reach the equilibrium displacement.

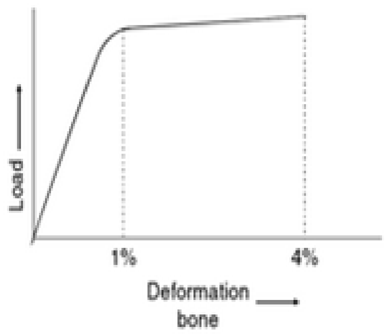
- In the above figure, typical displacement of cartilage tested in a confined compression test.
- A constant load is applied to the cartilage, and the displacement is measured over time.
- Initially, the deformation is rapid, as relatively large amounts of fluid are exuded from the cartilage.
- As the displacement reaches a constant value, the flow slows to zero.
- By fitting the mathematical biphasic model to the measured displacement, two material properties of the cartilage are determined:
    - The aggregate modulus
    - permeability
- The aggregate modulus is a measure of the stiffness of the tissue at equilibrium when all fluid flow has ceased.
- The higher the aggregate modulus, the less the tissue deforms under a given load.
- The aggregate modulus of cartilage is typically in the range of 0.5 to 0.9 MPa.
- The permeability indicates the resistance to fluid flow through the cartilage matrix.
- Permeability was first introduced in the study of flow through soils.
- The average fluid velocity through a soil sample (v_{ave}) is proportional to the pressure gradient (Vp).
- The constant of proportionality (k) is called the permeability. This relationship is expressed by Darcy's law,

$$v_{ave} = kVp$$

MODELING

Cartilage is a tough and pliant material with its external surface very smooth and without any geometrical irregularities. Articular cartilage line the terminal ends of hard and mineralized bones at their articulating ends with other bones. The geometrical congruence of two opposing cartilage plates in any joint is extremely well adapted to the functional movements. Mechanically, cartilages are adapted to resist and endure compression and bending forces. Structurally, they are made of two major components. First the collagen fibers are arranged in wide arches with their domes directed outward. The intervening space is densely packed with a protein- carbohydrate complex (proteoglycan), which has enormous affinity for water and, hence , maintains a high turgor pressure. Owing to structural arrangement, any compressive or bending forces, one cartilage slides on the other, preventing any bone- on-bone grinding.





4. Explain the structure, function, material properties and modeling of tendon and ligament or Describe the structure and function of ligaments and tendons (M/J 2016) (N/D 2015)

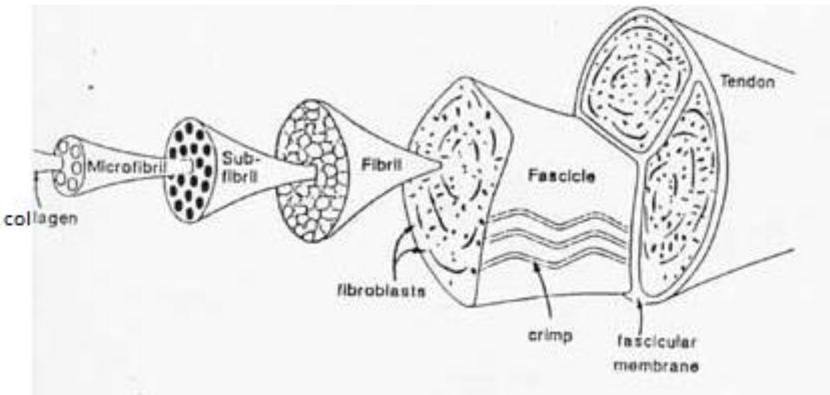
STRUCTURE AND FUNCTIONS OF TENDON AND LIGAMENT

Introduction:

- Ligaments and tendons are soft collagenous tissues.
- Ligaments connect bone to bone and tendons connect muscles to bone.

- Ligaments and tendons play a significant role in musculoskeletal biomechanics.
- They represent an important area of orthopedic treatment for which many challenges for repair remain.
- A good deal of these challenges has to do with restoring the normal mechanical function of these tissues.
- Again, as with all biological tissues, ligaments and tendons have a hierarchial structure that affects their mechanical behavior.
- In addition, ligaments and tendons can adapt to changes in their mechanical environment due to injury, disease or exercise.
- Thus, ligaments and tendons are another example of the structure-function concept and the mechanically mediated adaptation concept that permeate this biomechanics course.

Hierarchical Ligament and Tendon Structure



- The largest structure in the above schematic is the tendon (shown) or the ligament itself.
- The ligament or tendon then is split into smaller entities called fascicles.
- The fascicle contains the basic fibril of the ligament or tendon, and the fibroblasts, which are the biological cells that produce the ligament or tendon.
- There is a structural characteristic at this level that plays a significant role in the mechanics of ligaments and tendons: the crimp of the fibril.
- The crimp is the waviness of the fibril; we will see that this contributes significantly to the nonlinear stress strain relationship for ligaments and tendons and indeed for basically all soft collagenous tissues.

TENDONS

Anatomy:

1. Tendons contain collagen fibrils (Type I)
2. Tendons contain a proteoglycans matrix
3. Tendons contain fibroblasts (biological cells) that are arranged in parallel row.

Basic Functions

- Tendons carry tensile forces from muscle to bone.
- They carry compressive forces when wrapped around bone like a pulley

Type I Collagen:

1. ~86% of tendon dry weight
2. Glycine (~33%)
3. Proline (~15%)
4. Hydroxyproline (~15%, almost unique to collagen, often used to identify)

LIGAMENTS

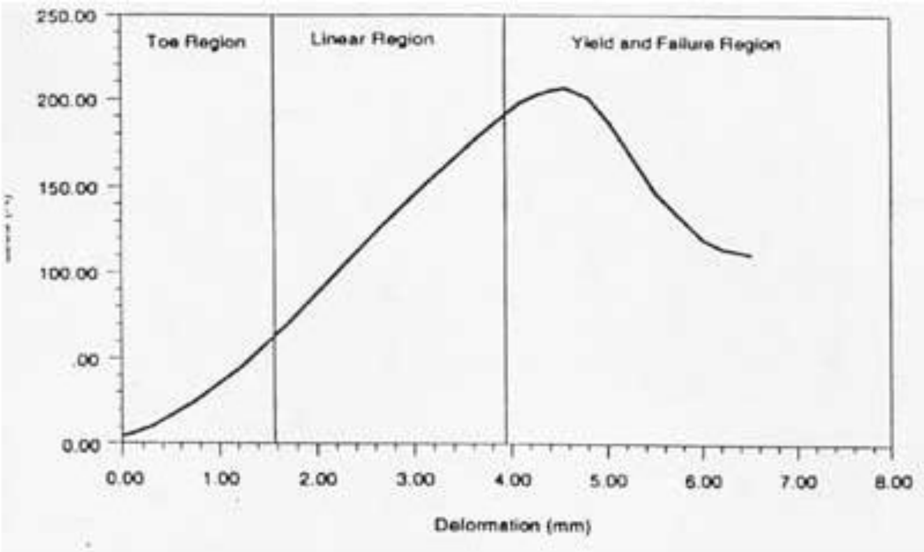
Anatomy

1. Similar to tendon in hierarchical structure
2. Collagen fibrils are slightly less in volume fraction and organization than tendon
3. Higher percentage of proteoglycan matrix than tendon
4. Fibroblasts

MATERIAL PROPERTIES AND MODELING OF TENDON AND LIGAMENT

- **Nonlinear Elasticity**

If one neglects viscoelastic behavior, a typical stress strain curve for ligaments and tendons can be drawn as:

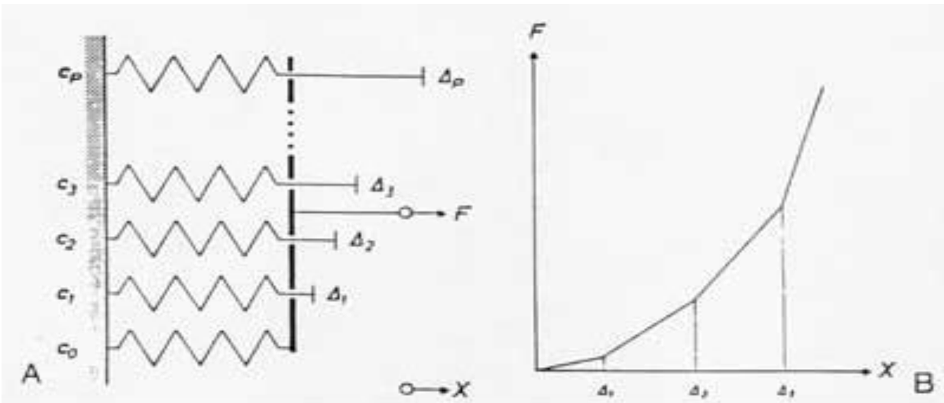


There are three major regions of the stress strain curve:

- The toe or toe-in region,
- The linear region and
- The yield and failure region.

- In physiologic activity, most ligaments and tendons exist in the toe and somewhat in the linear region.
- These constitute a nonlinear stress strain curve, since the slope of the toe-in region is different from that of the linear region.

A simple model illustrating the dependence of ligament/tendon nonlinear stress/strain relationships is shown below:

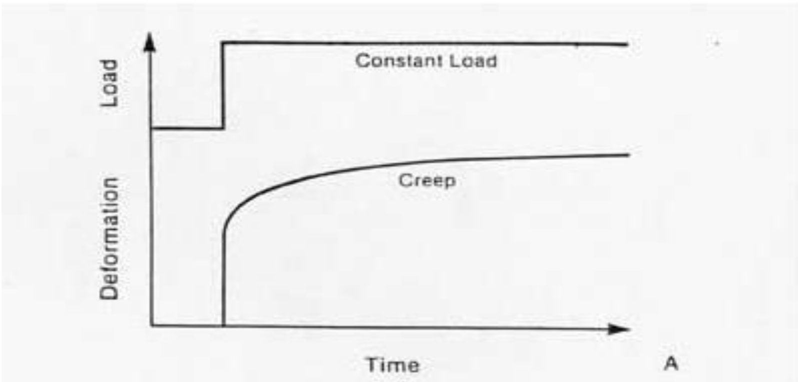


- In this case, as a spring is stretched to its limit its stiffness increases.

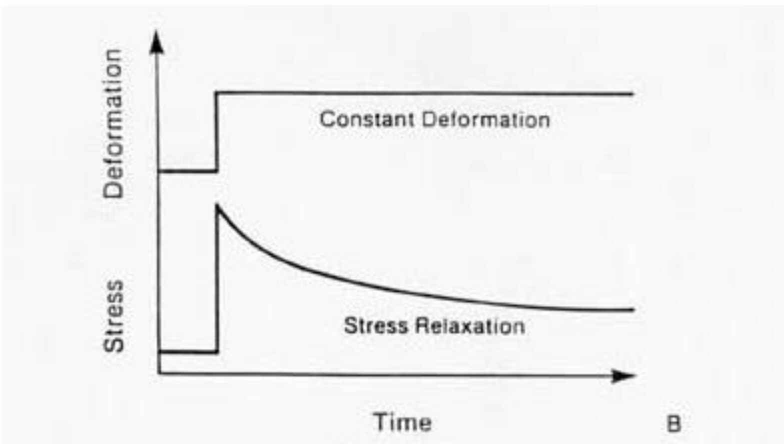
- This can easily be seen if the effective ligament stiffness is modeled using the Voight model, with each fibril contributing a small part to the overall stiffness.

- **Viscoelasticity**

- Another important aspect of ligament/tendon behavior is viscoelasticity.
- The relationship between stress and strain is not constant but depends on the time of displacement or load.
- There are two major types of behavior characteristic of viscoelasticity.
- The first is creep. Creep is increasing deformation under constant load.
- Creep is illustrated schematically below:



- The second significant behavior is stress relaxation.
- This means that the stress will be reduced or will relax under a constant deformation. This behavior is illustrated below:



Modeling Ligaments and Tendons

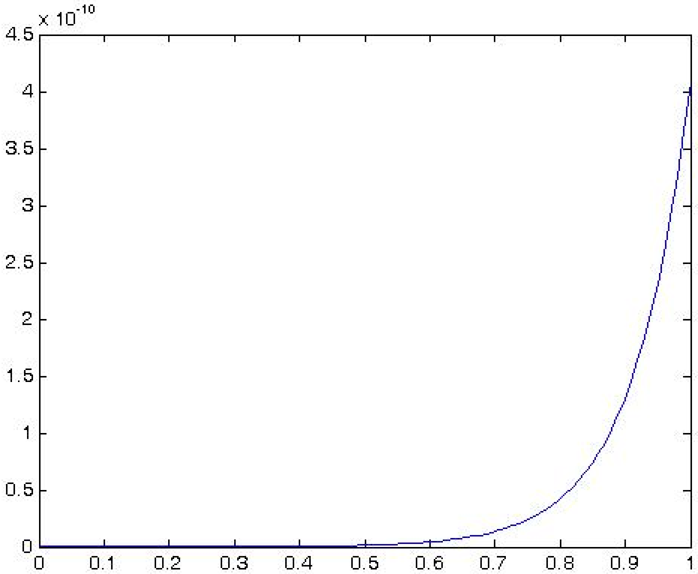
- **Nonlinear Elastic Materials**
- To represent the nonlinear elastic behavior of ligaments a hyper elastic model including strain energy function. The strain energy function is given below:

$$W = F_1(I_1, I_2) + F_2\left(\sqrt{I_4}\right) + F_3\left(I_1, I_2, \sqrt{I_4}\right)$$

- Another model described ligament behavior as nonlinear elastic using a transversely isotropic strain energy function. This was written as:

$$W = F_1(I_1, I_2) + F_2(\lambda) + F_3(I_1, I_2, \lambda)$$

Where F1 represents the contribution of the isotropic ground matrix, F2 represents the contribution of the collagen fibers, and F3 represents the interaction between the collagen fibers and the matrix.



Stress- strain curve explaining nonlinear behavior

- **Viscoelastic Materials**
  - As ligaments and tendons exhibit time dependent behavior, a number of studies have modeled ligaments and tendons using viscoelasticity.
  - The 2nd PK (2nd Piola-Kirchoff stress) stress tensor as:

$$S_{ij}(t) = \int_{-\infty}^{t} G_{ijpm}(t-\tau) \frac{\partial S_{pm}}{\partial E_{kl}} \frac{\partial E_{kl}}{\partial \tau} d\tau$$

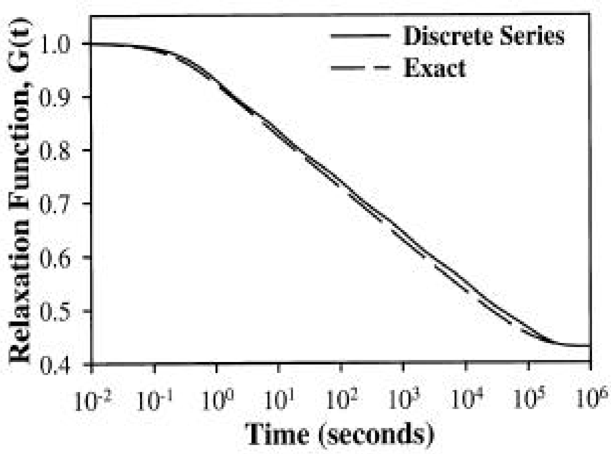
- it is generally assumed that the relaxation function is the same in all directions, which means that the stress relaxation tensor is actually a scalar and the above is written as:

$$S_{ij}(t) = \int_{-\infty}^{t} G(t-\tau) \frac{\partial S_{ij}}{\partial E_{kl}} \frac{\partial E_{kl}}{\partial \tau} d\tau$$

- For purposes of finite element analysis approximated the reduced relaxation function as:

$$G(t) = G_e + \frac{G_0 - G_e}{N_d + 1} \sum_{I=0}^{N_d} e^{\left(\frac{t}{10^{(I+I_0)}}\right)}$$

Analytical forms of the stress relaxation function very well as shown below:



5. Explain the structure, function, material properties and modeling of muscle or identify the three classes of muscle models involved in predicting muscle force. Describe about them. (M/J 2016), (N/D 2015)

STRUCTURE AND FUNCTIONS OF MUSCLE:

There are 3 types of muscle tissue:

- Skeletal muscle tissue
- Cardiac muscle tissue and
- Smooth muscle tissue.

- The structure of these muscle tissues can be described from the level of detail of the muscle fibers (muscle cells).
- The functions of muscle tissues depend on the type of muscle tissues and their locations in the body.

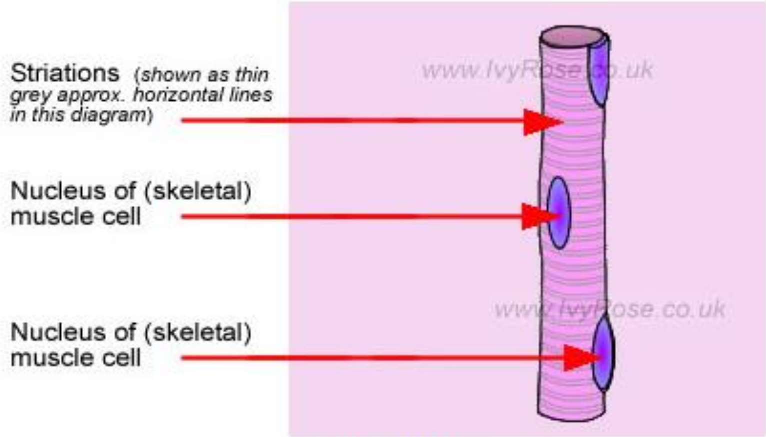
(a) Skeletal Muscle Tissue

Structure:

- Skeletal muscle is called "striated" because of its appearance consisting of light and dark bands visible using a light microscope.
- As shown in the diagram a single skeletal muscle cell is long and approximately cylindrical in shape, with many nuclei located at the edges of the cell.

Function:

- Movement of the skeleton under conscious control, including movement of limbs, fingers, toes, neck, etc.
- Movement of tissues of facial expression under conscious control, e.g. ability to smile and to frown.



Above: Diagram of Skeletal Muscle Tissue

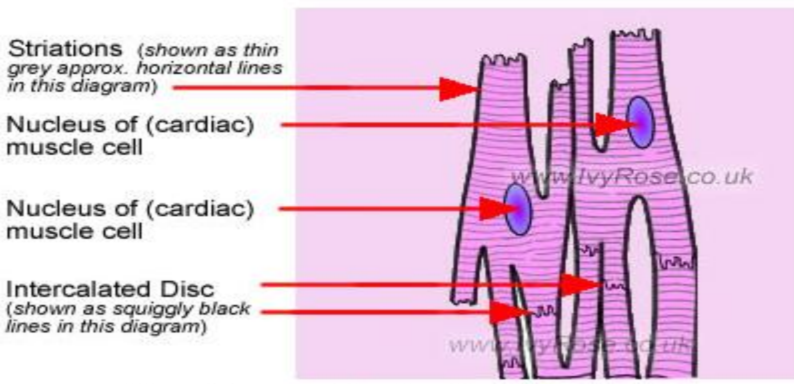
(b) Cardiac Muscle Tissue

Structure:

- Cardiac muscle fibers are striated, branched (sometimes described as Y-shaped), and have a single central nucleus.
- These fibers are attached at their ends to adjoining fibers by thick plasma membranes called intercalated discs.

Function:

- **Pumping of blood through the heart**:
  - De-oxygenated blood through the Right Atrium and Right Ventricle to the lungs, and
  - Oxygenated blood through the Left Atrium and Left Ventricle to the aorta, then the rest of the body.



Above: Diagram of Cardiac Muscle Tissue

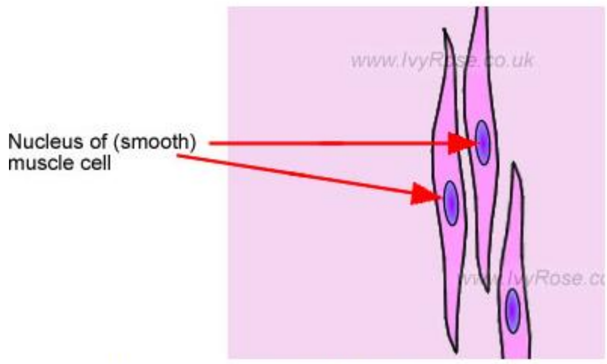
(c) Smooth Muscle Tissue

Structure:

- Unlike Skeletal and Cardiac muscle tissue, Smooth muscle is not striated.
- Smooth muscle fibers are small and tapered with the ends reducing in size, in contrast to the cylindrical shape of skeletal muscle. Each smooth muscle fiber has a single centrally located nucleus.

Function:

- Contractions of smooth muscle constrict (narrow = reduce the diameter of) the vessels they surround. This is particularly important in the digestive system in which the action of smooth muscle helps to move food along the gastrointestinal tract as well as breaking the food down further.
- Smooth muscle also contributes to moving fluids through the body and to the elimination of indigestible matter from the gastrointestinal system.

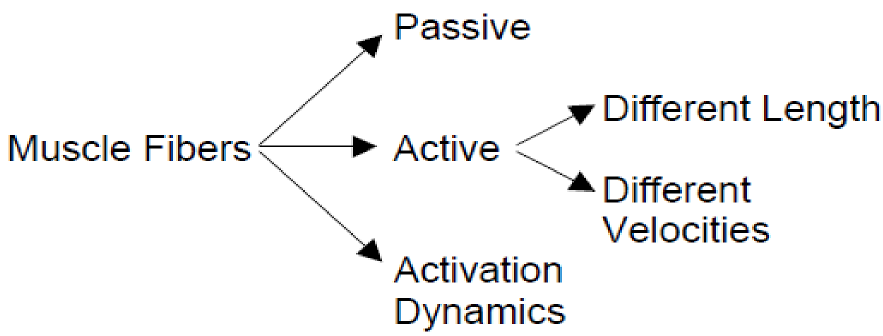


Above: Diagram of Smooth Muscle Tissue

MATERIAL PROPERTIES AND MODELLING OF MUSCLE:

- The anatomical components of the human body can be divided into active and passive structures. Active structures produce force whereas passive structures do not.
- The only active structure in the human body is muscle and its capability of active contraction makes it distinct from other soft biological tissues.

- Because of its contractile properties, muscle is less easily associated with strictly mechanical properties than are bones or ligaments, for which force-elongation or stress-strain relations can be well-defined. So, dealing with the mechanics of muscle.
- It is believed that the development of realistic constitutive models for the mechanical behavior of the muscle tissue can enhance the understanding of the complex muscle function.



- Additionally, it is the basis for realistic simulation of the human musculoskeletal system. It is well-proved that the presence of active muscle contraction to numerical models of biological structures provides a better insight to the whole structure loadings during movement and more accurate predictions of their behavior.
- There are three ways one could use to model the mechanical behavior of a muscle:

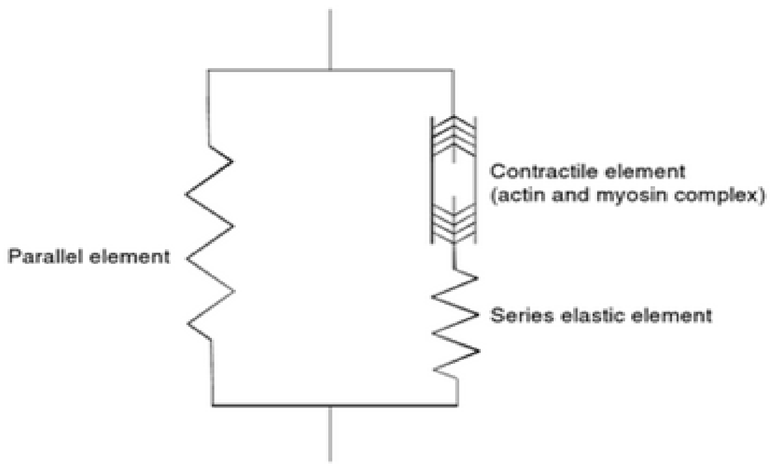
MODEL REPRESENTATION – Hill's model:

- **Potential Model Components**
  Models of muscle normally include some of the following components.
- **Contractile Component or contractile element (CE)**– normally representing (some) properties of the muscles (force-length, force-velocity, activation dynamics).
- **Parallel Elastic Component (PEC)** – normally a linear elastic component representing elastic material in parallel to the muscle fibers (connective tissue).

- **Series Elastic Component (SEC)** – normally a linear elastic component representing elastic properties of material in series with the contractile component (tendon, muscle cross-bridge elasticity).

The challenge to describe mechanical characteristics of this contractile tissue is enhanced by the range of elastic properties exhibited by each muscle. The differences are not only in gross morphology, but also in ultimate structural profiles. Regardless of differences, Hill's model describes each muscle with a contractile element, a series elastic element, and a parallel element. In the active state, the actin filaments slide on the myosin filament during contraction causing muscle length shortening. Skeletal muscle can shorten up to 50% of their resting length.

This shortening pulls the tendon, which in turn moves the bone at the joint, causing the segment motion. The muscles have a viscous element as well. It shows transient tension decay with a rate constant closely related to intrinsic maximum speed with which the muscle can shorten when tetanised (maximal contraction).
The force of contraction of muscle is dependent on their cross sectional areas and the rate of the neural input and is maximum when muscle is stimulated at 100Hz. While shortening of the muscles takes place due to contractile element behavior, the return to its resting position after the contraction is achieved by the release of stored energy in the series elastic components.



Hill's three element muscle model

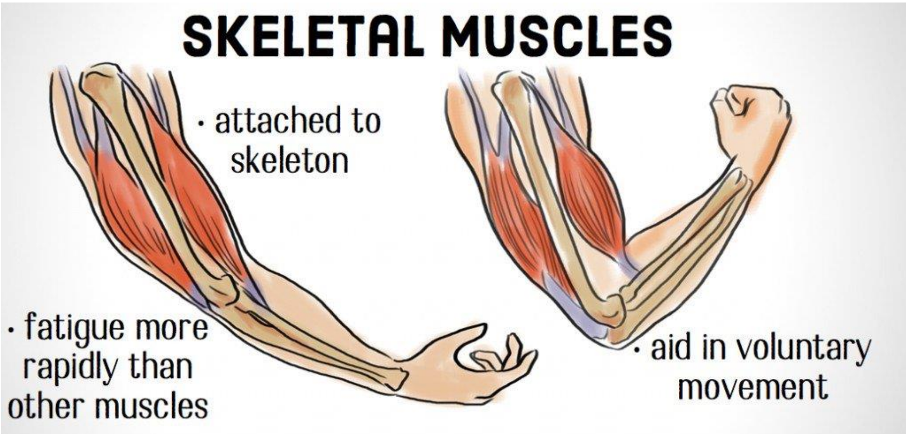
6. Describe the three types of muscle tissue: smooth, skeletal and cardiac. (N/D 2016)

There are three types of muscle found in the human body:

- **Skeletal Muscle**
- **Smooth Muscle**
- **Cardiac Muscle** (heart muscle)

Skeletal muscle

- Skeletal Muscles are those which attach to bones and have the main function of contracting to facilitate movement of our skeletons.
- They are also sometimes known as striated muscles due to their appearance. The cause of this 'stripy' appearance is the bands of Actin and Myosin which form the Sarcomere, found within the Myofibrils.
- Skeletal muscles are also sometimes called voluntary muscles, because we have direct control over them through nervous impulses from our brains sending messages to the muscle.
- Contractions can vary to produce powerful, fast movements or small precision actions.
- Skeletal muscles also have the ability to stretch or contract and still return to their original shape.

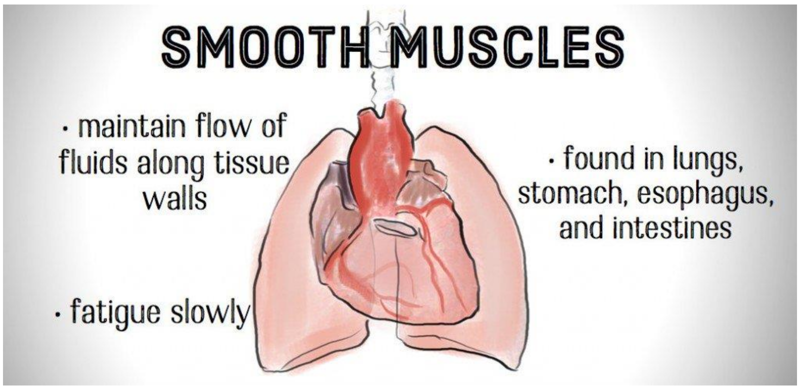


Skeletal muscle fibre type

- Not all fibres within Skeletal muscles are the same.
- Different fibre types contract at different speeds, are suited to different types of activity and vary in colour depending on their Myoglobin (an oxygen carrying protein) content.

Smooth muscle

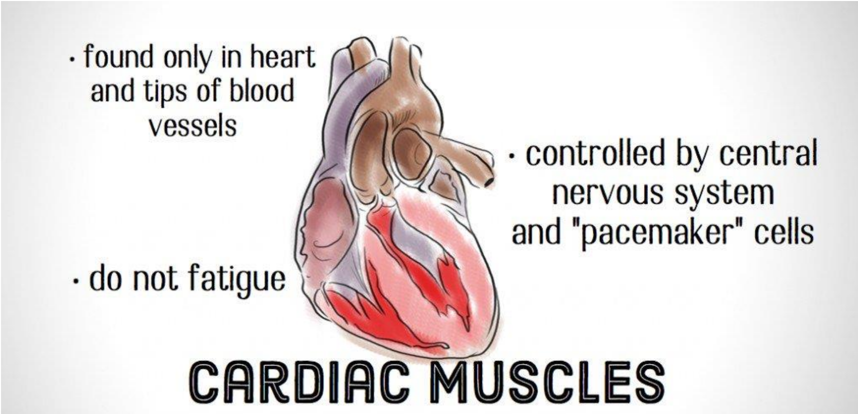
- Smooth muscle is also sometimes known as Involuntary muscle due to our inability to control its movements, or Unstriated as it does not have the stripy appearance of Skeletal muscle.
- Smooth muscle is found in the walls of hollow organs such as the Stomach, Oesophagus, Bronchi and in the walls of blood vessels.
- This muscle type is stimulated by involuntary neurogenic impulses and has slow, rhythmical contractions used in controlling internal organs, for example, moving food along the Oesophagus or contricting blood vessels during Vasoconstriction.



Cardiac muscle (heart muscle)

- This type of muscle is found solely in the walls of the heart. It has similarities with skeletal muscles in that it is striated and with smooth muscles in that its contractions are not under conscious control.
- However this type of muscle is highly specialised. It is under the control of the autonomic nervous system, however, even without a nervous imput contractions can occur due to cells called pacemaker cells.

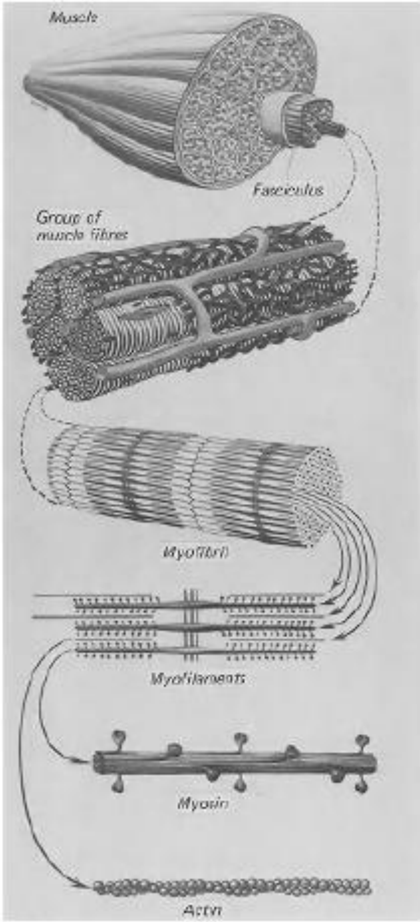
- Cardiac muscle is highly resistant to fatigue due to the presence of a large number of mitochondria, myoglobin and a good blood supply allowing continuous aerobic metabolism.



7. Describe how the force developed between the actin and myosin filaments of a sarcomere is transferred to the tendon of the muscle.(N/D 2016)

- A skeletal muscle fiber is elongated, having a diameter of 10-60 *firn,* and a length usually of several millimeters to several centimeters; but sometimes the length can reach 30 cm in long musc1es.
- The fibers may stretch from one end of musc1e to another, but often extend only part of the length of the musc1e, ending in tendinous or other connective tissue intersections.
- The flattened nuclei of muscle fibers lie immediately beneath the cell membrane.
- The cytoplasm is divided into longitudinal threads or *myofibrils,*each about 1 *µm* in diameter. These myofibrils are striated when they are stained by dyes and when they are examined optically.
- Some zones stain lightly with basic dyes such as hematoxylin, rotate the plane of polarization of light weakly, and are called *isotropie* or *I bands.*
- Others, alternating with the former, stain deeply with hematoxylin and strongly rotate the plane of polarization of light to indicate a highly ordered substructure.
- They are called *anisotropie* or *A bands.* The I bands are bisected transversely by a thin line also stainable with basic dyes: this line is called the

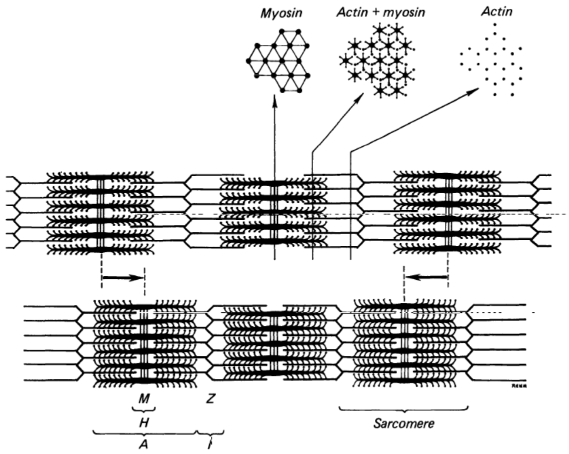
zwisehenscheibe or Z *band.* The A bands are also bisected by a paler line called the *H* band.



hierarchy of skeletal muscle

- If the muscle contracts greatly, the I and H bands may narrow to extinction, but the A bands remain unaltered.
- Each myofibril is composed of arrays of *myofilaments.*
- These are divided transversely by the Z bands into serially repeating regions termed *sarcomeres,* each about 2.5 *µm* long, with the exact length dependent on the force acting in the muscle and the state of excitation. Two types of myofilament are distinguishable in each sarcomere, fine ones about 5 nm in diameter and thicker ones about 12 nm across. The fine ones are *actin* molecules.
- The thick ones are *myosin* molecules. The actin filaments are each attached at one end to a Z band and are free at the other to interdigitate with the myosin filaments.

- It is seen that the A band is the band of myosin filaments, and the I band is the band of the parts of the actin filaments that do not overlap with the myosin. The H bands are the middle region of the A band into which the actin filaments have not penetrated. Another line, the M *band,* lies transversely across the middle of the H bands, and else examination shows this to consist of fine strands interconnecting adjacent myosin filaments.



The structure of a myofilament, showing the spatial arrangement of the actin and myosin molecules.

8. Explain stresses in bone.

Stresses

Bones such as the femur are subjected to a bending moment, and the stresses (both tensile and compressive) generated by this bending moment account for the structure and distribution of cancellous and cortical bone.

In the upper section of the femur, the cancellous bone is composed of two distinct systems of trabeculae. One system follows curved paths from the inner side of the shaft and radiates outwards to the opposite side of the bones, following the lines of maximum compressive stress. The second system forms curved paths from the outer side of the shaft and intersect the first system at right angles. These trabeculae follow the lines of maximum tensile stress, and in general are lighter in structure than those of the compressive system.

The thickness of the trabeculae varies with the magnitude of the stresses at any point, and by following the paths of the principal compressive and tensile stresses they carry these stresses economically. The greatest strength is therefore achieved with the minimum of material.

The distribution of the compact bone in the shaft is also due to the requirement to resist the bending moment stresses. To resist these stresses, the material should be as far from the neutral axis as possible. A hollow cylinder is the most efficient structure, again achieving the greatest strength with the minimum of material.

a. Formation and remodeling of bone

Bone formation is an essential process in the development of the human body. It starts during the development of the foetus, and continues throughout childhood and adolescence as the skeleton grows. Bone remodelling meanwhile is a life-long process, consisting of *resorption* (the breaking down of old bone) and ossification (formation of new bone), and is key to shaping the skeleton and to the repair of bone fractures.

There are three types of cell present in bone that are of particular interest – osteoblasts, osteocytes and osteoclasts, which are respectively responsible for the production, maintenance and resorption of bone.

- **Osteoblasts**
  Mononucleated "bone-forming" cells found near the surface of bones. They are responsible for making *osteoid*, which consists mainly of collagen. The osteoblasts then secrete alkaline phosphatase to create sites for calcium and phosphate deposition, which allows crystals of bone mineral to grow at these sites. The osteoid becomes mineralised, thus forming bone.
- **Osteocytes**
  These are osteoblasts that are no longer on the surface of the bone, but are instead found in lacunae between the lamellae in bone. Their main role is homeostasis – maintaining the correct oxygen and mineral levels in the bone.
- **Osteoclasts**
  Multinucleated cells responsible for bone resorption. They travel to

specific sites on the surface of bone and secrete acid phosphatase, which unfixes the calcium in mineralised bone to break it down.

During foetal development there are two mechanisms for creating bone tissue:

- Endochondral ossification
- Intramembranous ossification

Intramembranous ossification occurs in the formation of flat bones such as those in the skull.

Endochondral ossification

This involves bone growth from an underlying cartilage model, and is seen in the formation and growth of long bones such as the femur.

The initial step involves the development of a cartilage model, which has the rough shape of the bone being formed. In the middle of the shaft is the primary ossification centre, where osteoblasts lay down osteoid on the shaft to form a bone collar.

The osteoid calcifies, and blood vessels grow into cavities within the matrix. Osteoblasts then use the calcified matrix as a support structure to lay down more osteoid and form trabeculae within the bone. Meanwhile osteoclasts break down spongy bone to create the medullary cavity, which contains bone marrow.

Initially the bone material is deposited with the collagen fibres in random directions, meaning the strength is much lower than at the final stage in which the fibres are aligned. The primary structure is called woven bone because the collagen fibres are woven together randomly. This is then converted into lamellar bone over time, which is much stronger due to the aligned fibers. The osteoid deposited by the osteoblasts calcifies to initially produce primitive cancellous bone. At sites where cortical bone is required, further deposition of osteoid occurs to increase the density of the structure.

At birth secondary ossification centres appear at either end of long bones. Between the primary and secondary centres is the epiphyseal plate, made of cartilage, which continues to form new cartilage and be replaced by bone such that the bone increases in length. This continues until a person is in their mid-twenties, when the plate is finally replaced by bone and no further growth occurs.

Remodelling of bone

Ossification is also essential in the remodelling of bone. This occurs throughout a person's lifetime, with ossification and resorption (removal of bone tissue) working together to reshape the skeleton during growth, maintain calcium levels in the body, and repair micro-fractures caused by everyday stress.

The remodelling of cortical bone follows the same process as shown above, but with a different geometry in order to form the concentric lamellae seen in osteons.

Responsive material

Bone is considered to be a responsive material. The formation and resorption of bone occur continuously: the body responds to stress levels in different areas of bone to ensure the right amount of healthy bone tissue is maintained and the bone can be continually reshaped.

A stress of 25–40 MPa is sufficient to maintain the correct levels of bone. If the bone is under-stressed for prolonged periods of time, bone wastage will set in, and the bones will become thinner. This can be an issue if a patient is bed-ridden for a long time, and is also observed in astronauts after long periods in space. A similar effect occurs during osteoporosis. This results in an imbalance of resorption and formation, causing bones to become thinner and weaker.

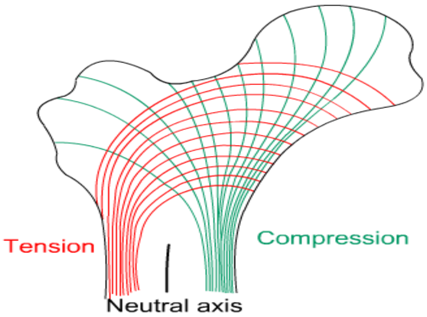
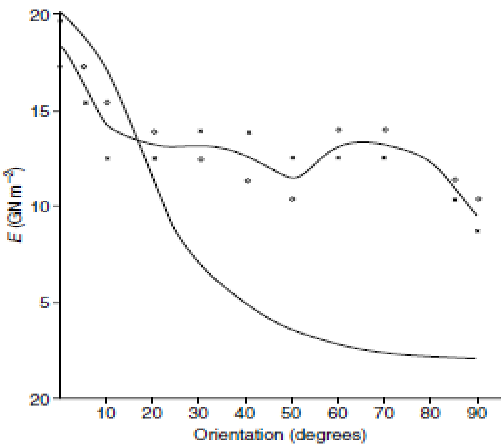


Diagram showing computed lines of constant stress from the analysis of various transverse sections

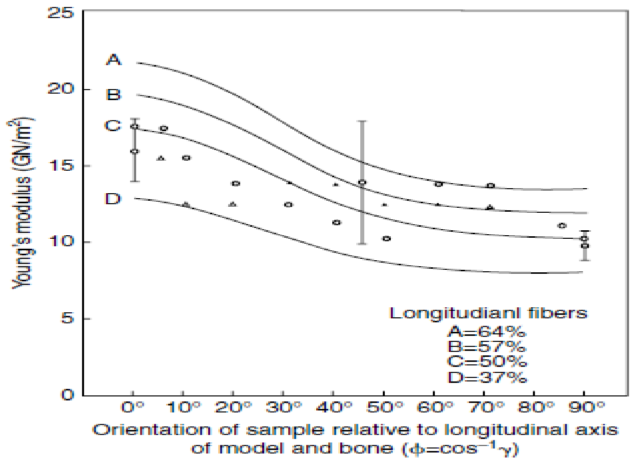
The opposite effect can be seen when bones are suddenly subjected to higher levels of stress than normal. Studies have been conducted that show an increase in bone mass in new recruits to the army as they begin intensive training.

9. Discuss about the viscoelastic properties of bone with Maxwell and voight model.

Bone (along with all other biologic tissues) is a viscoelastic material. Clearly, for such materials, Hooke's law for linear elastic materials must be replaced by a constitutive equation that includes the time dependency of the material properties. The behavior of an anisotropic linear viscoelastic material may be described by using the Boltzmann superposition integral as a constitutive equation:



Variation in young's modulus of bovine femur specimens (E) with the orientation of specimen axis to the long axis of the bone, for wet and dry conditions compared with the theoretical curve predicted from a fiber reinforced composite model.



Comparison of predictions of Karlz two-level composite model with the experimental data. Each curve represents a different lamellar configuration within a single osteon, with longitudinal fiber a=60% , B= 57%, C=50% and D= 37%. And the rest of the fiber assumed horizontal.

$$\sigma_{ij}(t) = \int_{-\infty}^{t} C_{ijkl}(t-\tau)\frac{d\epsilon_{kl}(\tau)}{d\tau}d\tau$$

where $\sigma_{ij}(t)$ and $\epsilon_{kl}(\tau)$ are the time-dependent second-rank stress and strain tensors, respectively, and $C_{ijkl}(t-\tau)$ is the fourth-rank relaxation modulus tensor.

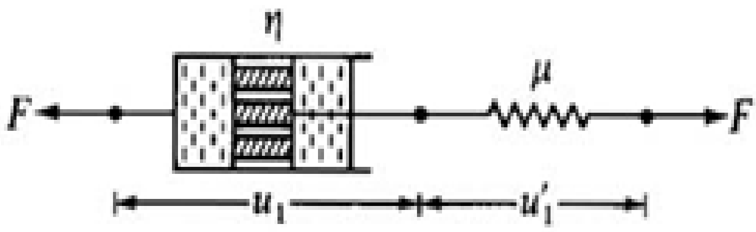
Thus, for a more complete understanding of bone's response to applied loads, it is important to know its rheological properties. There have been a number of early studies of the viscoelastic properties of various long bones. However, none of these was performed over a wide enough range of frequency (or time) to completely define the viscoelastic properties measured, for example, creep or stress relaxation. Thus it is not possible to mathematically transform one property into any other to compare results of three different experiments on different bones.

In the first experiments over an extended frequency range, the biaxial viscoelastic as well as uniaxial viscoelastic properties of wet cortical human and bovine femoral bone were measured using both dynamic and stress relaxation techniques over eight decades of frequency (time). The results of these experiments showed that bone was both nonlinear and thermorheologically complex, that is, time–temperature superposition could not be used to extend the range of viscoelastic measurements. A nonlinear constitutive equation was developed based on these measurements.

Viscoelasticity considers in addition a dissipative phenomenon due to ‘‘internal friction,’’ such as between molecules in polymers or between cells in wood. Here again, isotropy, linearity, and small strains allow for simple models.

Quadratic functions for the state potential and the dissipative potential lead to either Kelvin-Voigt or Maxwell’s models, depending upon the partition of stress or strains in a reversible part and in an irreversible part.

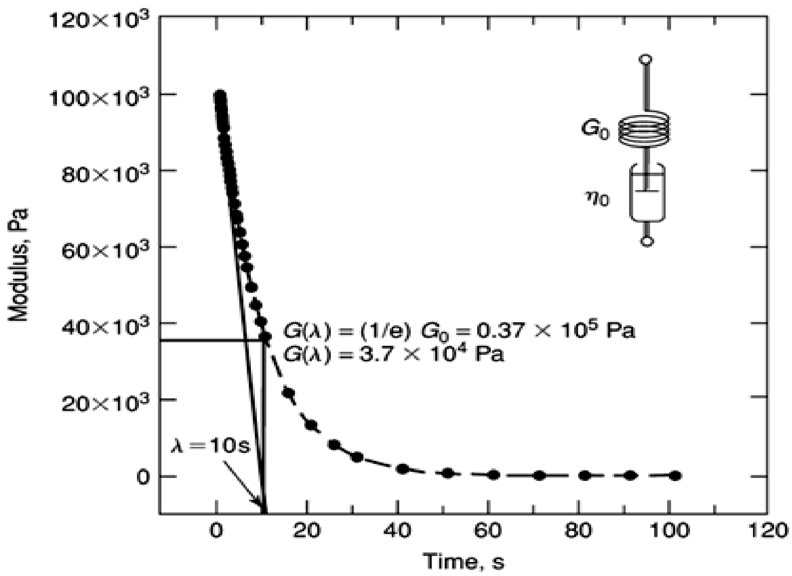
Maxwell model:



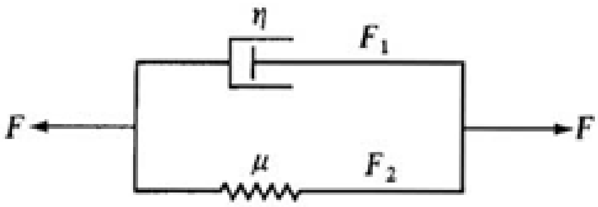
- Represented by a purely viscous damper and a purely elastic spring connected in series
- The model can be represented by the following differential equation:

$$\frac{d\epsilon_{Total}}{dt} = \frac{d\epsilon_D}{dt} + \frac{d\epsilon_S}{dt} = \frac{\sigma}{\eta} + \frac{1}{E}\frac{d\sigma}{dt}$$

- Predicts/models a stress that decays exponentially with time to zero with permanent deformation
- Model doesn’t accurately predict creep (constant stress). Predicts that strain will increase linearly with time. Actually strain rate decreases with time



Voigt model:

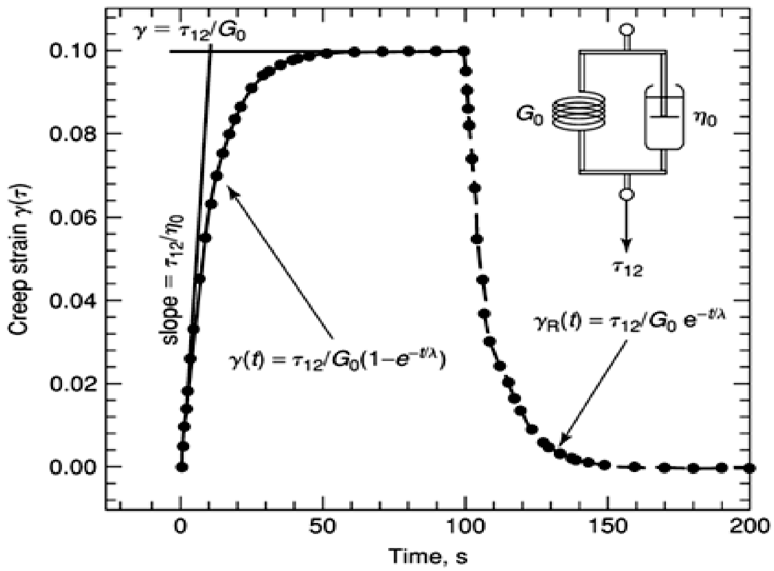


- Represented by a Newtonian damper and Hookean elastic spring in parallel.

- The model can be expressed as a linear first order DEQ

$$\sigma(t) = E\varepsilon(t) + \eta\frac{d\varepsilon(t)}{dt}$$

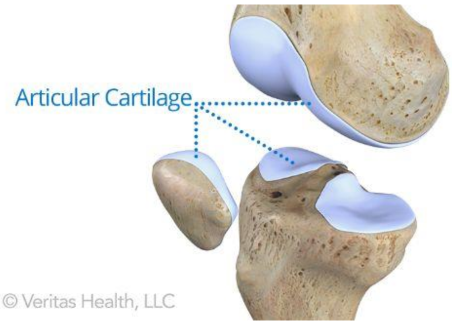
- Represents a solid undergoing reversible, viscoelastic strain.
- Models a solid that is very stiff but will creep (e.g. crystals, glass, apparent behavior of cartilage). At constant stress (creep), predicts strain to tend to σ/E as time continues to infinity
- The model is not accurate for relaxation in a material/tissue



Creep and recovery response

10. Describe the functions of articular cartilage.(N/D 2016)

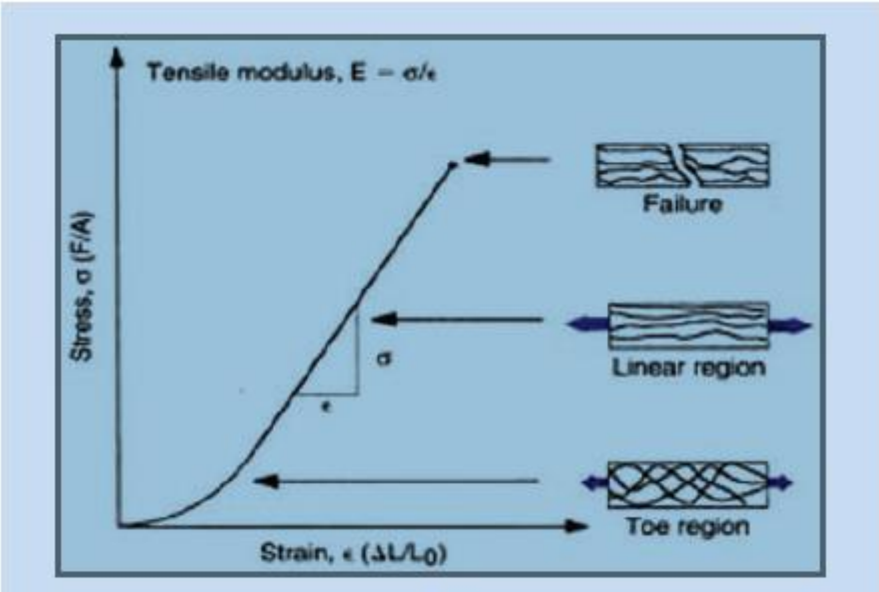
- Articular cartilage is the highly specialized connective tissue of diarthrodial joints. Its principal function is to provide a smooth, lubricated surface for articulation and to facilitate the transmission of loads with a low frictional coefficient .
- Articular cartilage is devoid of blood vessels, lymphatics, and nerves and is subject to a harsh biomechanical environment. Most important, articular cartilage has a limited capacity for intrinsic healing and repair.
- In this regard, the preservation and health of articular cartilage are paramount to joint health.
- Injury to articular cartilage is recognized as a cause of significant musculoskeletal morbidity.
- The unique and complex structure of articular cartilage makes treatment and repair or restoration of the defects challenging for the patient, the surgeon, and the physical therapist.
- The preservation of articular cartilage is highly dependent on maintaining its organized architecture.



BIOMECHANICAL FUNCTION

- Articular cartilage is a thin layer of specialized connective tissue with unique viscoelastic properties. Its principal function is to provide a smooth, lubricated surface for low friction articulation and to facilitate the transmission of loads to the underlying subchondral bone.
- Articular cartilage is unique in its ability to withstand high cyclic loads, demonstrating little or no evidence of damage or degenerative change.
- The biomechanical behavior of articular cartilage is best understood when the tissue is viewed as a biphasic medium.
- Articular cartilage consists of 2 phases: a fluid phase and a solid phase. Water is the principal component of the fluid phase, contributing up to 80% of the wet weight of the tissue. Inorganic ions such as sodium, calcium, chloride, and potassium are also found in the fluid phase.
- The solid phase is characterized by the ECM, which is porous and permeable.
- The relationship between proteoglycan aggregates and interstitial fluid provides compressive resilience to cartilage through negative electrostatic repulsion forces.
- The initial and rapid application of articular contact forces during joint loading causes an immediate increase in interstitial fluid pressure.
- This local increase in pressure causes the fluid to flow out of the ECM, generating a large frictional drag on the matrix.
- When the compressive load is removed, interstitial fluid flows back into the tissue.
- The low permeability of articular cartilage prevents fluid from being quickly squeezed out of the matrix.

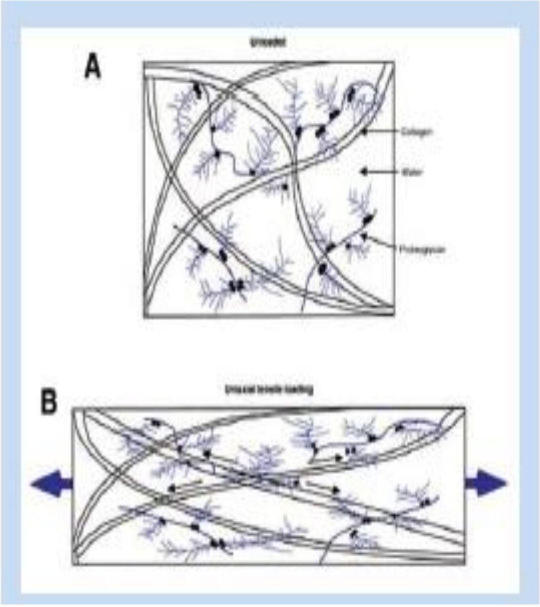
- The 2 opposing bones and surrounding cartilage confine the cartilage under the contact surface. These boundaries are designed to restrict mechanical deformation.
- Articular cartilage is viscoelastic and exhibits time-dependent behavior when subjected to a constant load or deformation.
- Two types of mechanisms are responsible for viscoelasticity in articular cartilage: flow dependent and flow independent.



stress-strain diagram for articular cartilage during tensile loading. The schematic representations on the right illustrate the orientation of the collagen fibrils in response to loading.

- The flow-dependent mechanism depends on interstitial fluid and the frictional drag associated with this flow.
- The drag resulting from the interstitial fluid is known as *biphasic viscoelastic behavior*.
- The flow-independent component of viscoelasticity is caused by macromolecular motion—specifically, the intrinsic viscoelastic behavior of the collagen proteoglycan matrix.
- As a result, the fluid pressure provides a significant component of total load support, thereby reducing the stress acting upon the solid matrix.
- Articular cartilage also exhibits a creep and stress-relaxation response.

- When a constant compressive stress is applied to the tissue, its deformation increases with time, and it will deform or creep until an equilibrium value is reached.
- Similarly, when cartilage is deformed and held at a constant strain, the stress will rise to a peak, which will be followed by a slow stress-relaxation process until an equilibrium value is reached.
- Because articular cartilage tends to stiffen with increased strain, it cannot be described by a single Young's modulus.
- Rather, the modulus of the tissue depends on the time at which the force measurement was taken during a stress-relaxation test, which was common practice in the preliminary studies of mechanical testing on articular cartilage.
- The current method is to apply a known strain, which is immediately followed by a peak in measured force and a slow stress-relaxation process; the force/stress value is recorded when it has reached equilibrium.



Schematic depiction of the main components of articular cartilage when the tissue is unloaded (A) and when tensile load is applied (B). When the tissue is loaded, collagen fibrils align along the axis of tension.

- This process is repeated across a range of strain values, and the equilibrium modulus is calculated as the slope of the stress-strain curve.
- The complex composition and organization of cartilage through the middle zones of cartilage contributes significantly to its shear-resistant properties.

- Stretching of the randomly distributed collagen fibrils provides cartilage with its shear stress response.
- The tensile force-resisting properties derive from the precise molecular arrangement of collagen fibrils.
- The stabilization and ultimate tensile strength of the collagen fiber are thought to result from the intra- and intermolecular cross-links.

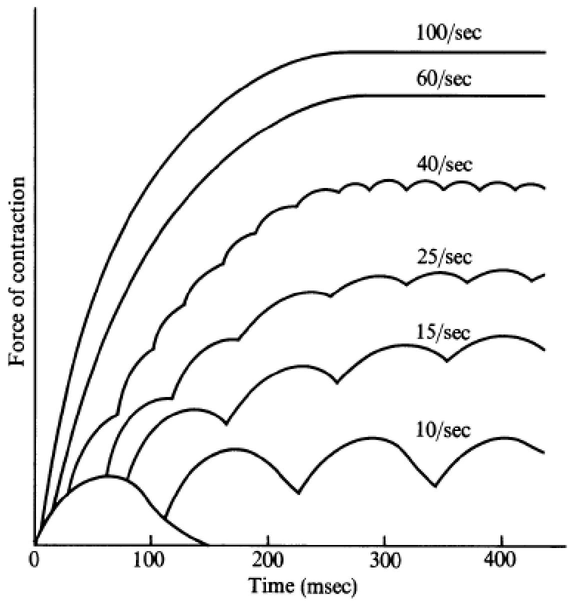
11. Discuss the factors that influence the force developed during muscular activity.(N/D 2016)

The Sliding Element Theory of Muscle Action

- Chemical and electron microscopic studies have revealed the fine structure of the myosin filaments. It is shown that each myosin filament consists of about 180 myosin molecules.
- Each molecule has a molecular weight of about 500 000 and consists of a long tail piece and a "head," which on close examination is seen to be a double structure.
- On further treatment the molecule can be broken into two moieties: *light meromyosin,* consisting of most of the tail, and *heavy meromyosin* representing the head with part of the tail.
- The myofilament is formed by the tails of the molecules which lie parallel in a bundle, with their free ends directed toward the midpoint of the long axis.
- The heads project laterally from the filament in pairs, at 180° to each other and at 14.3 nm intervals. Each pair is rotated by 120° with respect to its neighbors to form a spiral pattern along the filament.
- These heads seem to be able to nod: they He close to their parent filament in relaxation, but stick out to actin filaments when excited. They are called *cross-bridges.*

Single Twitch and Wave Summation

- Skeletal muscle responds to stimulation by nerve, electric, or chemical impulses. Each adequate stimulation elicits a single twitch lasting for a fraction of a second.

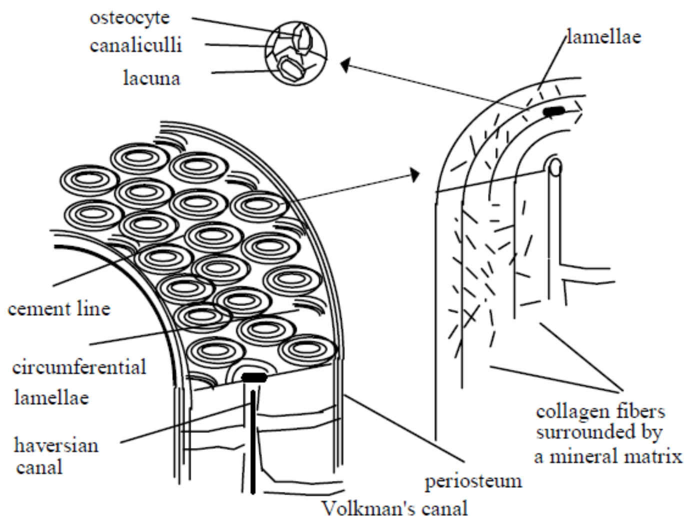


Wave summation and tetanization

- A single isometric twitch is shown in the lower left-hand corner. It is followed by successive twitches at varying frequencies.
- When the frequency of twitch is 10 per second, the first musc1e twitch is not completely over when the second one begins. This results in a stronger contraction. The third, fourth, and additional twitches add still more strength.
- This tendency for summation is stronger if the twitches come at a higher frequency. Finally, a *critical frequency* is reached at which the successive contractions fuse together and cannot be distinguished one from the other.
- This is then the *tetanized state.* For frequencies higher than the critical frequency further increase in the force of contraction is slight. Th
- is kind of reinforcement by successive twitches is called *wave summation.*

12. With a neat diagram explain the composition and mechanical property of cortical bone.(A/M 2017)

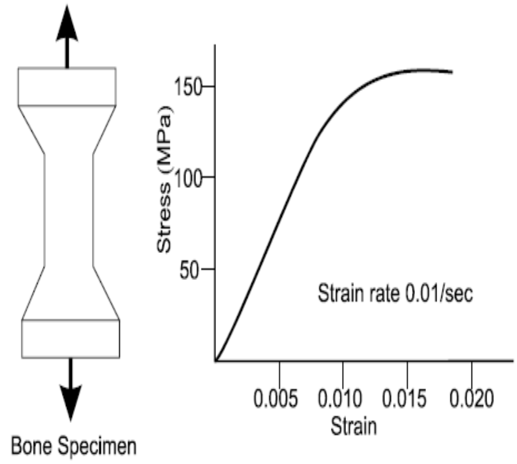
- Cortical bone is typically found in the shaft of long bones and as a shell around spongy bone.
- Regions within cortical bone are often described as either **haversian** or **laminar** .



Structure of cortical bone.

- The haversian type consists of an osteon with a central channel called the haversian canal.
- This canal contains blood vessels and nerves. Surrounding the canal are lamellae, lacunae and osteocytes, same as those described above for cancellous bone.
- The network of canaliculi allows blood and nutrients from the haversian canal to reach the osteocytes.
- Each osteon has a cement line at its periphery. Neither the canaliculi nor the collagen fibers of an osteon cross the cement line.
- A typical osteon is about 200 micrometers in diameter. Every point in an osteon is no more than 100 micrometers away from the centrally located blood supply.
- Osteons usually run longitudinally in long bones, but they branch often and intertwine with each other.
- The second type of cortical bone span the regions between osteons. These interstitial lamellae are continuous with the osteon and they are composed of the same material but in a different configuration.
- The properties of bone and other biological tissues depend on the freshness of the tissue.
- The properties of bone can change within a matter of minutes if allowed to dry out.

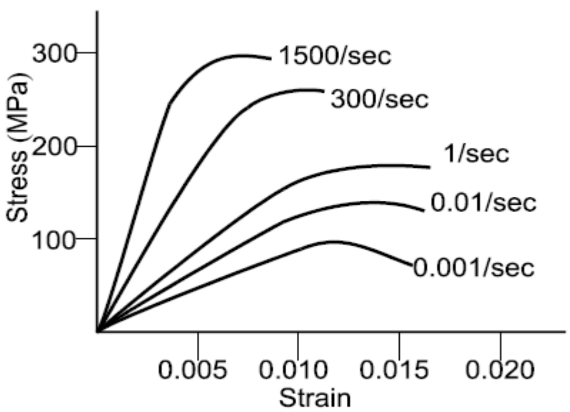
- Cortical bone has an ultimate strain of around 1.2 % when wet and about 0.4 % if the water content is not maintained.
- Mechanical testing consists of applying tensile, compressive, torsional, or shear loads to bone specimens and recording the deformation of the material.
- Force-deformation (structural properties) or stress-strain (material properties) curves can then be determined.



- Mechanical testing of bone is complicated because of the anisotropy of bone (requires multiple samples for testing) and the small sample sizes that are obtainable.
- For this reason efforts have been made to use ultrasonic techniques. These techniques make use of the relations between the speed of sound in a material and the elastic properties of that material.
- Ultrasound requires fewer and smaller specimens to characterize the material properties of bone.
- The disadvantage is that the relationship between the speed of sound and the tissues elastic properties must already be known.
- Bone shows a linear range in which the stress increases in proportion to the strain. The slope of this region is defined as Young's Modulus or the Elastic Modulus.
- Both steel and glass are stiffer and stronger than bone. Bone is strain rate sensitive and tends to be more strain-rate sensitive than other biological tissues.

Three important parameters that characterize some of the mechanical properties of bone:

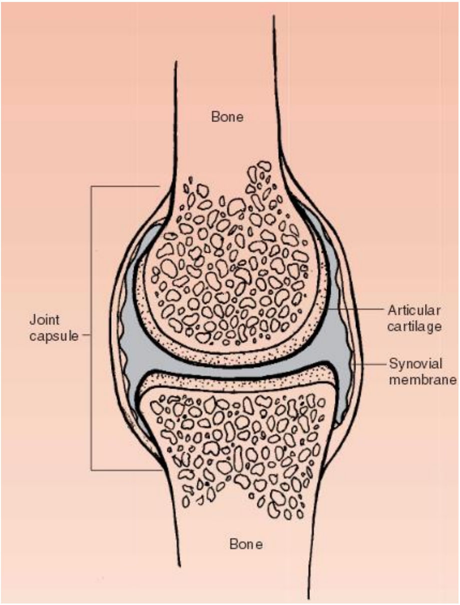
- Ultimate force, maximum deformation to failure, and energy that it can store before failing, can be obtained from a force-deformation curve.
- The ultimate force represents the maximum load that the bone can sustain before it breaks.
- The ultimate force varies depending on the type of load applied (e.g. tensile, compressive, shear) and the loading rate.
- The deformation at failure is self-explanatory and also depends on the loading rate and direction.
- The energy absorbed before failing can be calculated from the area under the force-deformation curve and therefore depends on both the ultimate force and the ultimate strain.
- Children's bones tend to absorb more energy before failure compared to adults (as much as 45 % more). Children's bones are weaker, but more compliant (children's bones can be 68% as stiff as adult bone).
- Material properties of the two types of bone differ. Cortical bone is stiffer than cancellous bone.
- It can sustain greater stress but less strain before failure. Cancellous bone can sustain strains of 75 % before failing in-vivo, but cortical bone will fracture if the strain exceeds 2 %.
- Cancellous bone has a greater capacity to store energy compared to compact bone.



Strain-rate sensitivity of cortical bone. As the strain-rate increases the ultimate stress increases and the ultimate strain decreases.

13. How the cartilages are classified? Explain its composition and mechanical property. (A/M 2017)

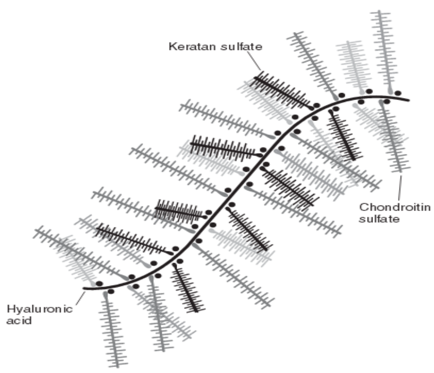
- The materials classed as cartilage exist in various forms and perform a range of functions in the body.
- Depending on its composition, cartilage is classified as articular cartilage (also known as hyaline), fibrocartilage, or elastic cartilage. Elastic cartilage helps to maintain the shape of structures such as the ear and the trachea.
- In joints, cartilage functions as either a binder or a bearing surface between bones.
- The annulus fibrosus of the intervertebral disc is an example of a fibrocartilaginous joint with limited movement (an amphiarthrosis).
- In the freely moveable synovial joints (diarthroses) articular cartilage is the bearing surface that permits smooth motion between adjoining bony segments.
- Hip, knee, and elbow are examples of synovial joints.
- In a typical synovial joint, the ends of opposing bones are covered with a thin layer of articular cartilage.



Schematic representation of a synovial joint.

Composition of articular cartilage

- Articular cartilage is a living material composed of a relatively small number of cells known as chondrocytes surrounded by a multicomponent matrix.
- Mechanically, articular cartilage is a composite of materials with widely differing properties.
- Approximately 70 to 85% of the weight of the whole tissue is water. The remainder of the tissue is composed primarily of proteoglycans and collagen. Proteoglycans consist of a protein core to which glycosaminoglycans (chondroitin sulfate and keratan sulfate) are attached to form a bottlebrush-like structure.
- These proteoglycans can bind or aggregate to a backbone of hyaluronic acid to form a macromolecule with a weight up to 200 million.
- Approximately 30% of the dry weight of articular cartilage is composed of proteoglycans.
- Proteoglycan concentration and water content vary through the depth of the tissue. Near the articular surface, proteoglycan concentration is relatively low, and the water content is the highest in the tissue.
- In the deeper regions of the cartilage, near subchondral bone, the proteoglycan concentration is greatest, and the water content is the lowest.
- Collagen is a fibrous protein that makes up 60 to 70% of the dry weight of the tissue. Type II is the pre-dominant collagen in articular cartilage, although other types are present in smaller amounts



A proteoglycan aggregate showing a collection of proteoglycans bound to a hyaluronic backbone. Proteoglycans are the bottlebrush-like structures consisting of a protein core with side chains of chondroitin sulfate and keratan sulfate. Negatively charged sites on the chondroitin and keratan sulfate chains cause this aggregate to spread out and occupy a large domain when placed in an aqueous solution.

Mechanical behavior

- In an aqueous environment, proteoglycans are polyanionic; that is, the molecule has negatively charged sites that arise from its sulfate and carboxyl groups.
- In solution, the mutual repulsion of these negative charges causes an aggregated proteoglycan molecule to spread out and occupy a large volume.
- In the cartilage matrix, the volume occupied by proteoglycan aggregates is limited by the entangling collagen framework.
- The swelling of the aggregated molecule against the collagen framework is an essential element in the mechanical response of cartilage. When cartilage is compressed, the negatively charged sites on aggrecan are pushed closer together, which increases their mutual repulsive force and adds to the compressive stiffness of the cartilage.
- Non-aggregated proteoglycans would not be as effective in resisting compressive loads, since they are not as easily trapped in the collagen matrix.
- Damage to the collagen framework also reduces the compressive stiffness of the tissue, since the aggregated proteoglycans are contained less efficiently.
- The mechanical response of cartilage is also strongly tied to the flow of fluid through the tissue. When deformed, fluid flows through the cartilage and across the articular surface.
- If a pressure difference is applied across a section of cartilage, fluid also flows through the tissue.
- These observations suggest that cartilage behaves like a sponge, albeit one that does not allow fluid to flow through it easily.

UNIT IV

BIOMECHANICS OF JOINTS AND IMPLANTS

- ✓ **Skeletal joints**
- ✓ **Forces and stresses in human joints**
- ✓ Analysis of rigid bodies in equilibrium
- ✓ Free body diagram
- ✓ **Design of orthopedic implants**
- ✓ **Manufacturing process of implants**
- ✓ **Biocompatibility**
- ✓ **Requirement of a biomaterial**
- ✓ Characteristics of different types of biomaterial
- ✓ **Fixation of implants**
- ✓ **Specifications for a prosthetic joint**
- ✓ **Biomechanical analysis of**
    - **Elbow**
    - **Shoulder**
    - **Spinal column**
    - **Hip Knee & Ankle**

LIST OF IMPORTANT QUESTIONS

PART A

1. **Define Biocompatibility.**
2. **What is sterilizability?**
3. **Define biomaterials. List its characteristics. (M/J 2016)**
4. **What are the requirements of biomaterials?**
5. **What the types of joint in skeletal system?(or) List the various types of joints? (N/D 2016)**
6. **How does joint angle affect the torque production capabilities of a muscle crossing that joint? (N/D 2016)**
7. **What factors influence the stability of synovial joints? (M/J 2016)**
8. **Give some examples of natural synovial joints. (N/D 2015)**
9. **What is an orthopedic implant?**
10. **Mention the steps in designing process of an orthopedic implant.**
11. **What is static equilibrium? (N/D 2015)**
12. **Enumerate the biomaterials used for making implants. (A/M 2017)**
13. **What is synovial joint? (A/M 2017)**
14. What is mean by a free-body diagram?
15. Describe fixation of implants?
16. Explain flexion and extension?
17. What is abduction and adduction?
18. Explain dorsiflexion and plantar flexion?
19. Mention the specification for a prosthetic joint.
20. Explain pronation and supination.

PART B

1. **Elaborate on the factors to be considered in the design of orthopedic implants. (M/J 2016), (N/D 2015), (N/D 2016)**
2. **Discuss about the mechanics of total knee replacement. (N/D 2015)**
3. **What is a joint? Describe the function of various types of joints. (M/J 2016)**
4. **Write a brief note on forces and stresses in human joints.**
5. **When you bend down to lift a heavy load off the floor, it is recommended to squat down (ie bending the keen) rather than just flexing the hips. Why? (N/D 2016)**
6. **What is free body diagram? Draw a free body diagram of women in skate. Describe about the forces acting on her.(N/D 2016)**
7. **Describe about the characteristics of biomaterials. (N/D 2016)**
8. **Explain biomechanical analysis physiological system or Explain the mechanics of (i) Shoulder (ii) Ankle (A/M 2017)**
9. **Explain about the materials and designs involved in making total hip replacement and elaborate its fixing procedures? (A/M 2017) or Explain fixation of implants.**
10. Explain the specifications for a prosthetic joint.
11. Explain analysis of a rigid body in equilibrium.

PART A

1. What is biocompatibility?

- Biocompatibility involves the acceptance of an artificial implant by the surrounding tissues and by the body as a whole.
- Biocompatible materials do not irritate the surrounding structures, do not provoke an abnormal inflammatory response, do not incite allergic or immunologic reactions, and do not cause cancer.

2. What is sterilizability?

- Sterilization is killing of microorganism; the material must be able to undergo sterilization. Sterilization techniques include dry heat, gamma, gas (ethylene oxide (ETO)) and steam autoclaving.
- Sterilizability of biomedical polymers is an important aspect of the properties because polymers have lower thermal and chemical stability than other materials such as ceramics and metals; consequently, they are also more difficult to sterilize using conventional techniques.

3. Define biomaterials. List its characteristics. (M/J 2016)

A biological or synthetic substance which can be introduced into body tissue as part of an implanted medical device or used to replace an organ, bodily function, etc.

- Hard Materials.
- Flexible Material.
- Must not react with any tissue in the body.
- Must be non-toxic to the body.
- Must not be biodegradable.

4. What are the requirements of biomaterials?

- It must be biocompatible
- non-carcinogenic
- corrosion-resistant

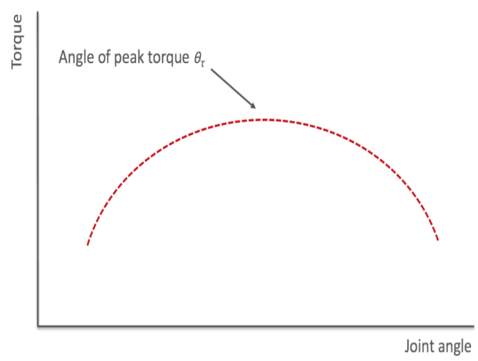
- has low toxicity and wear
- Long fatigue life
- Adequate Strength
- Modulus equivalent to that of bone

5. What the types of joint in skeletal system? (or) List the various types of joints? (N/D 2016)

i. Immovable (or fixed) joints
ii. Slightly movable joints
iii. Movable (or synovial) joints

6. How does joint angle affect the torque production capabilities of a muscle crossing that joint? (N/D 2016)

The joint angle where we are strongest is called the *angle of peak torque.*



This curve is the "joint torque-angle" relationship. There are lots of factors that contribute, both peripheral and central, including:

- Moment arm length of the muscle
- Normalized fiber length
- Regional muscle size
- Tendon stiffness
- Muscle stiffness
- Neural drive

7. What factors influence the stability of synovial joints? (M/J 2016)

- **The shape of the articular surfaces of the bones**
- **The ligaments** strong bands of dense fibrous connective tissue which bind the adjacent bones together
- **Muscles** which extend between the two bones comprising the joint.

8. Give some examples of natural synovial joints. (N/D 2015)

- Elbow
- Knee
- Fingers
- Hip
- Wrist joint
- Carpals of the wrist
- Radioulnar joint

9. What is an orthopedic implant?

- An orthopedic implant is a medical device manufactured to replace a missing joint or bone or to support a damaged bone.
- The medical implant is mainly fabricated using stainless steel and titanium alloys for strength and the plastic coating that is done on it acts as an artificial cartilage.
- The implant fixation can be done by both internally and externally.
- Eg. of internal fixation- Plates and screw, intramedullary nails etc.,
- Eg. of external fixation- Ilizarov external fixator,pins etc.,

10. Mention the steps in designing process of an orthopedic implant.

- Design (or new idea) and prototyping
- Milling
- Finishing
- Inspection
- Passivation
- Packing

- Sterilization

11. What is static equilibrium? (N/D 2015)

Static equilibrium is a form of equilibrium that occurs when an object is at rest. "Static" refers to the object being motionless while "equilibrium" refers to the object either having no net forces acting upon it or having all of its net forces balanced.

12. Enumerate the biomaterials used for making implants. (A/M 2017)

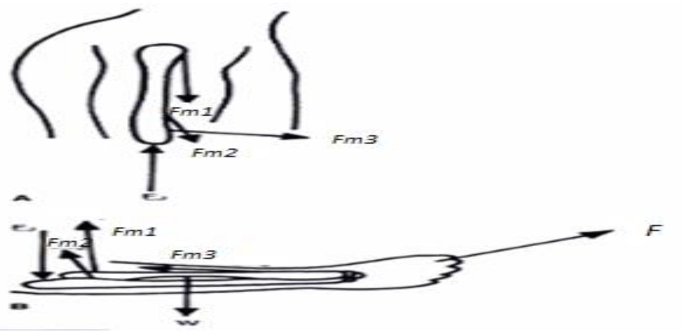
- **Metal implants:** Gold, Lead, Iridium, Tantalum, stainless steel and cobalt alloy
- **Polymers:** ultrahigh molecular weight polyurethane, polyamide, polymethylmethacrylate resin, polytetrafluoroethylene, and polyurethane
- **Others:** zirconia, roxolid, surface modified titanium implants

13. What is synovial joint? (A/M 2017)

- A **synovial joint**, also known as diarthrosis, joins bones with a fibrous **joint** capsule that is continuous with the periosteum of the joined bones, constitutes the outer boundary of a **synovial** cavity, and surrounds the bones' articulating surfaces.
- The**synovial** cavity/**joint** is filled with **synovial** fluid.

14. What is mean by a free-body diagram?

Free- body diagrams are constructed to help identify the forces and moments acting on individual parts of a system and to ensure the correct use of the equations of mechanics to analyze the system. For this purpose, the parts constituting a system are isolated from their surroundings and the effects of surroundings are replaced by proper forces and moments.



Forces involved at and around the elbow joint and the free body diagram of the lower arm

15. Describe fixation of implants?

Implant fixation mainly divides in to two. They are

- Internal fixation- The implants are placed internally to support the bone.
  Eg. - Plates and screw, Intramedullary nails etc.,
- External fixation- The implants are placed externally to support bone.
  Eg. - Ilizarov external fixator, pins etc.,

16. Explain flexion and extension?

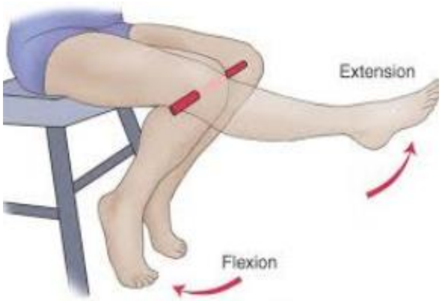
Flexion

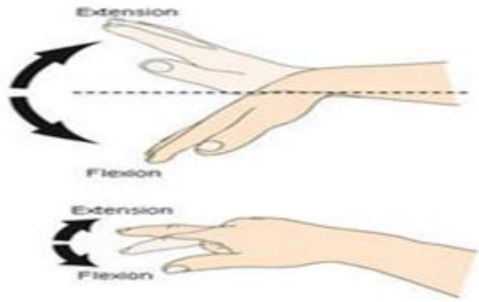
- Flexion describes a bending movement that decreases the angle between a segment and its proximal segment.
- Bending the elbow, or clenching a hand into a fist, are examples of flexion. When sitting down, the knees are flexed. When a joint can move forward and backward, such as the neck and trunk, flexion refers to movement in the anterior direction. Flexion of the shoulder or hip refers to movement of the arm or leg forward.

Extension

- Extension is the opposite of flexion, describing a straightening movement that increases the angle between body parts.
- When a joint can move forward and backward, such as the neck and trunk, extension refers to movement in the posterior direction. For example, when

standing up, the knees are extended. Extension of the hip or shoulder moves the arm or leg backward. When the chin is against the chest, the head is flexed, and the trunk is flexed when a person leans forward.





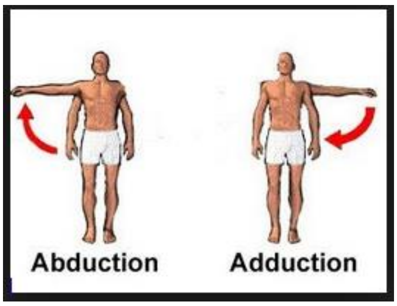
17. What is abduction and adduction?

Abduction

- Abduction refers to a motion that pulls a structure or part away from the midline of the body.
- For example, raising the arms up, such as when tightrope-walking, is an example of abduction at the shoulder. When the legs are splayed at the hip, such as when doing a star jump or doing a split, the legs are abducted at the hip.

Adduction

- Adduction refers to a motion that pulls a structure or part toward the midline of the body, or towards the midline of a limb.
- Adduction of the wrist is also called ulnar deviation. For example, dropping the arms to the sides, or bringing the knees together, are examples of adduction

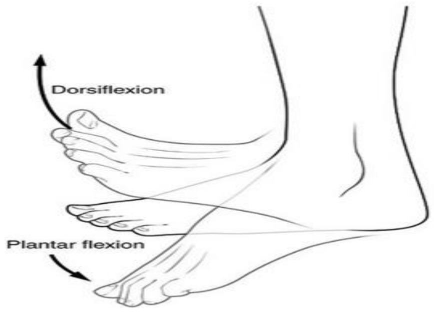


18. Explain dorsiflexion and plantar flexion?

Dorsiflexion and plantar flexion refers to extension or flexion of the foot at the ankle.

Dorsiflexion where the toes are brought closer to the shin (the front part of the leg from the knee to the ankle). This decreases the angle between the dorsum of the foot and the leg. For example, when walking on the heels the ankle is described as dorsiflexion.

Plantar flexion is the movement which decreases the angle between the sole of the foot and the back of the leg. For example, the movement when depressing a car pedal or standing on the tiptoes can be described as plantar flexion



19. Mention the specification for a prosthetic joint.

1. Appropriate articulation
2. Good stability
3. Adequate strength
4. Good fixation
5. Correct choice of materials
6. Low friction forces
7. Acceptable wear rate
8. Good salvage potential
9. Standardization
10. Sterilization
11. Cost effectiveness
12. Surgical instrumentation

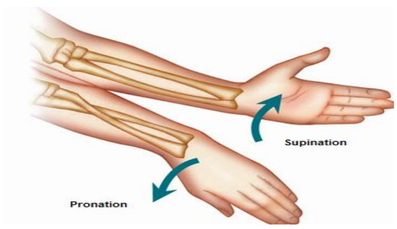
20. Explain pronation and supination.

Pronation:

- Pronation refers to the inward roll of the foot during normal motion and occurs as the outer edge of the heel strikes the ground and the foot rolls inward and flattens out.
- A moderate amount of pronation is required for the foot to function properly, however damage and injury can occur during excessive pronation.

Supination:

- Supination (or under-pronation) is the opposite of pronation and refers to the outward roll of the foot during normal motion.
- Supination occurs during the running gait as the heel lifts off the ground and the forefoot and toes are used to propel the body forward.



PART B

1. Elaborate on the factors to be considered in the design of orthopedic implants. (M/J 2016), (N/D 2015), (N/D 2016)

Most implant requires custom designed surgical instruments and it is vital to consider the product design specification for the instruments at an early stage as they interact with the implant. Any design changes to the implant will have a knock-on effect for the design of the instrument. In many cases the complexity and design time of the surgical instruments are greater than the actual implant itself. The standard BS EN 12011 has been produced to help identify the requirements for surgical instruments. It is also important to consider the packaging, sterilization.

Design requirements

The design requirements for the implant device will include:

- Intended performance
- Design attributes
- Materials
- Design evaluation
- manufacture
- Testing
- Instruments required
- Sterilization
- Packaging
- Information to be supplied by the manufacturer

Design Reviews

A design review is required, at each stage of the design process, to formally document comprehensive, systematic examination of a design to:

- Evaluate design requirements
- Assess capability of the design
- Identify problems

Concept Design

The concept, or conceptual, design stage is where solutions are generated to meet the design requirements. The aim is to generate as many ideas as possible

Concept design may involve:

- Simple sketches of ideas
- Computer aided design models
- Analytical calculations
- Initial manufacturer consultation.

Design Verification

Design verification involves confirmation by examination that a medical device meets the design requirements It is essential that verification is considered early in the design process. Ideally, when a design requirement is decided, a method for verifying that requirement should also be developed. Design verification is possible by one of the method is the rapid prototyping.

Rapid prototyping

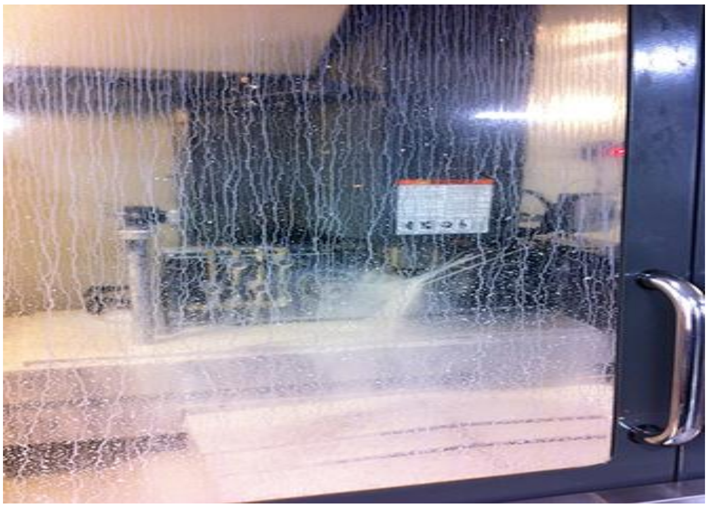
The design process usually involves a 3D CAD mockup

Rapid prototyping is a very effect technique for verifying the design of medical devices as it aids communication between engineers and surgeons. Models of implants can be produced within hours by a variety of methods such as Selective Laser Sintering (SLS), Fused Deposition Modeling (FDM) or three-dimensional printing .The surgeon can then inspect the models and size them against a skeleton. Rapid prototyping can also be used to produce models of human bones, which is particular useful if there is abnormal anatomy into which an implant will be used. Multi-component systems can also benefit from the use of rapid prototyping as they allow designers to evaluate the interaction between components, such as a screw and a screw driver, thus minimizing the opportunity for component incompatibility.

Milling

Every implant and instrument starts from a solid block of metal or PEEK plastic

After the designing process, pretty much all the implants and instruments were produced from a single block of metal or PEEK composite plastic. The metal blocks (or rods) are secured to the inside of large industrial multi-axis milling machines (CNC's). These milling machines consist of a number of robotic arms and drilling tips that can carve very fine features. They're basically the same machines that you would find carving automobile parts, except on a smaller scale. A white, milky looking cooling liquid is aggressively sprayed onto the drill bit as it's carving the part to cool it, as well as to remove metal shards.

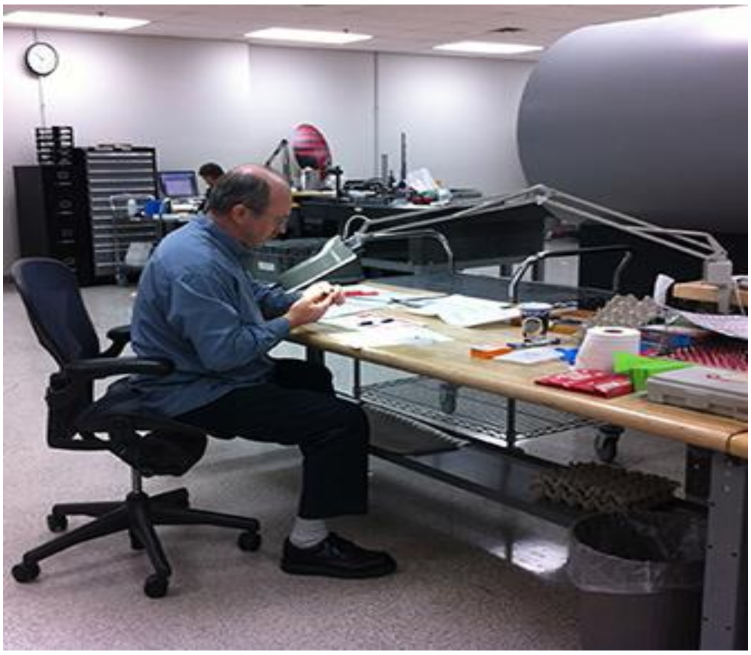


An implant being milled and sprayed with coolant

Often, the five most important faces of a block of metal are milled, with the sixth kept intact as the base. After milling is complete, a wire electrical discharge machining (EDM) cutter removes the base, and the sixth face is milled. It’s important that as much of the implant or instrument is milled at a time; moving a partially finished piece between different machines or adjusting their position could potentially cause variations between parts, and also naturally increases the amount of time it takes to produce a finished part.

Finishing, inspection, and passivation

A robot checks a part for precise measurements



A quality technician also gives the parts a look over

Though milling is done with industrial robots, most of the remaining tasks are done by the hands of skilled machinists and technicians. If the implant or instrument requires a dull or matte finish, it next goes to a sandblaster to remove the metal’s shine. Technicians also will use a rotary sander to deburr and finish the part. Around this time, the parts will also be laser-etched with information, such as the part’s ID number and manufacturing date. A sampling of the finished parts then goes to the quality lab, where technicians use automated instruments, as well as their eyes and hands, to meticulously check various properties of the parts. We also saw a device called an optical comparator that projects a magnified silhouette of a finished piece onto a screen which is compared with a schematic of the piece printed on mylar. Finally, all the parts are then cleaned ultrasonically using water, and then passivated using nitrate or citrate to prevent corrosion.

Packing and sterilization

Our finished medical devices are almost done, but unlike automobile parts, these parts can’t simply be placed into plastic bags or cardboard boxes to be shipped out. Larger companies often will sterile package the finished parts in-house, however, some of the smaller companies we visited outsourced this step to another firm nearby.

Lotron's electron-beam technology blasts packaged parts with electricity to completely sterilize them

To ensure that the packaged, finished goods are sterile enough to be used in a surgical environment, they're often sent to a third-party to irradiate the parts using radiation or gas. Over at Iotron, pallets of parts move down a long conveyor belt into an area called the "shield", which is a concrete-lined, maze-like passageway with a giant electron beam gun at the heart of it.

<u>Manufacturing of an orthopedic implant</u>

7-Axis High Speed Contour Milling:

- Machining of complex parts from bar or flat stock material.
- Dedicated equipment for production of implant grade PEEK components.
- Automated manufacturing process allows for single operation machining of complete 6-sided part and eliminates the need for secondary operations and fixturing requirements.
- Reduced cycle time and tighter tolerance compared to conventional 5-Axis machines.

Swiss Micromachining:

- Experienced in process development and production handling of small diameter, high precision orthopedic implants and devices.

  13-Axis manufacturing with gun drilling, thread whirling, small hole broaching, and 3D profiling capabilities.

Automated Deburring & Polishing:

- Capable of producing a 2 Ra microfinish on all metallic components, including stainless steel, titanium, nitinol, cobalt chrome, etc.
- Efficient process utilizing a wide variety of engineered burnishing compounds and high energy, driven-media finishing technology.
- Equipment available for both process development and production.

  Programmable Logic Control eliminates operator variation and allows for a validated process.

Inspection Equipment & Quality Assurance:

- Multi sensor systems with laser, optical and touch probe for full three-dimensional measurement of 5-axis parts, such as bone fixation plates.
- Process validation per ISO 13485:2012.

Clean room Packaging:

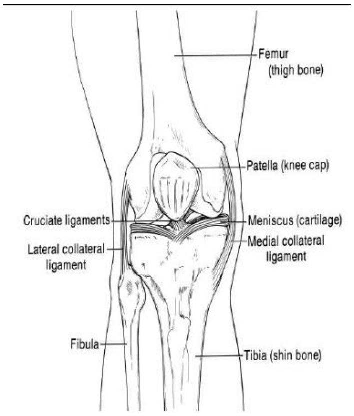
- ISO Class 8 room with adaptable microenvironments for tighter particulate control.
- Controls temperature, humidity & microbial growth.
- Lot specific Endotoxin reports available

2. Discuss about the mechanics of total knee replacement. (N/D 2015)

A total knee replacement is an operation to replace the worn or damaged parts of your knee joint. The surfaces of the diseased joint are removed and replaced with a mechanical, artificial joint that is called a prosthesis. This surgery can relieve the pain and stiffness in your knee joint. Pain in your knee or leg prevents you from doing your usual activities. Your xrays show irregular surfaces at the knee. When more conservative treatments such as medication and physical therapy can no longer relieve your pain and disability, it is time to consider having a total knee replacement. Your orthopaedic surgeon will encourage you to use your new joint as soon as possible after your operation. Patients often stand and begin walking the day of or the day after surgery. Physical therapy (PT) will begin in the PT department the afternoon of surgery or the day after. You will walk with a walker, then crutches or cane at first as you recover. Most patients have some temporary pain after joint replacement as the tissues heal and muscles regain strength. This pain should go away in a few weeks or months. Pain medication will be ordered for you and your pain level will be monitored. Your health care team will make every effort to keep you comfortable. With your new knee replacement and the help of your orthopaedic team, you may be able to resume some of the activities you once enjoyed. You may be permitted to go on long walks, dance, play golf, garden and ride a bicycle. Total joint replacement has an excellent track record for improving quality of life, allowing greater independence and reducing pain.

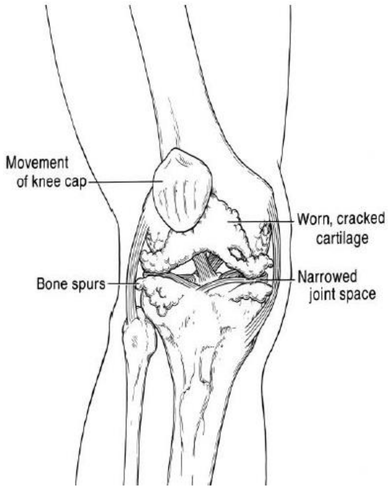
The Normal Joint

Your knee is a hinge joint where the end of the thigh bone (femur) meets the beginning of the large bone in your lower leg (tibia). A healthy knee has smooth cartilage that covers the ends of the femur and tibia. The smooth cartilage lets the surfaces of the two bones glide smoothly as you bend your knee. The muscles and ligaments around the knee joint support your weight and help move the joint smoothly so you can walk without pain.



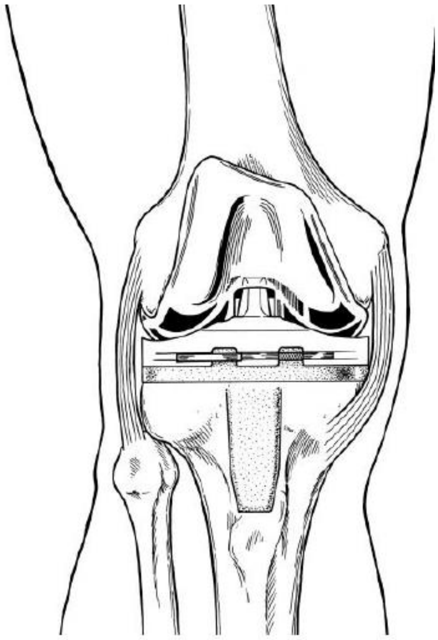
The Degenerated Knee Joint

The smooth cartilage layers can wear down on the ends of the femur and tibia. This degeneration can happen because of injury, arthritis, or as a side effect from medicines, such as steroids. When the smooth surfaces become rough, the surfaces are like sandpaper. Instead of the joint gliding when you move your leg, the bones grind and you have pain and / or stiffness. When pain in your knee or leg prevents you from doing your usual activities and your x-rays show irregular surfaces at the knee, your doctor might suggest that you have a knee replacement.



Your Replacement Knee Prosthesis

To create a new knee joint, the ends of the bones forming the joint are surgically removed. They are replaced with parts like the pieces shown here. The parts of the prosthesis are made of metal and very strong plastic. The pieces may be cemented in place with a special bone cement, or the metal may have a porous surface that bone will grow into to create a tight fit.



3. What is joint? Explain types of joint present in the human body(M/J 2016).

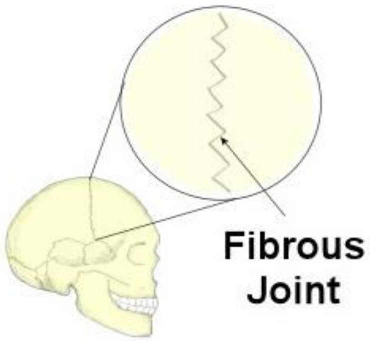
Joint is the area where two bones are attached for the purpose of permitting body parts to move. A joint is usually formed of fibrous connective tissue and cartilage.

A need for strength makes the bones rigid, but if the skeleton consisted of only one solid bone, movement would be impossible. Nature has solved this problem by dividing the skeleton into many bones and creating joints where the bones intersect. Joints, also known as articulations, are strong connections that join the bones, teeth, and cartilage of the body to one another. Each joint is specialized in its shape and structural components to control the range of motion between the parts that it connects.

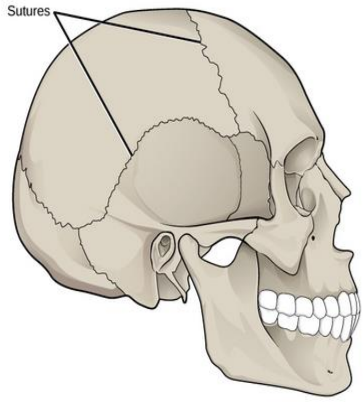
A joint is the point where two or more bones meet. There are three main types of joints;

- ✓ **Fibrous** (immoveable),
- ✓ **Cartilaginous** (partially moveable) and
- ✓ **Synovial** (freely moveable) joint

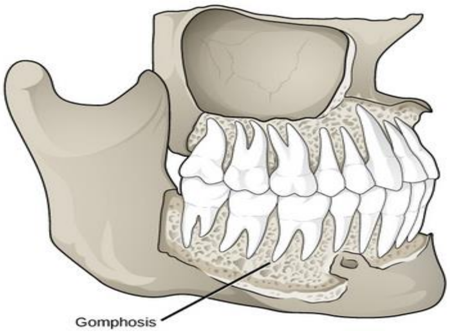
Fibrous joints (immoveable)



The bones of fibrous joints are held together by fibrous connective tissue. There is no cavity, or space, present between the bones, so most fibrous joints do not move at all. There are three types of fibrous joints: sutures, syndesmoses, and gomphoses. **Sutures** are found only in the skull and possess short fibers of connective tissue that hold the skull bones tightly in place.



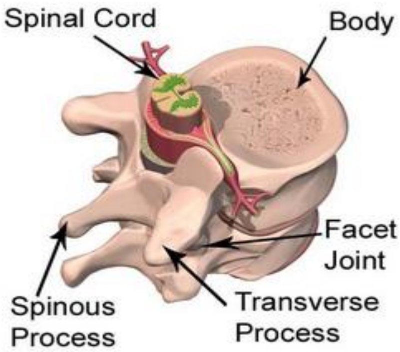
Syndesmoses are joints in which the bones are connected by a band of connective tissue, allowing for more movement than in a suture. An example of a syndesmosis is the joint of the tibia and fibula in the ankle. The amount of movement in these types of joints is determined by the length of the connective tissue fibers. **Gomphoses** occur between teeth and their sockets; the term refers to the way the tooth fits into the socket like a peg. The tooth is connected to the socket by a connective tissue called the periodontal ligament. Fibrous joints classified as synarthroses, or immovable, include: sutures, gomphoses, and synchondroses.



Cartilaginous (partially moveable)

Cartilaginous joints are those in which the bones are connected by cartilage. There are two types of cartilaginous joints:

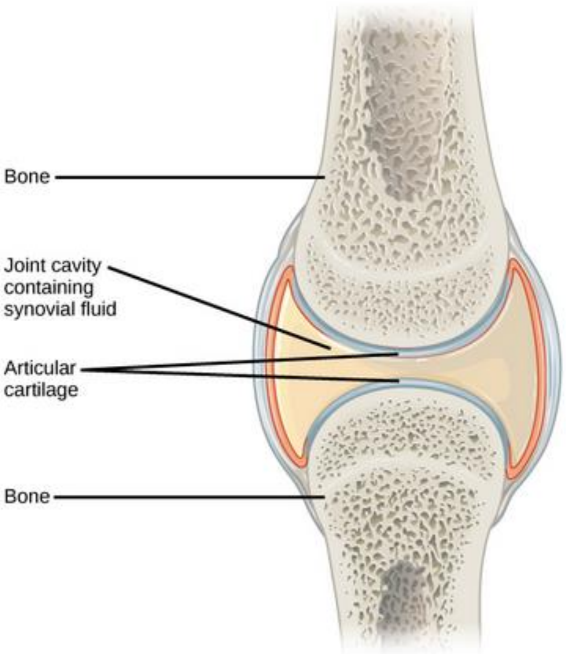
- **Synchondroses** are temporary joints which are only present in children, up until the end of puberty. For example the epiphyseal plates in long bones.
- **Symphesis** joints are permanent cartilaginous joints, for example the pubic symphesis. Example for cartilaginous joint is vertebrae in the spine.



A cartilagenous joint between two vertebrae

Synovial Joints (freely moveable)

Synovial (diarthrosis): Synovial joints are by far the most common classification of joint within the human body. They are highly moveable and all have a synovial capsule (collagenous structure) surrounding the entire joint, a synovial membrane (the inner layer of the capsule) which secretes synovial fluid (a lubricating liquid) and cartilage known as hyaline cartilage which pads the ends of the articulating bones. There are 6 types of synovial joints which are classified by the shape of the joint and the movement available.



Types of Synovial Joint

Hinged Joint

Hinged joints include the elbow, fingers, toes and knee. Movement occurs in only one direction or one plane. Innerbody.com states that the hinged joint in the knee is unusual because it allows the knee to swivel, turning the foot from side to side.

Pivot Joint

A pivot joint is a synovial joint designed with one end fitting like a cylinder inside a ring. Pivot joints at the base of the skull allow the head to rotate. Other pivot points allow the rotation of the palm.

Ball-and-Socket Joint

It describes a ball-and-socket joint as one in which the rounded surface of a bone fits into and moves within a cup-shaped depression. Examples of this type of synovial joint are the hip and shoulder joints. The ball-and-socket joint allows freedom of movement up, down, right, left and in a full 360-degree rotation.

Saddle Joint

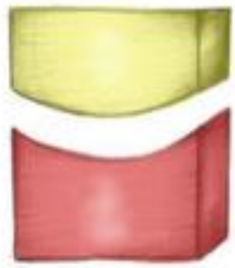
The saddle joint is a biaxial joint that allows movement on two planes--flexion/extension and abduction/adduction. The thumb is the only bone structure in the human body with a saddle joint

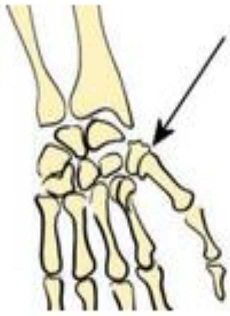
Condyloid Joint

Condyloid joints are biaxial joints that permit up, down and side-to-side motions. Medical-look.com lists the radiocarpal joint in the wrist as an example of a condyloid joint.

Gliding Joint

Gliding joints allow two or more flat or slightly rounded bones to move easily together without friction or grinding. Enotes.com, an online nursing encyclopedia, states the function of a gliding joint is to allow motions such as smooth sliding of bone past bone, bending, stretching and circular motion. Examples of gliding joints include the forearm to wrist joint and the lower leg to ankle joint.

Joint Type	Movement at joint	Examples	Structure
Hinge	Flexion/Extension	 Elbow/Knee	 Hinge joint
Pivot	Rotation of one bone around another	 Top of the neck (atlas and axis bones)	 Pivot Joint
Ball and Socket	Flexion/Extension/Adduction/ Abduction/Internal & External Rotation	 Shoulder/Hip	 Ball and socket joint

Saddle	Flexion/Extension/Adduction/ Abduction/Circumduction	 CMC joint of the thumb	 Saddle joint
Condyloid	Flexion/Extension/Adduction/ Abduction/Circumduction	 Wrist/MCP & MTP joints	 Condyloid joint
Gliding	Gliding movements	 Intercarpal joints	 Gliding joint

4. Write a brief note on forces and stresses in human joints.

Forces and Stresses

The application of the physical laws of force and motion to the body is termed biomechanics.

In physics, a force is any external agent that causes a change in the motion of a free body or that tries to deform a fixed body (create internal stress).

The Shoulder joint

The stresses exerted on the bones in a shoulder-hand system during the pull-up motion onto a bar, a common fitness activity, due to both external loads and strained muscles. The maximum stress values represent the effect of strained muscles, which was first taken into consideration.

During the design of an implant at the shoulder end of the humerus bone, the stress distribution must be taken into consideration, particularly with respect to the strength of the bones. It is also necessary to take into account the stress fluctuations in the human body occurring during daily activities; fluctuating stresses acting on an implant may create fatigue cracks at the shoulder end of the humerus and, additionally, may impose a high stress concentration factor due to shape and cross-sectional size changes in the humerus bone. As a result, the stress distribution in the humerus bone calculated in this study may help in the appropriate design of an implant for a humerus bone.

The hip joint

The mechanics of the human hip joint was found that the inter-cartilaginous space was variable in size and location, and that it changed shape with different positions of the joint. It was concluded that this space would be of limited value as a load-distribution mechanism and that cartilage should be well able to distribute the applied load.

The loading characteristics of the hip joint have been discussed for many years, and several hypotheses have been produced. Testing of these hypotheses, however, has been impeded by the lack of an easily and reliably performed technique for measuring the load over the whole articular surfaces. Fujifilm prescale meets this need and provides quantified values for load at any point.

If the femoral head and acetabulum were perfectly congruent, or incongruent but symmetrical, then the pressure prints would be uniform and regular in shape. Hips from children and young adults to see if loading patterns change with age. But even at this stage we can identify an area of high pressure in the antero-superior segment of the joint. Whatever the reason for this loading pattern, whether developmental or acquired, the site coincides with the most usual location of the progressive cartilage loss seen in osteoarthritis.

The knee joint

There is a lack of fundamental information on the knee biomechanics in deep flexion. Mechanical loads during activities requiring deep flexion were quantified on normal knees and compared with those in walking and stair climbing. The deep flexion activities generate larger net quadriceps moments and net posterior forces than routine ambulatory activities. Moreover, the peak net moments and the net posterior forces were generated between 90^{0} and 150^{0} of flexion.

The large moments and forces will result in high stress at high angles of flexion. These loads can influence pathological changes to the joint and are important considerations for reconstructive procedures of the knee. The posterior cruciate ligament should have a substantial role during deep flexion, since there was a large posterior load that must be sustained at the knee. The mechanics of the knee in deep flexion are likely a factor causing problems of posterior instability in current total knee arthroplasty. Thus, it is important to consider the magnitude of the loads at the knee in the treatment of patients that commonly perform deep flexion during activities of daily living.

The spinal column

Four elemental forces of the spine: compression, tension, torque, and shear.

- **Compression and tension**

Forces acting along the axis of the spine can apply compression or tension. As implied, compressive forces act to flatten the disks and vertebrae. Tension acts to elongate them. Forces develop from many internal and external sources, not just gravity, although gravity is a persistent force.

Eg: Jessica is beginning her jump. As she accelerates upward, her spine experiences a compressive force. As she hangs freely from the bar, gravity exerts tension on her spine.

- **Shear forces**

Forces acting perpendicular to the axis of the spine apply a shearing force that tries to slide the components away from their normal axis. Stresses develop in the interior of the structure. If the shear forces are great enough, ligament and disk tears may result as well as shear fractures of the vertebrae.

Eg: As Michael leans from side to side, gravity develops shear stresses on his spine. If he were holding weights in his hands, the forces would be greater. (Movement exaggerated.)

- **Torsional forces**

Forces acting perpendicular to the axis of the spine, but not in the plane of the spine, apply a rotational force that tries to rotate the components around their normal axis. This rotational force is formally called torque. Stresses develop in the interior of the structure. If the torsional forces are great enough, ligament and disk tears may result, as well as torsional fractures of the vertebrae.

The disc is subjected to a variety of loads: compression by gravity, pushing; traction by being pulled; tensile stress by flexion, extension, lateral bending; torsion and shear by axial rotation; and various multiple combinations of each. Since the back muscles use the spine as a lever in maintaining balance, the forces upon the IVD's (intervertebral discs) and vertebrae are much greater than the forces of the body weight above.

5. When you bend down to lift a heavy load off the floor, it is recommended to squat down (ie bending the keen) rather than just flexing the hips. Why? (N/D 2016)

No matter how hard you try, avoiding all the situations that may be stressful to your back is impossible. Sometimes you have to lift heavy objects. Of course, as the weight of the object increases, so does the risk of injuring your back. However, the actual stress on your back is also related to the position of your body when you lift the object. Understanding how body positioning affects your activities is called body mechanics. You can decrease your chance of a back strain or injury by using good body mechanics.

Before you lift any object, you must first make sure that you are capable of lifting it. You can safely lift only a certain amount of weight. Even if you are extremely strong and you can lift 300 pounds by yourself, attempting to lift 310 pounds would be futile and hazardous -- your chance of injury increases dramatically with every pound over your limit. So the first thing to do when lifting an unfamiliar object is to test its weight, or load. Pushing it with your foot is usually enough to give you an idea.

Protecting Your Back When Moving Objects

You can do just as much damage transporting an object as you can trying to lift it. Here's a few safety tips to keep in mind.

Push, Don't Pull

Moving heavy objects on a cart is, of course, much less strenuous than carrying them, but even with a cart, you can still hurt your back if you're not careful. As a rule, it is safer to push an object than pull it. When you push, you use the strength of your legs and your back to move the object; you can really get your weight behind it. When you pull, the tendency is to stand flat-footed and to yank, relying solely on your back without using the leg muscles. Also, the back is often in a poor position when pulling, increasing the risk of a strain. Next time you have a choice, remember to stand tall, lean into the object, use your legs and arms, keep your head up, and push.

Diet and Back Pain

Most people see no relation between what they eat and their back's strength and health. Just like the engine in your car, your body needs the proper nutrition so that the muscles can continue to move and support the spine. If your car runs out of gas, the engine quits and the car cannot move. If you have not eaten, your back muscles may quit working for you; they can weaken, tighten up, and become more susceptible to fatigue-related injury.

Eat Power Foods

Clearly, your muscles need food to maintain their vigor, but not just any food. The kind of food you eat matters. In our fast-paced society, eating right can be difficult. Fast food may satisfy your hunger and may even give you an energy boost, but your muscles and your body need power foods. Power foods are the ones that provide a great deal of energy slowly, over the course of a few hours. They can keep your muscles constantly supplied with the fuel they need to maintain the support and protection of your spine. Power foods are high in complex carbohydrates and low in simple sugars and fat and contain an adequate amount of protein.

6. What is free body diagram? Draw a free body diagram of women in skate. Describe about the forces acting on her.(N/D 2016)

Free body diagrams are used to show which forces are acting on a body at a particular instant in time. Arrows indicate the position, direction and size of the force acting. The most likely forces acting on a sports performer are friction, air resistance, weight and reaction forces.

Physics of skating

- For the most part, the physics behind ice skating comes down to analyzing the movement of skates over the ice.
- The skates do two things: They glide over the ice and they push off the ice with the edge, which causes a gain in speed. With practice, this combination of movements can become as effortless as walking. Another part of the physics is the low friction of the skate blade with the ice.
- This low friction allows a skater to easily glide over the ice surface, and in addition, the physical properties of the ice allows a skater to dig in with his skate in order to go around a turn, speedup,orstop.
- A skater propels himself forward by pushing off the ice with a force perpendicular to the skate blade. Since the friction of the blade with the ice is almost zero, this is the only way he can propel himself forward.

- As the skater pushes off with his rear leg, a perpendicular force *F* is exerted on the skate by the ice. The component of the force *F* that points forward (in the direction of motion) is what pushes the skater forward.
- At the same time, his other skate is either raised or gliding on the ice. As the skater moves forward he then switches to the other leg and pushes off the ice with that one, and the process is mirrored.
- To push off the ice with greater forward force (and accelerate faster), the skater increases the angle *α*, which increases the component of force in the direction of motion.
- A skater can also skate backwards using a gliding pattern in the shape of a lazy "S" (as shown below), in which the skater's blades never leave the ice.
- However, the skater cannot push off against the ice as hard as he does when skating forward, which means he cannot go as fast. In this technique, the skater pushes against the ice with his push-skate facing inward, while his other skate glides.
- As the skater moves backwards he then switches to the other leg and pushes off the ice with that one, and the process is mirrored. Thus, the physics of skating backward is similar to the physics of skating forward.
- A skater can at most move his feet at about 7 m/s, and the greatest forward push force will be when he begins skating from rest.
- At this point the velocity of his foot relative to the ice is 7 m/s. As the skater gains speed this relative velocity changes.
- For example, if the skater reaches a speed of 5 m/s, the relative velocity of his foot relative to the ice is 2 m/s (assuming he moves his leg backwards, with no sideways component of velocity), and the push force is less as a result.
- Consequently, there is a maximum speed a skater can reach, which is directly influenced by how fast he can move his feet on the ice. However, the maximum speed the skater can reach is not necessarily 7 m/s.
- It can be much more than this if the skater, when pushing off the ice, moves his leg backward with a sideways component of velocity. Analyzing the physics to determine the maximum speed involves looking at the biomechanics of the skater.

- To maintain his balance when accelerating forward, a skater will crouch (or bend) forward in the direction of motion. This prevents him from falling (tipping) backwards due to the torque caused by the forward component of the force *F*.
- By crouching (or bending) forward, the skater is moving his center of mass forward which creates a counter-torque. This counter-torque balances the torque caused by the forward component of *F*, and this prevents him from falling (tipping) backwards.
- Crouching (or bending) forward also reduces a skater's air resistance (drag) by reducing his frontal area. This allows him to accelerate to, and maintain, a greater speed.
- In the sport of speed skating, skaters use clapskates which hinge at the front.
- This allows the blade to remain flattened against the ice when the skater raises his heel, during the stride. This increases the time that the skater can push off against the ice, which enables him to accelerate faster. This design trick is of a more practical nature, while still being related to the physics of ice skating.
- In some ice skating sports, such as figure skating, movement on the ice involves artistry and technical skill, while in other sports, such as speed skating, movement on the ice is of a strictly technical nature and is geared towards speed and efficiency.
- When the skater is on the straight part of the track his strides are wider than on the curved part of the track.
- The reason for this is because on the curved part of the track it is easier to steer around the turn if he keeps the lateral distance between his strides small.
- However, the trade-off is that he can't go as fast around a turn as he can on the straightpart of the track. In short track speed skating the radius of the turn is quite small. This introduces considerable "lean" of the skaters towards the inside of the turn, due to centripetal acceleration.

- As a result, short track speed skaters can't do much skating around the turns. They must mostly ride the turns using the momentum gained on the straight part of the track.

7. **Write a brief note on characteristics of different types of biomaterial or Describe about the characteristics of biomaterials.**

Biomaterials

Biomaterials are used to make devices to replace a part or a function of the body in safe, reliably economically, and physiologically acceptable manner. A variety of devices and materials are used in the treatment of disease or injury. Commonplace examples include suture needles, plates, teeth fillings, etc.

The different types of biomaterial and the characteristics are the following;

Polymers

There are a large number of polymeric materials that have been used as implants or part of implant systems. The polymeric systems include acrylics, polyamides, polyesters, polyethylene, polysiloxanes, polyurethane, and a number of reprocessed biological materials.

Some of the applications include the use of membranes of ethylene-vinyl-acetate (EVA) copolymer for controlled release and the use of poly-glycolic acid for use as a resorbable suture material. Some other typical biomedical polymeric materials applications include: artificial heart, kidney, liver, pancreas, bladder, bone cement, catheters, contact lenses, cornea and eye-lens replacements, external and internal ear repairs, heart valves, cardiac assist devices, implantable pumps, joint replacements, pacemaker, encapsulations, soft-tissue replacement, artificial blood vessels, artificial skin, and sutures.

As bioengineers search for designs of ever increasing capabilities to meet the needs of medical practice, polymeric materials alone and in combination with metals and ceramics are becoming increasingly incorporated into devices used in the body.

Metals

The metallic systems most frequently used in the body are:

(a) Iron-base alloys of the 316L stainless steel
(b) Titanium and titanium-base alloys, such as (i)Ti-6% Al-4%V, and commercially pure ³ 98.9% (ii) Ti-Ni (55% Ni and 45% Ti)
(c) Cobalt base alloys of four types (i) Cr (27-30%), Mo (5-7%), Ni (2-5%) (ii) Cr (19-21%), Ni (9-11%), W (14-16%) (iii) Cr (18-22%), Fe (4-6%), Ni (15-25%), W (3-4%) (iv)Cr (19-20%), Mo (9-10%), Ni (33-37%)

The most commonly used implant metals are the 316L stainless steels, Ti-6%-4%V, and Cobalt base alloys of type "i" and "ii". Other metal systems being investigated include Cobalt-base alloys of type "iii" and "iv", and Niobium and shape memory alloys, of which (Ti 45% - 55%Ni) is receiving most attention. Further details of metallic biomedical materials will be given later.

Composite Materials

Composite materials have been extensively used in dentistry and prosthesis designers are now incorporating these materials into other applications. Typically, a matrix of ultrahigh-molecular-weight polyethylene (UHMWPE) is reinforced with carbon fibers. These carbon fibers are made by pyrolizing acrylic fibers to obtain oriented graphitic structure of high tensile strength and high modulus of elasticity.

The carbon fibers are 615mm in diameter, and they are randomly oriented in the matrix. In order for the high modulus property of the reinforcing fibers to strengthen the matrix, a sufficient interfacial bond between the fiber and matrix must be achieved during the manufacturing process. This fiber reinforced composite can then be used to make a variety of implants such as intramedullary rods and artificial joints. Since the mechanical properties of these composites with the proportion of carbon fibers in the composites, it is possible to modify the material design flexibility to suit the ultimate design of prostheses.

Composites have unique properties and are usually stronger than any of the single materials from which they are made. Workers in this field have taken advantages of this fact and applied it to some difficult problems where tissue in-growth is necessary. Examples:

Deposited Al_2O_3 onto carbon;

Carbon / PTFE;

Al_2O_3 / PTFE;

PLA-coated carbon fibers.

Ceramics

The most frequently used ceramic implant materials include aluminum oxides, calcium phosphates, and apatites and graphite. Glasses have also been developed for medical applications. The use of ceramics was motivated by:

(i) Their inertness in the body,
(ii) Their formability into a variety of shapes and porosities,
(iii) Their high compressive strength, and
(iv) Some cases their excellent wear characteristics.

Selected Applications of ceramics include:

(a) Hip prostheses,
(b) Artificial knees,
(c) Bone grafts,
(d) A variety of tissues in growth related applications in orthopedics

dentistry and heart valves.

Applications of ceramics are in some cases limited by their generally poor mechanical properties: (a) in tension; (b) load bearing, implant devices that are to be subjected to significant tensile stresses must be designed and manufactured with great care if ceramics are to be safely used.

Biodegradable Materials

Another class of materials that is receiving increased attention is biodegradable materials. Generally, when a material degrades in the body its properties change from their original values leading to altered and less desirable performance. It is possible, however, to design into an implant's performance the controlled degradation of a material, such that natural tissue replaces the prosthesis and its function.

8. Explain biomechanical analysis of physiological system Explain the mechanics of (i) Shoulder (ii) Ankle (A/M 2017)

BIOMECHANICAL ANALYSIS OF HIP

The Hip

The hip joint has a unique importance in the human body. It is one of the largest and most stable joints. As a consequence of bipedal walking, from the mechanical point of view, it becomes the most exposed place, where the carrying free limb is connected with the solid pelvic girdle on which the spine system is connected.

A right bipedal system is observed only in humans and in birds. The difference is in the location of a central point. Birds (and in the past dinosaurs too) have the central point below the hip joint level, so the bird body resembles a hanged pendulum. This system is very stable. It demands a minimal force for keeping an erect position, which was very important for dinosaurs with their enormous weight.

The hip joint has an intrinsic stability provided by its relatively rigid ball and-socket configuration. It has also a great deal of mobility, which allows normal locomotion in the performance of daily activities. For this posture the whole human skeleton is adapted, but especially the skeleton of the pelvis, which is flat with an open pelvis muscle apparatus.

Moreover, the hip joint is basically a spherically shaped joint with bounded dynamic mobility, with the shape of a rotational ellipsoid, whose lengths of axes are not that different. Joint cartilage covering the femur head and the acetabulum is very flexible hyaline cartilage, which because of its deformations allows movements corresponding to the ball-shaped joint, which is its main function from a biomechanical point of view. The femoral head is covered with hyaline cartilage from 2.2 to 3.7 mm thickness.

The hip joint has a unique status in the body because it connects a limb carrying the whole body weight with a relatively non movable pelvic round. It is substantially different, for example, from the shoulder joint, inside which the humerus is connected with a trunk by an inter link—scapula and the heterogeneous muscles.

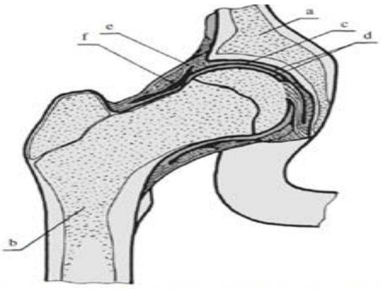
To determine the value of pressure force functioning on the femur head we will assume that individual skeleton segments are perfectly solid and mutually connected bodies. Let point *C* be the center of the femoral head. Then from a moment balance condition to this point we determine a reactive force *Fs* as a resultant of muscle forces keeping the balance. We have,

$$F_s = G' \frac{a}{b} \quad [\mathrm{N}]$$

where *G* is the weight of the body *G* lowered by the weight of the supporting lower limb, *a* is an arm of the acting weight force *G* with respect to the point *C*, and *b* is an arm of the acting force *Fs* with respect to the point *C.*

It means that when standing on one leg a force acts on the femur head that is 3.7 times greater than the weight of the human body. This force acts on the femur head in a direction diverted medially from 16◦ from the vertical. The resultant muscle forces *Fs* and the resulting pressure force *P* in this static situation act in one plane. This plane goes in a direction of an acting pressure force of the human body weight and the center of rotation of the femur head.

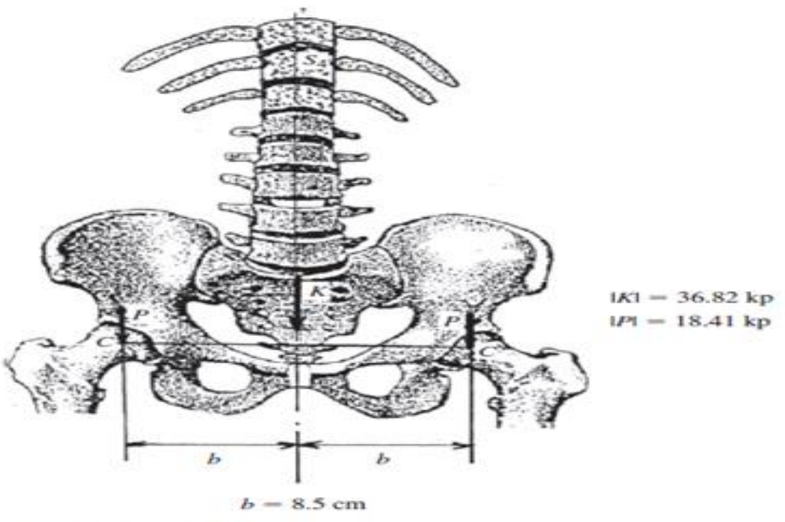
The hip joint is stabilized by the muscle system, which is important for biomechanical relationships in a joint, because it affects force relationships in a limb and its movements. Movements in the hip joint are controlled by muscles, which do not lay directly on the hip joint. Where muscles grip and the direction of their operation are very important from a biomechanical point of views. The below figure shows the schematic cross section of the hip joint.



Schematic cross section of the hip joint [modified after Beznoska et al. (1987) and Nedoma et al. (2006)], where a = the pelvis, b = the femur, c = the contact surface, d = the cartilages, e = the joint capsule, and f = the joint cavity.

Two basic trajectory systems—tension and pressure, as it follows from experimental and theoretical biomechanics—are a morphological substrate of tensile and compression forces during loading of the end of the upper femur. Biomechanically less active, less exposed places do not have such a structure, and they are labeled like the so-called Ward triangle, seen on the frontal cut of the end of the upper femur. These anatomical experiences and basic knowledge are necessary to the mathematical simulation function of the hip joint and during the mathematical simulation as well as design of total replacement, but also during its construction and own implantation of an artificial joint (THA). The basic ordinary model is the static Fischer hip joint model.

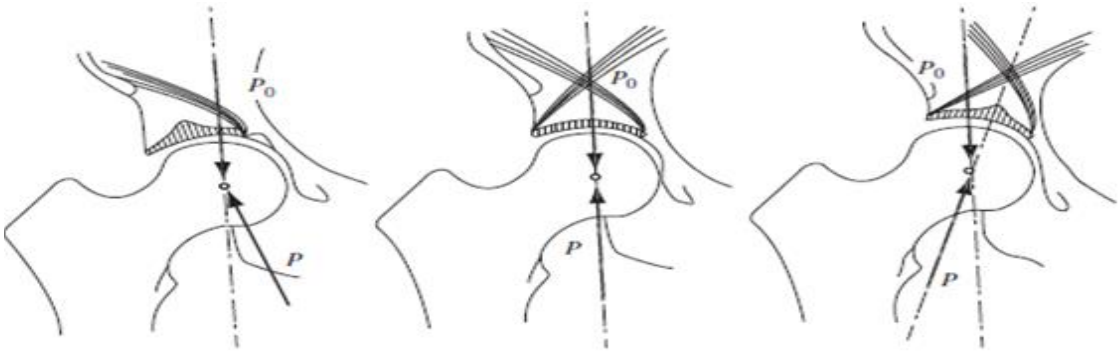
Fischer hip joint model:



The Fischer model.

The center of gravity is situated in the middle sagittal plane at the Th 10–11 disk area. Both joints carry the weight of the body, which is distributed between both limbs and the median halves vertically cut the line connecting the centers of both hip joints. The head is in contact with the acetabular surface and we refer to it as the load zone. The resultant of post-operating forces acts on the head of the hip joint in a vertical direction.

Bombeli model:



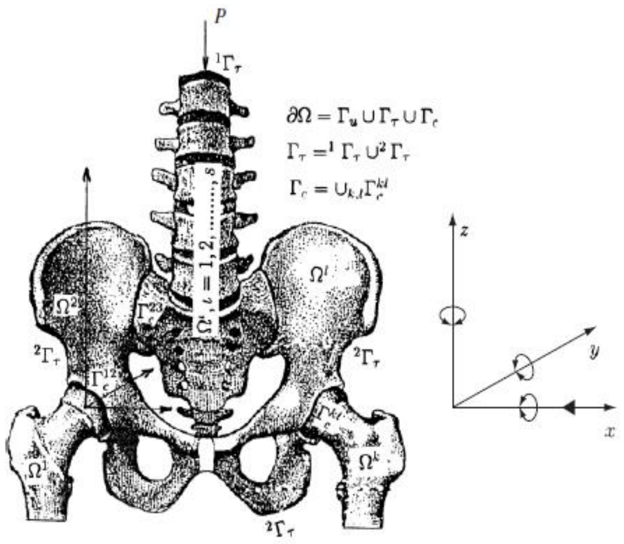
The Bombelli model.

Bombelli (1983) showed that the forces acting on the hip joint were vertical, going through the center of the loading surface, the so-called weight-bearing surface (WBS). Under normal anatomical relations this force goes through the top of the Gothic arch. During the inclination of the loading plane from a horizontal line, damage to the biomechanical balance is observed.

Nedoma Model

During the years 1986–1989 Nedoma formulated the problem of the distribution of stresses in the hip joint based on the theory of contact problems in linear elasticity and thermoelasticity, and on this theory he presented the fundamental biomechanical model of the hip joint.

This model allows one to study biomechanical relations in the hip joint as a system of viscoelastic bodies, which are in mutual contact. If this model is investigated as a three dimensional model, then it describes the hip joint and its biomechanical behavior very exactly. The exactness of this model depends on the exactness of the determination of the geometry of the investigated skeleton part and a location and a size of acting muscular forces. Contemporary computed tomography (CT) and magnetic resonance imaging (MRI) techniques allow one to obtain a real geometry of the investigated part of the skeleton with sufficiently high accuracy.



The Nedoma model.

➢ BIOMECHANICAL ANALYSIS OF KNEE

The Knee

- The knee joint is created by the contact of three bones: femur, tibia, and patella.
- The joint contact surfaces on the femur condyles are egg shaped, whereas the medial condyle is more convex than the lateral one.
- On the front side between the femur condyles there is a groove-shaped area leading to the patella.
- The joint contact surfaces on the tibia condyles are slightly concave and do not correspond to curved surfaces of the femur condyles.
- This lack of harmony is balanced by ligament cartilages— menisci that are situated medially and laterally.
- The knee joint is basically a combination of the trochlear-shaped and wheel-shaped connections.

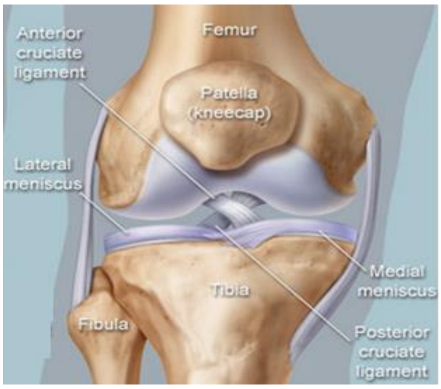
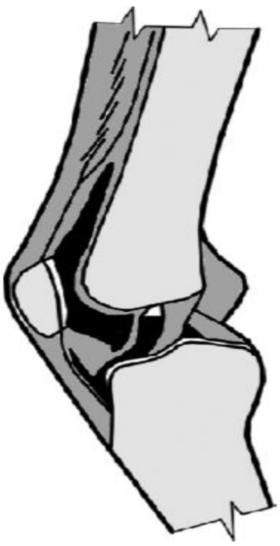
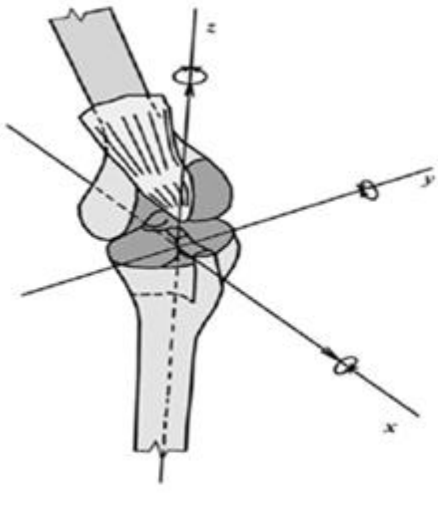


Diagram of the knee

In comparison with the hip joint the mobility of the knee joint is much more complicated. Nonconcentric curvatures of the femur condyles and shallow hollows of the tibia condyles together with a functional influence of the ligament apparatus give the knee joint from 5◦ to 6◦ of freedom. There are 2◦ of freedom in rotation and 2◦ of freedom in displacement .

The elastic connection and stability of the femoral and tibial joint parts are established mainly by the cruciate ligaments, which are situated among condyles a nd collateral ligaments (medial and lateral).





Degree of freedom in the knee joint

The rotary motion along the *y* axis is created during normal knee bending. But the center of bending is moved along the surfaces that form a circle of 20–25mm in diameter. It is established by the curvature of the femoral condyles.

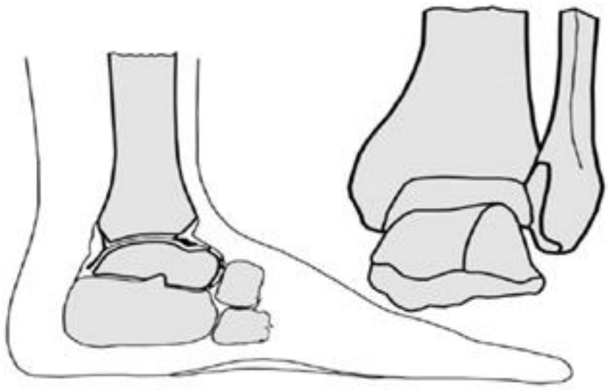
Rotation along the *z* axis occurs when the knee is influenced by different femur condyle shapes, mainly before reflexion ends. Limited rotation is allowed also along the *x* axis. Displacement between the femur condyle and the tibia occurs along *x* and *y* axes and during flexion along the *y* axis. It is also established by the condyle femur shape. The curvature of contact surfaces of the femur condyle allow the knee joint to flex up to 160◦. On the longitudinal – Sagittal plane the femur condyle has the shape of a spiral. The curvature gets smaller toward the back. The result of this shaping is the continual transfer of the center of rotation, which is dependent on the angle of the knee bending. As a consequence of this shaping of femur condyles and the difference between the size of the medial and lateral condyles, there is a combination of their generating and sliding along surfaces of the tibia condyles during joint flexion. This finding is important in determining the reasons for the degenerative changes in the knee joint contact surfaces.

- **BIOMECHANICAL ANALYSIS OF ANKLE(A/M 2017)**

The Ankle

The function of the lower limb is adapted to a supporting function. This is why the skeleton of the leg and the foot are fundamentally different types of the limb construction, dependent on posture and gait. The ankle consists of three bones forming the ankle mortise. This joint system consists of the tibiotalar, the fibullotalar, and the tibiofibullar joints. The ankle joint is of a hinge-type joint. Its stability depends on joint congruency and the medial, lateral, and syndesmotic ligaments. Ankle stability increases and depends more on the articular surface congruency during weight-bearing activities.

The skeleton of the foot is similar to that of the hand. It consists of 28 bones, of which 7 bones are ossa tarsi (talus, os naviculare, os cuneiforme I–III, that is, mediale, intermedium, laterale, calcareus, and os cuboideum), 5 ossa metatarsalia, and digiti pedis—created by the phalanges.



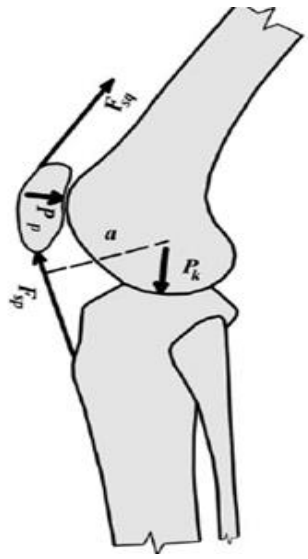
schematic image of ankle joint

- The greatest mobility is in the ankle joint. The motion of the foot occurs around three axes (i.e., abduction–adduction, dorsiflexion–plantarflexion, and eversion– inversion) and on three planes.
- Flexion–extension acts in the sagittal plane, abduction– adduction acts in the horizontal or tranverse plane, and inversion–eversion acts in the coronal or frontal plane.
- The function of the ankle joint is rendered possible because the joint surfaces are kept in contact by the lateral ligaments.
- The ankle joint is a special kind of the trochlea-shaped joint with screw-shaped surfaces (helicoid surfaces and helical surfaces).

➢ **BIOMECHANICAL ANALYSIS OF SHOULDER(A/M 2017)**

The Shoulder Joint

- The shoulder connects the upper limb to the trunk, it functions together with the elbow to position the hand in space, and allows for its effective function.
- The shoulder consists of the glenohumeral, acronioclavicular, sternoclavicular, and scapulothoracic joints and the musculature structures that support these joints.



Force relation in the knee joint: p_k =a compression force.F_{sp} = a tensile force , F_{sq}= a quadriceps force, P_p = a force acting in the patella-femoral connection, and a=an arm of the patellar tendom force.

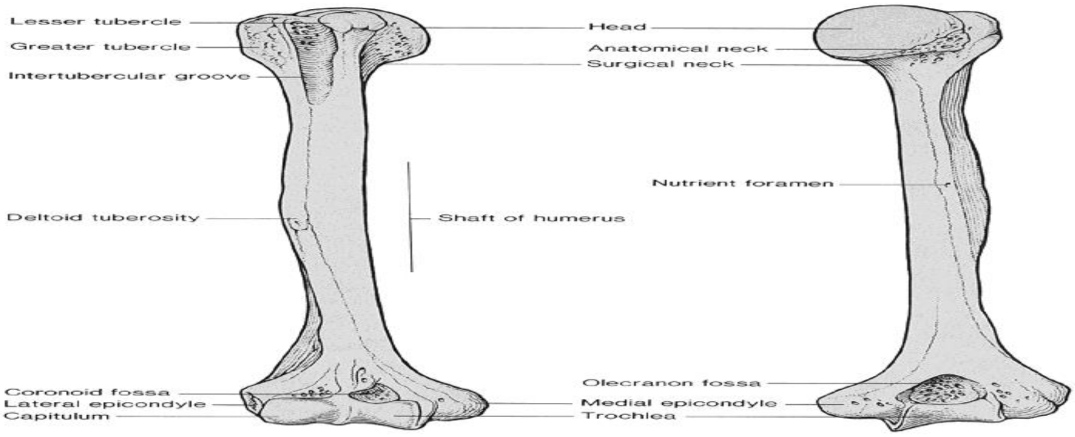
Shoulder motions include flection and extension, abduction, and internal-external rotation . The joint pit is situated on a shoulder scapula and the head is situated on the humerus. The contact joint surfaces are covered by cartilage. On the periphery the joint pit is enlarged about the cartilage border. According to the shape of the contact surface, the shoulder joint is a balland- socket joint .It has from 4◦ to 6◦ degrees of freedom, that is, three in the rotation around the *x*, *y*, and *z* axes and one during the shifting process, which occurs only in luxation. Theoretically, the forward elevation is possible at about 180◦, the average value in men is 167◦ and in women it is 171◦ extension or posterior elevation averages.

These values are limited by capsular torsion. Abduction in the coronal plane is limited by bony tuberosity on the acromion. Forward elevation in the scapula plane is more functional because in this plane the inferior portion of the capsule is not twisted and the musculature of the shoulder is optimally aligned for elevation of the arm. The synovial fluid acts by cohesion and adhesion to stabilize the glenohumeral joint, and it adheres to the articular cartilage overlying the glenoid and proximal humerus. Numerous muscles act on various components in the shoulder area to provide mobility and dynamic stability. To understand muscle functions and force acting, one must consider a given orientation, size, and activity of the acting muscles.

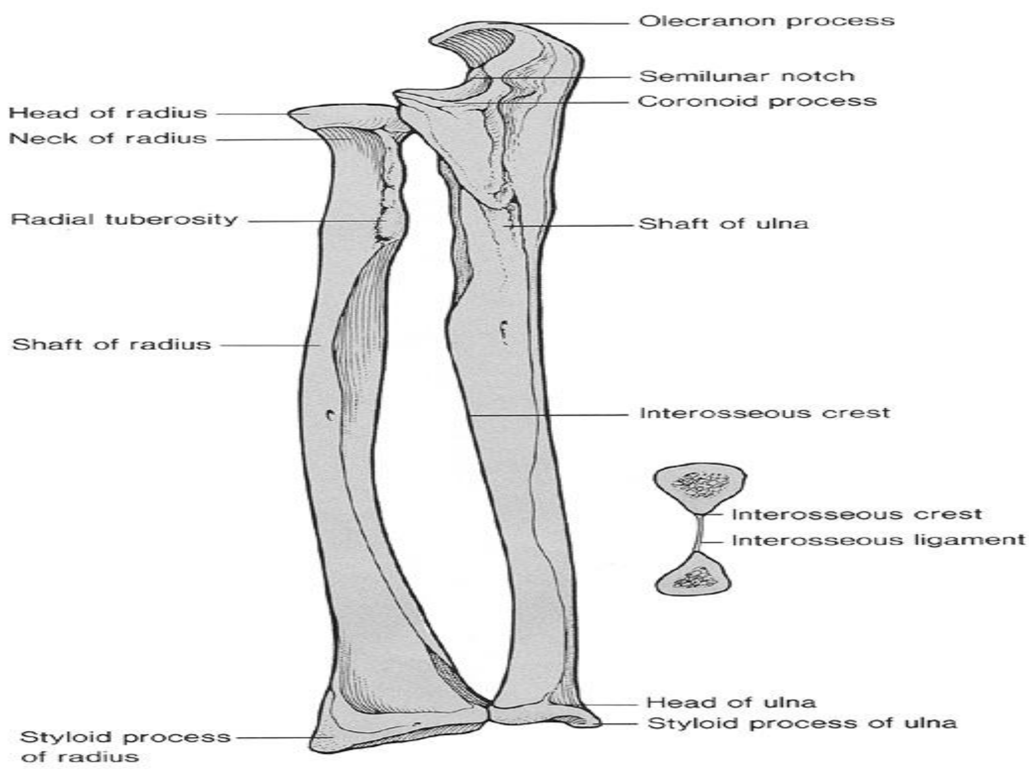
BIOMECHANICAL ANALYSIS OF ELBOW

Elbow:

- The elbow complex is a compound synovial joint which contains three articulations within a Common fibrous capsule.
- The trochlea of the humerus with the semilunar notch of the ulna form the hemerounular joint, the capticulam of the humerus and the head of the radius form the humeroradial joint, and the head of the radius in the radial notch of the ulna form the superior radioulnar joint.
- The humeroulnar joint is a hinge type joint which allows for flexion and extension of the forearm.
- The head of the radius is also free to rotate on the capitulum of the humerus and does so during pronation and supination. (Pronation- inward movement, supination- outward movement).



The radioulnar articulation is a pivot type joint in which the head of the radius spins in the radial notch when the forearm is rotated. The bony interlocking of the ulna with the humerus provides much stability to the elbow joint. Three ligaments are also major contributors to this joints stability.



The flexion- extension axis of the elbow joint passes through the center of the trochlea of the humerus and changes only slightly during complete flextion-extension.

The axis of rotation of the forearm for supination or pronation is through the head of the radius and distal ulnar head. The radius rotates around the relatively fixed ulna during pronation –supination. The upper arm and the forearm form an angle in the valgus-varus (angulation (or bowing) within the shaft of a bone or at a joint) direction known as the carrying angle.

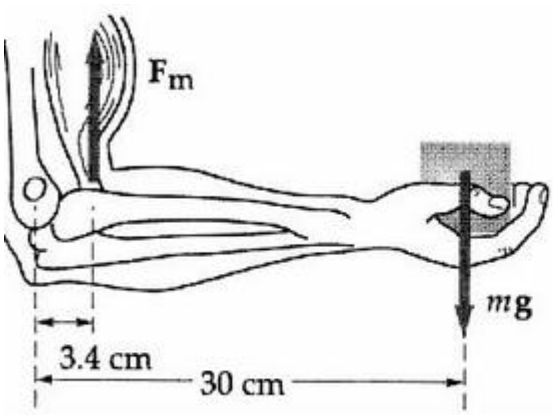


Diagram shows the force action in elbow.

An extensive examination of the musculature of the upper limb predicted maximum forces up to 3 kN during flexion on both the humero-radial and humero-ulnar articulations. A biomechanical analysis of elbow joint function was carried out investigating eating, dressing, pulling a heavy object and assisted standing from the sitting position. They found compressive load of 300 N acting on both sides of the trochlear notch during the dressing and eating activities, and that the total joint load reaches peaks up to 2500N in the seat rise and table pull activities.

➢ BIOMECHANICAL ANALYSIS OFSPINAL COLUMN

The spinal column

The vertebral column is a mechanical marvel in that it must afford both rigidity and flexibility. The segmental design of the vertebral column allows adequate motion among the head, trunk, and pelvis; affords protection of the spinal cord; transfers weight forces and bending moments of the upper body to the pelvis; offers a shock absorbing apparatus; and serves as a pivot for the head. Without stabilization from the spine, the head and upper limbs could not move evenly, smoothly, or support the loads imposed upon them.

Weight Distribution

The flexible vertebral column is balanced upon its base, the sacrum. In the erect position, weight is transferred across the sacroiliac joints to the ilia, then to the hips, and then to the lower extremities. In the sitting position, weight is transferred from the sacroiliac joints to the ilia, and then to the ischial tuberosities.

Spinal Length

About 75% of spinal length is contributed by the vertebral bodies, while 25% of its length is composed of disc material. The contribution by the discs, however, is not spread evenly throughout the spine. About 20% of cervical and thoracic length is from disc height, while approximately 30% of lumbar length is from disc height. In all regions, the contribution by the discs diminishes with age.

The Spinal Curves

A curved column has increased resistance to compression forces. This is just as true in the spine as for a rib or long bone. The spinal curves offer the vertebral column increased flexibility and shock-absorbing capability while still maintaining an adequate degree of stiffness and stability between vertebral segments.

The six degrees of freedom involved in spinal kinematics.

Action and brake mechanisms

Flexion.

During flexion, the intervertebral discs (IVD's) tends to compress at its anterior aspect, the inferior set of articular facets glide anterosuperiorly upon the mating set of superior facets of the vertebra below, and the range of motion is checked by the posterior anulus of the disc, the posterior longitudinal ligament, the intertransverse ligaments, the supraspinous ligament, the extensor muscles, and the dorsolumbar aponeurotic sheet of fascia.

Extension.

Extension has a much lower magnitude than flexion. The IVD tends to compress at its posterior aspect, and the inferior set of articular facets glide posteroinferiorly upon the mating superior facets below. The motion is checked by the anterior anulus of the disc, the anterior longitudinal ligament, all the anterior and lateral muscles that contribute to flexion, the anterior fascia and visceral attachments, and probably spinous process and/or laminae jamming at maximum extension.

Rotation

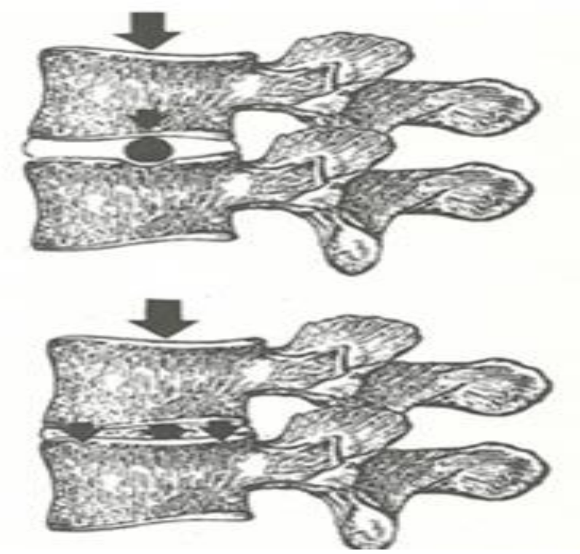
Spinal rotation is limited by the planes of the articular facets, the thickness of the associated IVD's, and the resistance offered by the fibers of the disc's anulus and the vertebral ligaments under torsion.

Lateral Bending

Sideward abduction involves a degree of tilting of vertebral bodies on their discs. The anterior aspect of the vertebral bodies in the upper spine also rotate toward the side of convexity, the posterior aspect swings in the opposite direction, and the facets tend to slide open on the convex side and override on the concave side. The motion is checked by the intertransverse ligaments and intercostal tissues on the convex side, behind the fulcrum, and the apposition of ribs on the concave side

in the thoracic region.

The Vertebral Motion Unit



In figure: The vertical body failure from excessive compression of a healthy disc when the greatest stress is central(Top). The failure compression of a degenerative disc(Bottom).

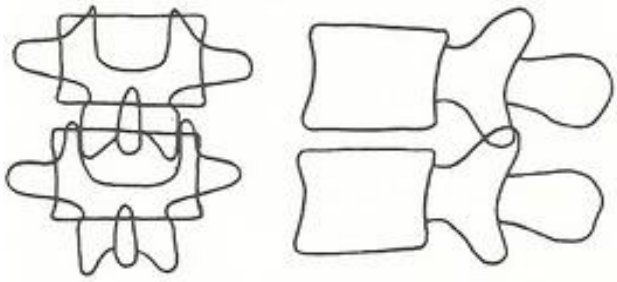
As it serves as the axial support of the trunk, the erect spinal column is a primary concern in static postural equilibrium. Since the body is never actually in a static state in life but exists in a state of "quiet dynamics" in the static postural attitude and a state of "active dynamics" in movement, the kinetic aspects of normal spinal biomechanics are an important consideration. Total spinal function is the sum of its individual component units.

An intervertebral "motion unit" consists of two vertebrae and their contiguous structures forming a set of articulations at one intervertebral level, thus conferring a quality and quantity of motion to the relationship of two vertebrae These units are firmly interconnected by the IVD and restraining ligaments and are activated by muscles that respond to both sensory and motor innervation.

The biomechanical efficiency of any one of the 25 vertebral motion units from atlas to sacrum can be described as that condition (individually and collectively) in which each gravitationally dependent segment above is (1) free to seek its normal resting position in relation to its supporting structure below, (2) free to move efficiently through its normal ranges of motion, and (3) free to return to its normal resting position after movement.

Spinal Movements

The important movers are shown in figure



Left, schematic posterior view of the intervertebral motion unit in the normal position; *right*, lateral view of the intervertebral motion unit in the normal position.

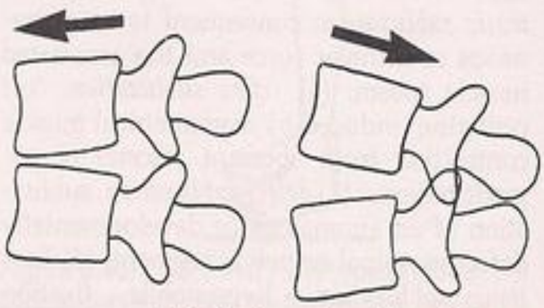
SPINAL EXTENSION

The spinal extensors span the entire length of the vertebral column, originating from the laminae, transverse processes, and ribs as diagonal strips and inserting as multiple tendon inserts on the spinous processes. This group of muscles, known collectively as the erector spinae, has the sole function of restoring the flexed spine to neutral and controlling flexion momentum as an antagonist guard.

Bilateral action by the splenius capitis, cervicis, rotatores, interspinales, and multifidi is also involved in spinal extension. The erector bundles are subdivided by innumerable connective tissue planes, and the entire group is enveloped by strong fascia in the lumbar region that is strongly anchored to the transverse processes. This tends to spread a mechanical load over a large area. As the erector spinae are the only muscles in the body supplied solely by the posterior rami of the spinal nerves, local pain, splinting, or unilateral weakness of this muscle points to spinal nerve involvement.

SPINAL FLEXION

The flexor muscles, developed most in the cervical and lumbar regions, are represented chiefly by the anterior longus coli and scalene muscles in the cervical area and the more lateral sternomastoid muscles, which are powerful assistants. In the lumbar region, the psoas major flexes the trunk on the thigh at the hip but has little effect on flexion of individual lumbar segments.



Left, flexion malposition; *right*, extension malposition.

9. Explain about the materials and designs involved in making total hip replacement and elaborate its fixing procedures? (A/M 2017) or Explain fixation of implants.

The fixation of orthopedic implants has been one of the most difficult and challenging problems. The fixation of implant mainly by

- Internal fixation
- External fixation.

INTERNAL FIXATION:

A broken bone must be carefully stabilized and supported until it is strong enough to handle the body's weight and movement. Until the last century, physicians relied on casts and splints to support and stabilize the bone from outside the body. The advent of sterile surgical procedures reduced the risk of infection, allowing doctors to internally set and stabilize fractured bones.

During a surgical procedure to set a fracture, the bone fragments are first repositioned (reduced) into their normal alignment. They are held together with special implants, such as plates, screws, nails and wires.

Internal fixation enables patients to return to function earlier, and reduces the incidence of nonunion (improper healing) and malunion (healing in improper position) of broken bones.

The implants used for internal fixation are made from stainless steel and titanium, which are durable and strong. If a joint is to be replaced, rather than fixed, these implants can also be made of cobalt and chrome. Implants are compatible with the body and rarely cause an allergic reaction.

Plates

Plates are like internal splints that hold the broken pieces of bone together. They are attached to the bone with screws. Plates may be left in place after healing is complete, or they may be removed (in select cases).

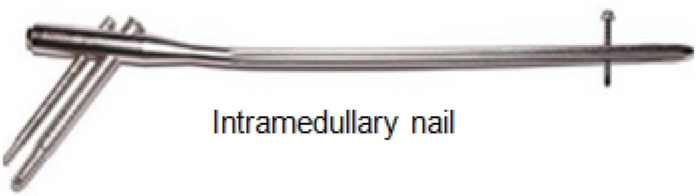
Screws

Screws are used for internal fixation more often other type of implant. Although the screw is a simple device, there are different designs based on the type of fracture and how the screw will be used. Screws come in different sizes for use with bones of different sizes. Screws can be used alone to hold a fracture, as well as with plates, rods, or nails. After the bone heals, screws may be either left in place or removed.

Intramedullary Nail

An intramedullary rod, also known as an intramedullary nail (IM nail) or inter-locking nail is a metal rod forced into the medullary cavity of a bone. IM nails have long been used to treat fractures of long bones of the body. IM nails resulted in earlier return to activity for the soldiers, sometimes even within a span of a few weeks, since they share the load with the bone, rather than entirely supporting the bone

The intramedullary Nail System consists of metallic implants including interlocking intramedullary nails, interlocking fusion nails, and nail caps. Intramedullary nails contain holes proximally and distally to accept locking screws. Intramedullary Interlocking Nails are provided with a variety of screw placement options based on surgical approach, nail type and indications.

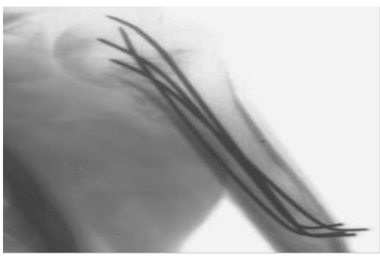


Xray showing the proximal portion of a fractured tibia with an intra medullary nail.

Wires/Pins

Wires are often used to pin the bones back together. They are often used to hold together pieces of bone that are too small to be fixed with screws. In many cases, they are used in conjunction with other forms of internal fixation, but they can be used alone to treat fractures of small bones, such as those found in the hand or foot. Wires are usually removed after a certain amount of time, but may be left in permanently for some fractures.

Example**: Kirschner wires** or **K-wires** or **pins** are sterilized, sharpened, smooth stainless steel pins. The wires are now widely used in orthopedics and other types of medical and veterinary surgery. They come in different sizes and are used to hold bone fragments together (pin fixation) or to provide an anchor for skeletal traction.



Intraoperative X-Ray of a Humerus fixated by Kirschner wires

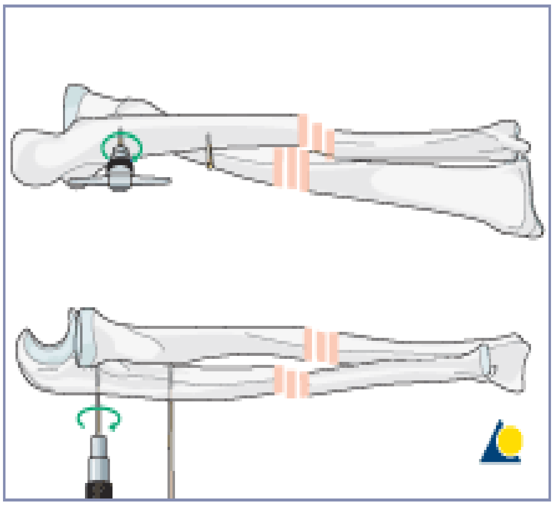
EXTERNAL FIXATION:

An external fixator acts as a stabilizing frame to hold the broken bones in proper position. In an external fixator, metal pins or screws are placed into the bone through small incisions into the skin and muscle. The pins and screws are attached to a bar outside the skin. Because pins are inserted into bone, external fixators differ from casts and splints which rely solely on external support.

In many cases, external fixation is used as a temporary treatment for fractures. Because they are easily applied, external fixators are often put on when a patient has multiple injuries and is not yet ready for a longer surgery to fix the fracture. An external fixator provides good, temporary stability until the patient is healthy enough for the final surgery.

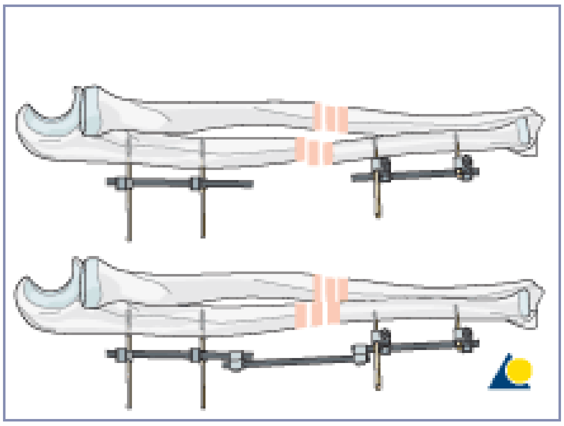
Ulnar threaded pins

Threaded pin insertion must always respect the anatomical safe zones. The ulna is subcutaneous along its posterior border and offers ease of access as a result.



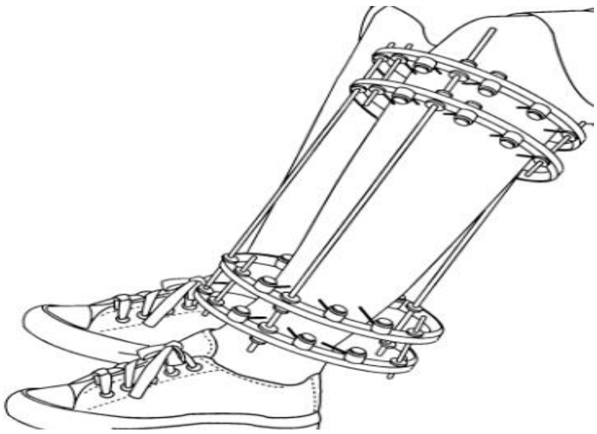
Bar-to-bar Clamp

Connect the threaded pins in each main fragment to a bar, using pin-to-bar clamps and tighten them. Connect the two bars loosely with an intermediate bar using bar-to-bar clamps.



Ilizarov external fixator

The Ilizarov external fixator is the name of a device that is used to lengthen bones. It is a very powerful tool that may also be used to stabilize fractures, regrow lost bone or correct deformities in the length rotation or angles of bones.



The device is attached to two ends of the bone during surgery. When the device is adjusted several times a day, tension pulls the ends of the bone slightly apart. The body's natural healing process fills in this space with new bone. The bone can be lengthened by about 1 mm each day. The device remains in place until the new bone becomes strong. This process may continue over several months.

Hydroxyapatite-coated schanz pins

- It is an external fixators used for distraction osteogenesis. (Lengthen the long bone of the body)

- Complications of external fixation include loosening of the fixation pins and pin-track infection.
- Coating pins with hydroxyapatite increases their fixation to bone and reduces the rate of infection and loosening during external fixation for distraction osteogenesis. Use of hydroxyapatite-coated pins should be considered in clinical situations requiring prolonged external fixation.
- Laboratory studies and clinical trials have suggested that hydroxyapatite coating improves the osteointegration of various orthopedic implants.

10. Explain the specifications for a prosthetic joint.

In recent years there has been a great deal of activity on the subject of prosthetic joint replacement, and many new implants have ensued. Commonly, these devices have been designed by their proponents to overcome specific problems but often the more general requirements of joint replacement have not been fully considered. Thus many prostheses have been devised without either a clear understanding of the specifications which must be met, or sufficient thought being given to the problems involved. This has naturally led to clinical problems with some replacements.

DESIGN CRITERIA

We consider that there are thirteen criteria which should be employed in designing prosthesis to meet the basic requirements. These are:

1. Appropriate articulation
2. Good stability
3. Adequate strength
4. Good fixation
5. Correct choice of materials
6. Low friction forces
7. Acceptable wear rate
8. Good salvage potential
9. Standardization

10. Sterilization
11. Cost effectiveness
12. Surgical instrumentation

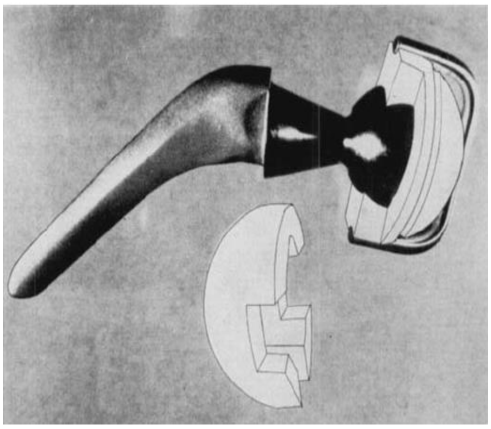
Appropriate articulation

Four factors are important in articulation.

- The degrees of freedom should be appropriate to the joint being replaced.
- The range of movement need not necessarily be as great as in the normal joint
- The components of the prosthesis should only articulate with each other, and not with bone or cartilaginous parts of the joint.
- The locus of the centre of rotation should closely approximate to that of the normal joint.

Stability

- The stability of the normal joint derives from a proper relationship between the shape of the articular surfaces, the muscles acting about the joint, and the length and position of ligaments.
- It is invariably erosion of the joint surfaces, loss of integrity of these ligaments and perhaps loss of muscle tone due to inactivity which cause instability of the diseased joints. In many cases the preservation of the ligament and reconstruction of the articular surfaces will restore adequate stability.



The Liverpool shoulder prosthesis which combines a full range of articulation with controlled inbuilt stability. A protective cap (shown below) protects the bearing surface of the metal scapular component and ensures correct orientation of the plastic humeral component during insertion. The plastic parts have been outlined for clarity of illustration.

Loading

It is clearly necessary that the implant be capable of supporting the loads to which it will be subjected within the body. Direct measurements of these loads are not usually possible. The technical and ethical difficulties involved with such tests, indirect load measurements may have to be used. Indirect measurements have been made of knee forces and similar work on other normal and prosthetic joints.

Fixation

Adequate fixation of the prosthesis is necessary for two reasons. Firstly, to prevent unwanted movements between the implant and bone, and secondly, to transmit joint Ioads into the bone in such a way that they will not cause excessive stresses in bone or implant. This requirement can be met by ensuring a good surface contact between prosthesis and bone over a sufficiently large surface area.

Materials

The material used in prosthesis must have adequate strength to carry the loads, be non-toxic, not corrode in the body nor produce the adverse tissue reaction.

The materials most commonly used are the four metals and three polymers listed below :

- Cobalt-chromium-molybdenum cast alloy
- Cobalt-chromium-tungsten wrought alloy
- Pure Titanium and titanium alloy
- Ultra high molecular weight polyethylene (UHMWP)
- Polymethylmethacrylate cement (known as acrylic bone cement).
- Polydimethylsiloxane (known as silicone rubber or Silastic)

- Stainless steel is especially useful during the development phase of new prostheses because it can be forged and is readily machined.
- Cobalt-chromium-molybdenum cast alloy, however, whilst being more difficult to machine Wrought cobalt-chromium-tungsten alloy may be used as an alternative to stainless steel.
- Titanium has the disadvantage of having poorer bearing properties

The mechanical properties are given below:

Material	Ultimate tensile strength (U.T.S.) $N/m^2 \times 10^{-7}$	Fatigue limit $N/m^2 \times 10^{-7}$	Modulus of elasticity $N/m^2 \times 10^{-10}$
Stainless steel 316 S16 & S17, 317 S16	65–100	28–30	20
Co – Cr – M cast alloy	69	30	20
Co – Cr – W wrought alloy	154	49	23
Titanium wrought	40–67	17–30	11
Polymethylmethacrylate	7	0.68	0.3
Polymethylsiloxane	0.5	......	up to 0.001
Ultra high molecular wt. polyethylene	2–4	......	0.055–0.07
Human bone (wet) a) cortical	6.5–11.7 (Compressive)	......	1.1–2.3
b) cancellous	0.03–1.59	......	0.003–0.1

Friction

- The co-efficient of friction of all these materials is an order of magnitude greater than that of articular cartilage.
- UHMWP against metal gave the lowest friction of the materials.
- The patient may not be aware of friction forces on his prosthesis, they have to be transmitted across the implant/bone interface and therefore should not be ignored, for high friction loads could endanger implant fixation.

Wear

- The best combination of currently available materials for joint replacement is considered to be cobalt-chromium molybdenum alloy or stainless steel in conjunction with ultra high molecular weight polyethylene.
- The combination caused minimal wear to the metal component.

- Wear or geometric changes in joint prostheses are usually confined to the plastic components and these should be replaceable, if bearing life is in doubt.

Salvage potential

It must be possible to salvage (safe) the joint in the event of failure. Clearly the smaller the amount of bone removed in the initial operation, the greater the choice of procedures during a second intervention. Cemented implants with long intramedullary stems can be particularly difficult to remove.

Standardization and cost effectiveness

To achieve universality the designer should aim at producing one size and configuration of prosthesis for either side of any patient, and if possible, a simple geometric shape which can be economically manufactured. For example, this has been achieved with the elbow by avoiding intramedullary stem fixation and by sacrificing the conjunct rotation of the natural joint. Universality has been implemented in different ways in other joint prostheses.

Sterilization

- The metal parts of prostheses may be steam autoclaved and thus easily sterilized.
- Heat sterilization is not satisfactory for UHMWP which has a melting point in the range **130-14 degree** C (close to steam autoclave temperature) and a heat

  distortion temperature of **70-80 degree C** at which stress relieving and dimensional changes can occur.
- There are various methods for sterilizing plastic components, but of these, gamma radiation of at least **2.5** M rad is widely used and this method is also suitable for sterilizing metallic components.

As this process must be carried out at specialist centers, implants which contain plastic components should be supplied pre-sterilized.

Instrumentation

Special purpose instrumentation may be necessary to ensure accuracy of insertion of the prosthesis. The need or otherwise of such instrumentation should be considered together with the intended operative procedure throughout the design phase of the prosthesis. Prostheses which do not require special instruments are at an advantage provided that the surgical procedure or their implanted function is not compromised.

However, in some circumstances elaborate instruments which ensure accuracy or serve to reduce the operative trauma are justified.

11. Explain analysis of a rigid body in equilibrium.

EQUILIBRIUM OF A RIGID BODY

A rigid body may be subjected to one of the force systems classified as follows:

(a) Concurrent force system: Collinear, plane or spatial
(b) Parallel force system: plane or spatial
(c) Coplanar force system: concurrent and non-concurrent, parallel and non-parallel
(d) Spatial force system: concurrent and non-concurrent, parallel and non-parallel.

(a) Concurrent Force Systems

The analysis of the static equilibrium of a rigid body under the action of concurrent forces is quite similar to that of a particle. Concurrent forces may be collinear, coplanar or spatial and the vector method or the algebraic method can be employed with advantage.

If there are two forces acting at a point in equilibrium, the forces must be collinear. If there are three forces acting at a point in equilibrium, the forces must lie in a plane, i.e., the force system must be coplanar. This follows from the fact that any two lines of forces acting at a point must constitute a plane and the third force cannot have a component normal to that plane; otherwise that unbalanced component would upset the equilibrium. If there are four or more forces acting at a point in equilibrium, these may be coplanar or spatial.

The condition of equilibrium for a rigid body under the action of concurrent forces

$$\mathbf{F}_1, \mathbf{F}_2, \mathbf{F}_3, \ldots$$

is that their resultant

$$\mathbf{R} = \Sigma\mathbf{F} = \mathbf{F}_1 + \mathbf{F}_2 + \mathbf{F}_3 + \ldots$$

must vanish, i.e.,

$$\mathbf{R} = \Sigma\mathbf{F} = 0$$

This vector equation for equilibrium stands for three scalar equations for a general concurrent force system:

$$R_x = \Sigma F_x = 0 = F_{1x} + F_{2x} + F_{3x} + \ldots$$
$$R_y = \Sigma F_y = 0 = F_{1y} + F_{2y} + F_{3y} + \ldots$$
$$R_z = \Sigma F_z = 0 = F_{1z} + F_{2z} + F_{3z} + \ldots$$

where x, y and z are the coordinate axes arbitrarily drawn through the point of concurrency as shown in Fig.

If, on the other hand, there is a rigid body subjected to a system of concurrent forces resulting in **R**, then an equilibrant force **E** given by

$$\mathbf{E} = -\mathbf{R}$$

must be applied passing through the point of concurrency in order to bring the body to equilibrium.

(b) Parallel Force System

If a rigid body is subjected to a parallel force system, the resultant of the forces may be a non-zero force or a zero force, unaccompanied or accompanied by a couple-moment. Equilibrium of the rigid body demands that the resultant force and resultant moment must vanish:

$$\Sigma \mathbf{F} = 0; \qquad \Sigma \mathbf{M} = 0$$

It may be understood that the moment about any arbitrary point O as shown in Fig. may be considered and should be equated to zero.

For a plane parallel-force system, the moment of the forces about any point in the plane of the forces must be perpendicular to the plane of the forces. It is interesting to note that moment summation about different points provide different equations, such as

$$\Sigma \mathbf{M}_1 = 0, \qquad \Sigma \mathbf{M}_2 = 0, \qquad \Sigma \mathbf{M}_3 = 0,$$

In fact, there can be only two of these equations mutually independent; one in its own right and the other in lieu of

$$\Sigma \mathbf{F} = 0$$

It follows that for a system of plane parallel forces acting on a body, the conditions of equilibrium may alternatively be stated as

$$\Sigma \mathbf{M}_1 = 0 \quad \text{and} \quad \Sigma \mathbf{M}_2 = 0$$

where 1 and 2 are two suitably chosen points.

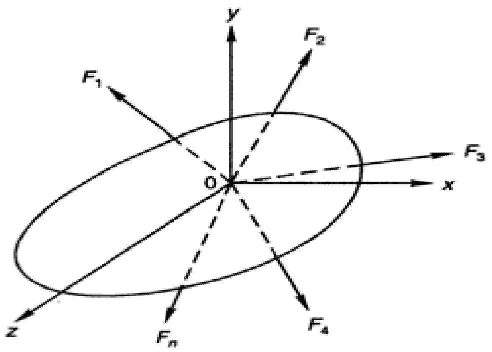
(c) Coplanar Force System

If the forces applied on a rigid body are such that their lines of action lie in the same plane and the moments due to the couples or otherwise are directed perpendicular to the plane, the body is said to be subjected to a coplanar force system. It is usual to choose the x-y plane in the plane of the coplanar force system and the x- and y-axes are chosen conveniently in regard to the directions of the forces as shown in Fig. The necessary conditions of equilibrium reduce to a set of three equations:

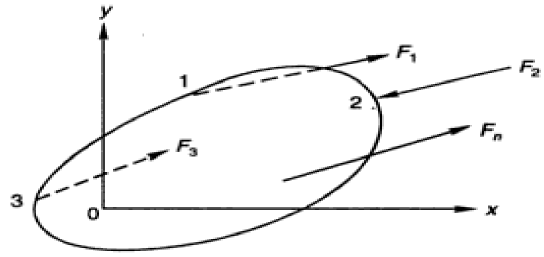
$$\Sigma F_x = 0 \quad \text{and} \quad \Sigma \mathbf{M} = 0$$

$$\Sigma F_y = 0$$

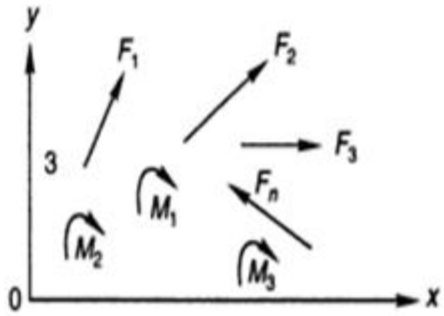
The moment summation referred to above is about any point in the plane of the forces; it is about the z-axis through the chosen point.



(a) Concurrent Forces



(b) Parallel Forces



(c) Plane Forces (in x-y Plane)

Different Force Systems Acting on a Rigid Body

An interesting and extremely useful point is to express the force summations of equilibrium in terms of equivalent moment summations. The advantage of doing so is that the moments can be taken about the line of action of a force which is unknown and needs to be eliminated at least temporarily. For example, if there are three unknown coplanar forces in a system, then moments about the line of action of each force, in turn, will yield three moment equations.

$$\Sigma M_1 = 0$$
$$\Sigma M_2 = 0$$
$$\Sigma M_3 = 0$$

It may as well be decided to consider the equivalent equations of equilibrium as

$$\Sigma M_1 = 0$$

$$\Sigma F_y = 0$$

$$\Sigma M_2 = 0$$

In each case, there are, in essence, two force equations and one moment equation; the apparent difference is only in the embodiment of these equations. It may so happen that the three equations set up in a particular case may not form an independent or a complete set, e.g., when a moment is taken about the point of intersection of two or more lines of forces. In such cases further equations for moment summations would provide an answer. It may also be noted that it is often advisable to set up a redundant equation to provide a check on the solution of the problem.

(d) Spatial Force System

The necessary conditions of equilibrium for a rigid body subjected to a general force system are those specified earlier by general equilibrium considerations

$$\Sigma \mathbf{F} = 0$$

$$\Sigma \mathbf{M} = 0$$

These two vector equations are equivalent to a set of six scalar equations:

$$\Sigma F_x = 0 \qquad \Sigma M_x = 0$$

$$\Sigma F_y = 0 \quad \text{and} \quad \Sigma M_y = 0$$

$$\Sigma F_z = 0 \qquad \Sigma M_z = 0$$

where the x, y and z axes are chosen arbitrarily but with due regard to convenience of handling the force system. For example, it may be preferable to have an axis in a direction in which a number of forces act and it may be advantageous to choose the x-y plane as the plane in which a number of forces lie.

EQUILIBRIUM OF A SYSTEM OF PARTICLES

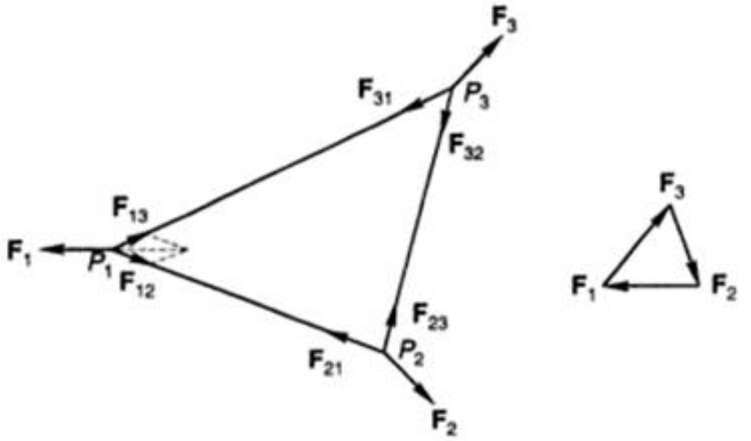
Consider a system consisting of three particles P_1, P_2 and P_3 as shown in Fig. The system is subjected to net external forces $\mathbf{F}_1$ at P_1, $\mathbf{F}_2$ at P_2 and $\mathbf{F}_3$ at P_3 as shown.

The resultant action on the system of particles consists of a single force $\mathbf{F}$ equal to the sum of the external forces.

$$\mathbf{F} = \mathbf{F}_1 + \mathbf{F}_2 + \mathbf{F}_3$$

The system will be in equilibrium if the resultant of the external forces vanishes, i.e.,

$$\mathbf{F} = \mathbf{F}_1 + \mathbf{F}_2 + \mathbf{F}_3 = 0 \quad \text{or } \mathbf{F}_1, \mathbf{F}_2 \text{ and } \mathbf{F}_3 \text{ constitute a closed triangle.}$$



A System of Three Particles

It is essential that each constituent particle of a system must also be in equilibrium. Take, for example, particle P_1. It is subjected to the external force $\mathbf{F}_1$ as well as the internal forces, $\mathbf{F}_{13}$ due to particle P_3 and $\mathbf{F}_{12}$ due to particle P_2. Obviously, $\mathbf{F}_1$, $\mathbf{F}_{12}$ and $\mathbf{F}_{13}$ must keep the particle in equilibrium.

Since action and reaction must be equal and opposite

$$\mathbf{F}_{12} = -\mathbf{F}_{21}; \ \mathbf{F}_{12} + \mathbf{F}_{21} = 0$$

$$\mathbf{F}_{13} = -\mathbf{F}_{31}; \ \mathbf{F}_{13} + \mathbf{F}_{31} = 0$$

$$\mathbf{F}_{23} = -\mathbf{F}_{32};\ \mathbf{F}_{23} + \mathbf{F}_{32} = 0$$

It follows that the sum of all internal forces in a system must be zero.

Extending the argument to a system of a number of particles, say n, the equilibrium demands that

1. *The vector sum of all external forces is zero*

$$\Sigma\mathbf{F}_e = 0$$

2. *The vector sum of all external plus all internal forces must be zero*

$$\Sigma\mathbf{F}_e + \Sigma\mathbf{F}_i = 0$$

in order that each particle be in equilibrium separately.

It is interesting to note that a system of three particles must lie in a plane and the forces on the particles must constitute a coplanar force system for equilibrium. A system of four or more particles may not be coplanar and may constitute spatial force systems for equilibrium.

UNIT V

MODELLING AND ERGONOMICS

- ✓ **Introduction to finite element analysis**
- ✓ **Analysis of biomechanical system using finite element methods**
- ✓ **Ergonomics**
- ✓ **Gait analysis**
- ✓ **Design of work station**
- ✓ **Sports biomechanics**
- ✓ **Injury mechanics**

LIST OF IMPORTANT QUESTIONS

PART A

1. **What is Ergonomics? Mention its importance in biomechanics or Define ergonomics(N/D 2016)(A/M 2017)**
2. **What are the advantages of ergonomics?**
3. **Define gait cycle.**
4. **List out major events during the gait cycle.**
5. **Explain the terms in gait cycle**
   i. **Cadence**
   ii. **Cycle time**
   iii. **Speed**
6. **What is finite element analysis(FEA)?(N/D 2015)**
7. **What are the applications of finite element analysis in biomechanics (FEA)?**
8. **List the limitations of finite element modeling. (N/D 2015)**
9. **Differentiate between intrinsic and extrinsic factors related to injury. (M/J 2016)**
10. **Define sport and exercise biomechanics. (M/J 2016).**
11. **What are the goals of sport mechanics?(N/D 2016)**
12. **What is stride length(A/M 2017)**
13. What are the primary factors for developing musculoskeletal disorder?
14. How can prevent injuries?
15. What are four basic methods of formulating the equations of finite element analysis?

PART B

1. Write short notes on finite element analysis (or) Analysis of biomechanical system using finite element methods (or) Explain FEA applications in biomechanics **or Describe the role of Finite element modeling in biomechanics (N/D 2015) or What is finite element analysis? Explain any one analysis of biomechanical system using FEM.(A/M 2017)**
2. **Explain about the gait cycle or explain about gait analysis.(A/M 2017)**
3. **Discuss the applications of gait analysis.(M/J 2016),(N/D 2015)**
4. **Explain ergonomic principles that contribute to good workplace design (or) work station.**
5. **Describe sports biomechanics**
6. **Write a brief notes on Injury mechanics or explain about injury mechanics(A/M 2017)**
7. **Differentiate between intrinsic and extrinsic factors related to injury development.(N/D 2016)**
8. **Identify the commonly measured parameters in gait analysis. Describe about them with illustrations.(N/D 2016)**
9. **Describe the methods used to achieve the goals of sport and exercise in biomechanics.(M/J 2016)**
10. Explain overview of the finite element method

PART A

1. **What is Ergonomics? Mention its importance in biomechanics or Define ergonomics(N/D 2016)(A/M 2017)**
    - Ergonomics is the science of fitting the job to the worker, matching the physical requirements of the job with the physical capacity of the worker.
    - Ergonomics is used to design an environment (layout, work methods, equipment, noise, etc) which is compatible with each individual's physical and behavioral characteristics.

    Importance of ergonomics

    - Ergonomics is important because when you're doing a job and your body is stressed by an awkward posture, extreme temperature, or repeated movement your musculoskeletal system is affected. Your body may begin to have symptoms such as fatigue, discomfort, and pain, which can be the first signs of a musculoskeletal disorder.
2. **What are the advantages of ergonomics?**
    - Increased savings
    - Fewer employees experiencing pain
    - Increased productivity
    - Increased morale
    - Reduced absenteeism
3. **Define gait cycle?**
    - The gait cycle is defined as the time interval between two successive occurrences of one of the repetitive events of walking.
    - Although any event could be chosen to define the gait cycle, it is generally convenient to use the instant at which one foot contacts the ground ('initial contact').
    - If it is decided to start with initial contact of the right foot then the cycle will continue until the right foot contacts the ground again.
    - The left foot, of course, goes through exactly the same series of events as the right, but displaced in time by half a cycle.

4. **List out major events during the gait cycle.**
    - Initial contact
    - Opposite toe off
    - Heel rise
    - Opposite initial contact
    - Toe off
    - Feet adjacent
    - Tibia vertical

5. **Explain the terms in gait cycle**
    - **Cadence-** The cadenceis the number of steps taken in a given time, the usual units being steps per minute.
    - **Cycle time-** A scientifically acceptable alternative would be to measure cadence in steps per second, but there is currently a trend to replace cadence entirely by a quantity which is inversely related to it – the cycle time(stride time) in seconds:

      Cycle time (s) = 120/cadence (steps/min)
    - **Speed -** The speedof walking is the distance covered by the whole body in a given time. It should be measured in meters per second.

      Speed (m/s) = stride length (m)/cycle time (s)

6. **What is finite element analysis(FEA) (N/D 2015)**
    - FEM is a simulation based displacement formulation to calculate component displacements, strains, and stresses under internal and external loads.
    - Estimations of material stress and strain are necessary during the course of device design to minimize the chance of device failure. For example, artificial hip joints need to be designed to withstand the loads that they are expected to bear without fracture or fatigue.
    - Applying analytical methods to such problems would require so many assumptions and simplifications.

- The most popular numerical method for solving problems in continuum mechanics is the finite element method (FEM), also referred to as finite element analysis (FEA).

7. What are the applications of finite element analysis in biomechanics (FEA)?

- Analysis of Commonplace Maneuvers at Risk for Total Hip Dislocation
- A Finite Element Model for the Lower Cervical Spine
- Finite Element Analysis of Indentation Tests on Pyrolytic Carbon
- Numerical Analysis of 3D Flow in an Aorta through an Artificial Heart Valve

8. List the limitations of finite element modeling.(N/D 2015)

- FEA is costly
- Modelling errors
- Incorrect boundary conditions
- Auto meshing
- Not all CAD software have compatible FEA software(Models not appropriate for software)
- Incorrect use of software

9. Differentiate between intrinsic and extrinsic factors related to injury. (M/J 2016)

Intrinsic factors

- Flexibility and joint laxity
- Nutrition
- Leg length discrepancies
- Fitness levels
- Age
- Weight/size

Extrinsic factors

- Coaching
- Incorrect techniques
- Environmental factors
- Clothing, footwear and equipment
- Safety hazards

10. Define sport and exercise biomechanics. (M/J 2016)

Sport and Exercise Biomechanics is a title that encompasses the area of science concerned with the analysis of mechanics of human movement. In other words it is the science of explaining how and why the human body moves in the way that it does.

Sports biomechanics:

- Sport biomechanics studies the effects of forces and motion on sport performance.
- By understanding and applying mechanical concepts, sports biomechanics assess the most optimal way to move the body in order to achieve maximal performance, whilst minimizing risk of injury.

11. What are the goals of sport mechanics?(N/D 2016)

- The major goal of biomechanics of sport and physical exercise is to improve performance in given sport or physical exercise.
- The secondary goal of sport biomechanics is to provide recommendations for injury prevention and rehabilitation.

12. What is stride length(A/M 2017)

- **Stride length** is measured from heel to heel and determines how far you walk with each **step**.
- the**stride length** is the distance **between** two successive placements of the same foot.
- It consists of two **step lengths**, left and right, each of which is the distance by which the named foot moves forward in front of the other one.

- In pathological gait, it is perfectly possible for the two **step lengths** to be **different**.

13. What are the primary factors for developing musculoskeletal disorder?

- Force
- Heaving lifting
- Push or pull
- Carrying
- Awkward or prolonged postures
- Repetitive activities
- Overhead work
- Extreme temperatures

14. How can prevent injuries?

- Use good body mechanics
- Stay physically fit and active
- Use mechanical assist devices when possible (lifts, carts, dolly)
- Get help when necessary.

15. What are four basic methods of formulating the equations of finite element analysis?

- The direct or displacement method
- The variational method
- The weighted residual method
- The energy balance method

PART B

1. Write short notes on finite element analysis (or) Analysis of biomechanical system using finite element methods (or) Explain FEA applications in biomechanics **or Describe the role of Finite element modeling in biomechanics (N/D 2015) or What is finite element analysis? Explain any one analysis of biomechanical system using FEM.(A/M 2017)**

INTRODUCTION

Estimations of material stress and strain are necessary during the course of device design to minimize the chance of device failure. For example, artificial hip joints need to be designed to withstand the loads that they are expected to bear without fracture or fatigue. Stress analysis is therefore required to ensure that all components of the device operate below the fatigue limit. For deformable structures such as diaphragms for artificial hearts, an estimate of strains or deformations is required to ensure that during maximal deformation, components do not contact other structures, potentially causing interference and unexpected failure modes such as abrasion.

For simple calculations, such as the sizing of a bolt to connect two components that bear load, simple analytical calculations usually suffice. Often, these calculations are augmented by reference to engineering tables that can be used to refine the stress estimates based on local geometry, such as the pitch of the threads. Such analytical methods are preferred because they are exact and can be supported by a wealth of engineering experience.

Unfortunately, analytical solutions are usually limited to linear problems and simple geometries governed by simple boundary conditions. The boundary conditions can be considered input data or constraints on the solution that are applied at the boundaries of the system. Most practical engineering problems involve some combination of material or geometrical nonlinearity, complex geometry, and mixed boundary conditions.

In particular, all biological materials have nonlinear elastic behavior and most experience large strains when deformed. As a result, nonlinearities of one form or the other are usually present in the formulation of problems in biomechanics. These nonlinearities are described by the equations relating stress to strain and strain to displacement.

Applying analytical methods to such problems would require so many assumptions and simplifications that the results would have poor accuracy and would thus be of little engineering value. There is therefore no alternative but to resort to approximate or numerical methods. The most popular numerical method for solving problems in continuum mechanics is the finite element method (FEM), also referred to as finite element analysis (FEA).

FEA is a computational approach widely used in solid and fluid mechanics in which a complex structure is divided into a large number of smaller parts, or elements, with interconnecting nodes, each with geometry much simpler than that of the whole structure. The behavior of the unknown variable within the element and the shape of the element are represented by simple functions that are linked by parameters that are shared between the elements at the nodes. By linking these simple elements together, the complexity of the original structure can be duplicated with good fidelity (Faithfulness). After boundary conditions are taken into account, a large system of equations for the unknown nodal parameters always results; these equations are solved simultaneously by a computer, using indirect or iterative means Finite element analysis is extremely versatile (Adapt). The size and configuration of the elements can be adjusted to best suit the problem; complex geometries can be discretized and solutions can be stepped through time to analyze dynamic systems.

Very often, simple analytical methods are used to make a first approximation to the design of the device, and FEA is subsequently used to further refine the design and identify potential stress concentrations. FEA can be applied to both solids and fluids or, with additional complexity, to systems containing both.

FEA software is very mature and computing power is now sufficiently cheap to allow finite element methods to be applied to a wide range of problems. In fluid flow, FEA has been applied to weather forecasting and supersonic flow around aircraft and within engines, and in the medical field, to optimizing blood pumps and cannulas. In solids, FEA has been used to design, build, and crash automobiles, estimate the impact of earthquakes, and reconstruct crime scenes.

In biomaterials, FEA has been applied to almost every implantable device, ranging from artificial joints to pacemaker leads. Although originally developed to help structural engineers analyze stress and strain, FEA has been adopted by basic scientists and biologists to study the dynamic environment within arteries, muscles and even cells.

(i) Analysis of Commonplace Maneuvers at Risk for Total Hip Dislocation

Dislocation is a frequent complication of total hip arthroplasty (THA). In this FE study, a motion tracking system and a recessed force plate were used to capture the kinematics and ground reaction forces from several trials of realistic dislocation-prone maneuvers performed byactual subjects. Kinematics and kinetic data associated with the experiments were imported into a FE model of THA dislocation. The FE model was used to compute stresses developed within the implant, given the observed angular motion of the hip and contact force inferred from inverse dynamics.

The FE mesh wascreated using PATRAN version 8.5 and the simulations were executed with ABAQUS version 5.8. In the FE analysis, the resultant resisting moment developed around the hip-cup center was tracked, as a function of hipangle. The peak of this resistive moment was a key outcome measure used to estimate the relative risk of dislocations from the motions. All seven maneuvers studied led to frequent instances of computationally predicted dislocation

FE mesh

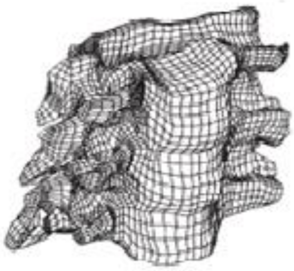
Maneuver	No. of trials	No. of dislocations	% of trials dislocating
Low sit-to-stand	47	41	87
Normal sit-to-stand	55	33	64
Tie	69	31	45
Leg cross	64	22	34
Stoop	42	6	14
Post. disloc. maneuvers	277	133	48
Pivot	58	23	40
Roll	19	12	63
Ant. disloc. maneuvers	77	35	45
Overall series	353	168	47

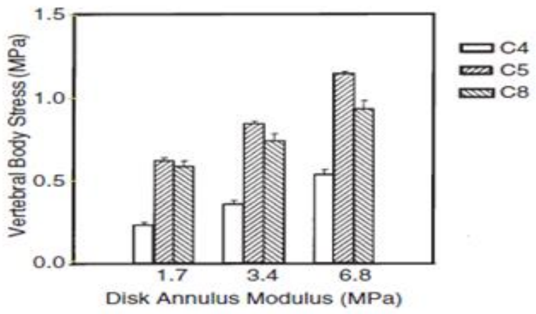
The conclusion that this library of dislocation-prone maneuvers appear to substantially extend the information base previously available to study this important complication of THA.

(ii) A Finite Element Model for the Lower Cervical Spine

A parametric study was conducted to determine the variations in the biomechanical responses of the spinal components in the lower cervical spine. Axial compressive load was imposed uniformly on the superior surface of the C4-C6 unit. The various components were assumed to have linear isotropic and homogeneous elastic behavior and appropriate material parameters were taken from the literature.

A detailed 3D finite element model was reconstructed from 1.0-mm CT scans of a human cadaver, resulting in a total of 10,371 elements. The results show that an increase in elastic moduli of the disks resulted in an increase in endplate stresses and that the middle C5 vertebral body produced the highest compressive stresses. The model appears toconfirm clinical experience that cervical fractures are induced by external compressive forces.



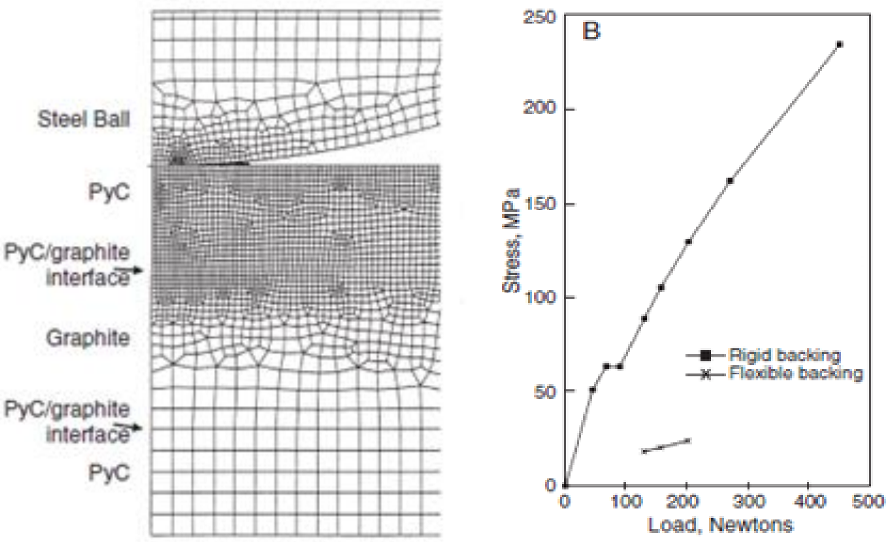


3D solid FE Mesh **stress as a function of disk annulus moduli**

(iii)Finite Element Analysis of Indentation Tests on Pyrolytic Carbon

Pyrolytic carbon (PyC) heart valves are known to fail through cracks initiated at the contact areas between leaflets and their housing. This phenomenon is simulated with a 5.1-mm steel ball indenting a graphite sheet coated on each side with PyC, similar to the makeup of real heart valves. Two types of contacts were analyzed: when the surface material is thick (rigid backing) and when it is fairly thin (flexible backing).

FEA was used to evaluate the stresses resulting from a range of loads. The geometry was taken to be axisymmetric, PyC was assumed to be an elastic material and quadrilateral solid elements were used. The mesh is refined in the contact areas but gets progressively coarser toward the noncontact areas. The maximum principal stress on the PyC surface adjacent to ball contact, as a function of the indentation load. “Flexible backing” is seen to greatly reduce the maximum principal stress in this area. The FE results were correlated with data from experiments and used to develop failure criteria for contact stresses. This in turn provided criteria for designing contact regions in pyrolytic heart valves.



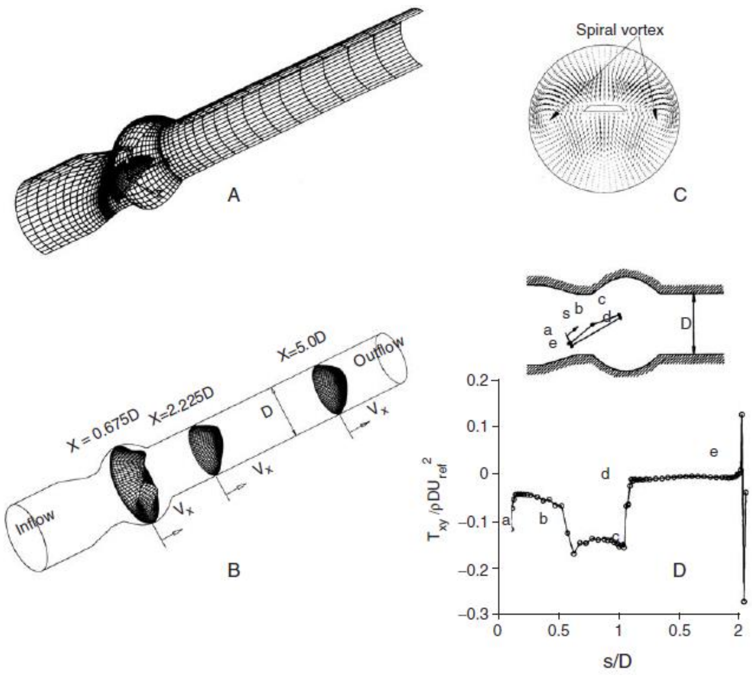
Part of the FE mesh showing aMaximum principal stress on thePyCSteel ball in contact with a PyC.surface adjacent to ball contact radius.

(iv) Numerical Analysis of 3D Flow in an Aorta through an Artificial Heart Valve

Three-dimensional transient flow past a Bjork-Shiley valve (Tilting disc) in the aorta is simulated by the FEM combined with a time stepping algorithm. The FE mesh comprises some 32,880 elements and 36,110 nodes. The results indicate that the flow is split into two major jet flows by the valve, which later merge downstream.

A 3D plot of velocity vectors show large velocities in the upper and lower jet flow regions in the sinus region, large velocities only in the upper part of the merged jet, and an almost uniform parabolic distribution near the outflow region.Twin spiral vortices are generated immediately downstream of the valve, in the sinus region and are convected (Transport) downstream, where they quickly die away by diffusion. Shear stress along the surface of the valve is shown to be a maximum in the vicinity (region) of its leading edge.

A study such as this provides useful information on the function of the valve in vivo.



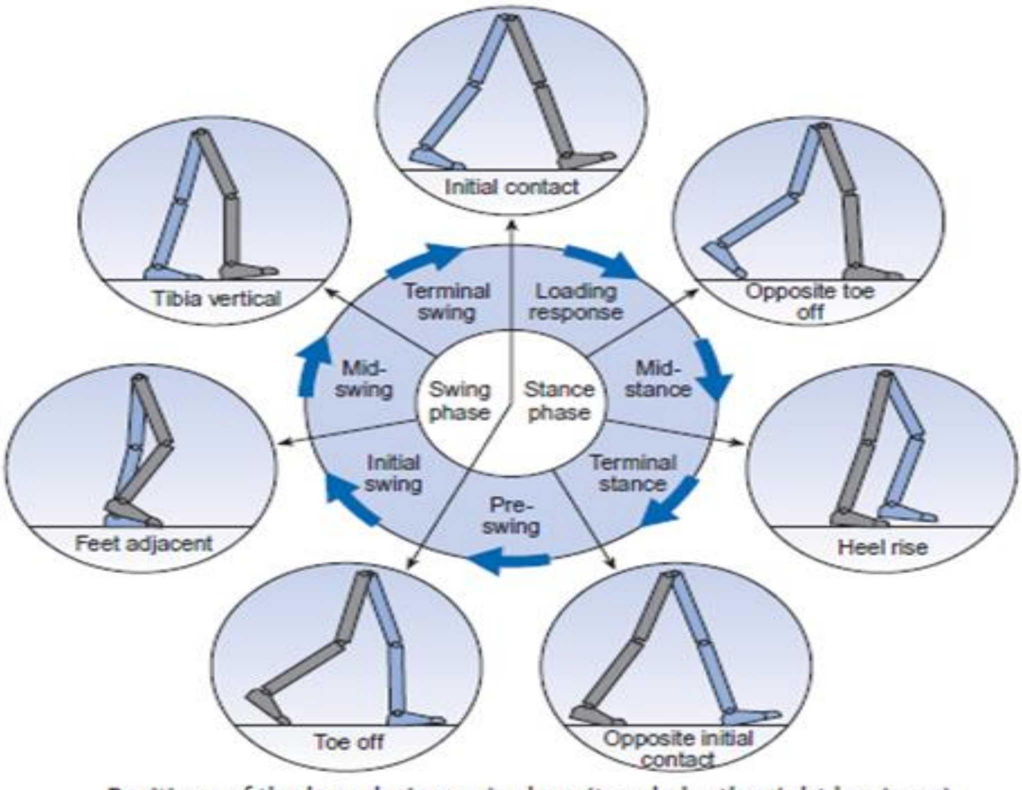
FE analysis of transient 3D flow past a Bjork-Shiley valve in the aorta

(A) Surface grid of aorta and fully opened Bjork-Shiley valve prosthesis, (B) carpet plot of axial velocity vectors, (C) secondary flow vector plot showing spiral vortices, and (D) shear stress along the valve surface in the symmetric mid-plane.

2. Explain about the gait cycle(M/J 2016)(A/M 2017).

The gait cycle is defined as the time interval between two successive occurrences of one of the repetitive events of walking. Although any event could be chosen to define the gait cycle, it is generally convenient to use the instant at which one foot contacts the ground ('initial contact'). If it is decided to start with initial contact of the right foot, as shown in Fig, then the cycle will continue until the right foot contacts the ground again. The left foot, of course, goes through exactly the same series of events as the right, but displaced in time by half a cycle. The following terms are used to identify major events during the gait cycle:

1. Initial contact
2. Opposite toe off
3. Heel rise
4. Opposite initial contact
5. Toe off
6. Feet adjacent
7. Tibia vertical



Positions of the legs during a single gait cycle by the right leg (gray).

These seven events subdivide the gait cycle into seven periods, four of which occur in the stance phase, when the foot is on the ground, and three in the swing phase, when the foot is moving forward through the air. The stance phase, which is also called the ‘support phase’ or ‘contact phase’, lasts from initial contact to toe off. It is subdivided into:

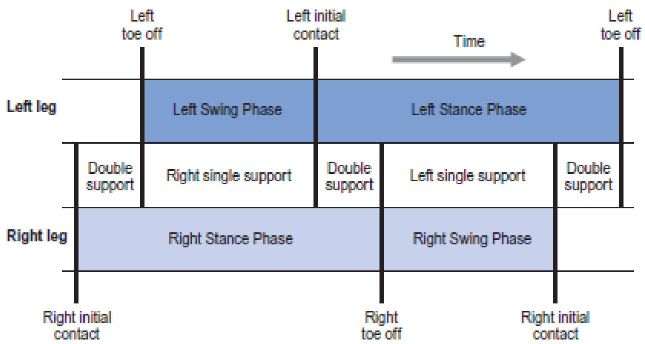
1. Loading response
2. Mid-stance
3. Terminal stance
4. Pre-swing.

The swing phase lasts from toe off to the next initial contact. It is subdivided into:

1. Initial swing
2. Mid-swing
3. Terminal swing.

Gait cycle timing

The duration of a complete gait cycle is known as the cycle time, which is divided into stance time and swing time. The timings of initial contact and toe off for both feet during a little more than one gait cycle. Right initial contact occurs while the left foot is still on the ground and there is a period of double support (also known as ‘double limb stance’) between initial contact on the right and toe off on the left. During the swing phase on the left side, only the right foot is on the ground, giving a period of right single support (or ‘single limb stance’),which ends with initial contact by the left foot.



Timing of single and double support during a little more than one gait cycle is starting with right initial contact.

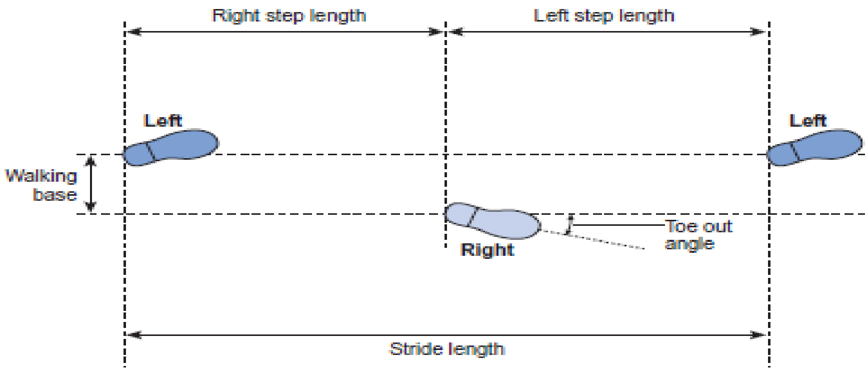
There is then another period of double support, until toe off on the right side. Left single support corresponds to the right swing phase and the cycle ends with the next initial contact on the right.

In each double support phase, one foot is forward, having just landed on the ground, and the other one is backward, being just about to leave the ground. When it is necessary to distinguish between the two legs in the double support phase, the leg in front is usually known as the ‘leading’ leg and the leg behind as the ‘trailing’ leg. The leading leg is in ‘loading response’, sometimes referred to as ‘braking double support’, ‘initial double support’ or ‘weight acceptance’.

The trailing leg is in ‘pre-swing’, also known as ‘second’, ‘terminal’ or ‘thrusting’ double support or ‘weight release’. In each gait cycle, there are thus two periods of double support and two periods of single support. The stance phase usually lasts about 60% of thecycle, the swing phase about 40%. However, this varies with the speed of walking, the swing phase becoming proportionately longer and the stance phase and double support phases shorter, as the speed increases. The final disappearance of the double support phase marks the transition from walking to running.Between successive steps in running there is a flight phase, also known as the ‘float’, ‘double-float’ or ‘non-support’ phase, when neither foot is on the ground.

Foot placement

The placement of the feet on the ground is shown in the figure. The stride length is the distance between two successive placements of the same foot. It consists of two step lengths, left and right, each of which is the distance by which the named foot moves forward in front of the other one.



The terms used to describe foot placement on the ground

In pathological gait, it is common for the two step lengths to be different. If the left foot is moved forward to take a step and the right one is brought up beside it, rather than in front of it, the right step length will be zero. It is even possible for the step length on one side to be negative, if that foot never catches up with the other one. However, the stride length, measured between successive positions of the left foot, must always be the same as the stride length measured between successive positions of the right foot, unless the subject is walking around a curve.

This definition of a 'stride', consisting of one 'step' by each foot, breaks down in some pathological gaits, in which one foot makes a series of 'hopping' movements while the other is in the air .There is no satisfactory nomenclature to deal with this situation.

The walking base (also known as the 'stride width' or 'base of support') is the side-to-side distance between the line of the two feet. The preferred units for stride length and step length are meters and for the walking base, millimeters. The pattern of walking known as 'tandem gait' involves walking with the heel of one foot placed directly in front of the toes of the other, i.e. with a walking base close to zero. Although this pattern is not typically seen, even as a pathological gait, it requires good balance and coordination and it is often used by the police as a test for intoxication!

The toe out (or, less commonly, toe in) is the angle in degrees between the direction of progression and a reference line on the sole of the foot. The reference line varies from one study to another.

It is common experience that you need to walk more carefully on ice than on asphalt. Whether or not the foot slips during walking depends on two things: the coefficient of friction between the foot and the ground, and the relationship between the vertical force and the forces parallel to the walking surface (front-to-back and side-to-side). The ratio of the horizontal to the vertical force is known as the 'utilized coefficient of friction' and slippage will occur if this exceeds the actual coefficient of friction between the foot and the ground. In normal walking, a coefficient of friction of 0.35–0.40 is generally sufficient to prevent slippage; the most hazardous part of the gait cycle for slippage is initial contact.

Cadence, cycle time and speed

The cadence is the number of steps taken in a given time, the usual units being steps per minute. In most other types of scientific measurement, complete cycles are counted, but as there are two steps in a single gait cycle, the cadence is a measure of half-cycles. Measurement in 'steps per minute' does not conform with the system International (SI).

cycle time (s) = 120/cadence (steps/min)

The normal ranges for both cadence and cycle time in both at different ages.

The speed of walking is the distance covered by the whole body in a given time. It should be measured in meters per second. The instantaneous speed varies from one instant to another during the walking cycle, but the average speed is the product of the cadence and the stride length, providing appropriate units are used. The cadence, in steps per minute, corresponds to half-strides per 60 seconds or full strides per 120 seconds.

The speed can thus be calculated from cadence and stride length using the formula:

Speed (m/s) = stride length (m) × cadence (steps/min)/120

If cycle time is used in place of cadence, the calculation becomes much more straightforward:

Speed (m/s) = stride length (m)/cycle time (s)

The walking speed thus depends on the two step lengths, which in turn depend to a large extent on the duration of the swing phase on each side. The step length is the amount by which the foot can be moved forwards during the swing phase, so that a short swing phase on one side will generally reduce the step length on that side. If the foot catches on the ground, this may terminate the swing phase and thereby further reduce both step length and walking speed. In pathological gait, the step length is often shortened, but it behaves in a way which is counterintuitive. When pathology affects one foot more than the other, an individual will usually try to spend a shorter time on the 'bad' foot and correspondingly longer on the 'good' one.

Shortening the stance phase on the 'bad' foot means bringing the 'good' foot to the ground sooner, thereby shortening both the duration of the swing phase and the step length on that side. Thus, a short step length on one side generally means problems with single support on the other side.

3. Discuss the applications of gait analysis.(N/D 2015)

- The applications of gait analysis are conveniently divided into two main categories: *clinical gait assessment* and *gait research*.
- Clinical gait assessment has the aim of helping individual patients directly, whereas gait research aims to improve our understanding of gait, either as an end in itself or in order to improve medical diagnosis or treatment in the future.
- There is obviously some overlap, in that many people performing clinical gait assessment use it as the basis for research studies. Indeed, this is the way that most progress in the use of clinical gait assessment is made.

Clinical gait assessment

- Clinical gait assessment seeks to describe, on a particular occasion, the way in which a person walks.
- This may be all that is required, if the aim is simply to document their current status.
- Alternatively, it may be just one step in a continuing process, such as the planning of treatment or the monitoring of progress over a period of time.
- The simplest form of gait assessment is practised every day in physician offices, physiotherapy and rehabilitation clinics, orthotic and prosthetic clinics, sports centres, and many other settings throughout the world.
- Every time a clinician watches a client or patient walk up and down a room, they are performing an assessment of the patient's gait.
- However, such assessment is often unsystematic and the most that can be hoped for is to obtain a general impression of how well the patient walks, and perhaps some idea of one or two of the main problems.
- This could be termed an 'informal' gait assessment. To perform a 'formal' gait assessment requires a careful examination of the gait, using a systematic approach, if possible, augmented by objective measurements.

- Such a gait assessment will usually produce a written report and the discipline involved in preparing such a report is likely to result in a much more carefully conducted assessment.
- The gait analysis techniques which are used in clinical gait assessment vary enormously, with the nature of the clinical condition, the skills and facilities available in the individual clinic or laboratory and the purpose for which the assessment is being conducted.
- In general, however, clinical gait assessment is performed for one of three possible reasons: it may form the basis of clinical decision making, it may help with the diagnosis of an abnormal gait, or it may be used to document a patient's condition. These will be considered in turn.

Clinical decision making

1. Gait assessment:

This starts with a full clinical history, both from the patient and from any others involved, such as doctors, therapists or family members. Where a patient has previously had surgery, details of this should be obtained, if possible from the operative notes. History taking is followed by a physical examination, with particular emphasis on the neuromusculoskeletal system. In many laboratories, physical examinations are performed by both a doctor and a physiotherapist. Finally, a formal gait analysis is carried out.

2. Hypothesis formation:

The next stage is the development of a hypothesis regarding the cause or causes of the observed abnormalities. This hypothesis is often informed by the specific questions raised by the referring doctor. Time needs to be set aside to review the data, and consultation between colleagues, particularly those from different disciplines, is extremely valuable. Indeed, almost all of that using gait assessment as a clinical decision-making tool stress the value of this 'team approach'. In forming a hypothesis as to the fundamental problem in a patient with a gait disorder

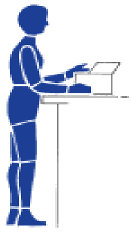
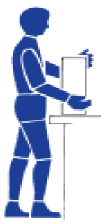
3. Hypothesis testing:

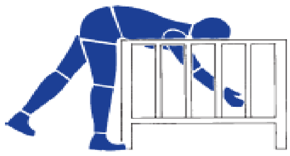
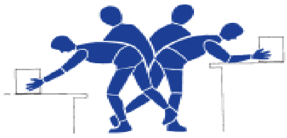
- This stage is sometimes omitted, when there is little doubt as to the cause of the abnormalities observed.
- However, where some doubt does exist, the hypothesis can be tested in two different ways – either by using a different method of measurement or by attempting in some way to modify the gait.
- Some laboratories routinely use a fairly complete 'standard protocol', including video recording, kinematic measurement, force platform measurements and surface electromyography (EMG).
- They will then add other measurements, such as fine wire EMG, where this is necessary to test a hypothesis. Other clinicians start the gait analysis using a simple method, such as video recording, and only add other techniques, such as EMG or the use of a force platform, where they would clearly be helpful.
- Different types of gait analysis data may be useful for different aspects of the gait assessment. Information on foot timing may be useful to identify asymmetries and may indicate problems with balance, stability, pain etc.
- The general gait parameters give a guide to the degree of disability and may be used to monitor progress or deterioration with the passage of time.
- The kinematics of limb motion describes abnormal movements, but do not identify the 'guilty' muscles.
- The most useful measures are probably the joint moments and joint powers, particularly if this information is supplemented by EMG data.
- Hemiparetic patients may show greater differences between the two sides in muscle power output than in any of the other measurable parameters, including EMG.

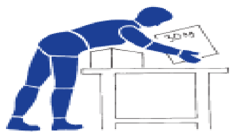
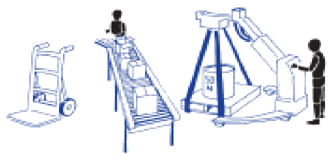
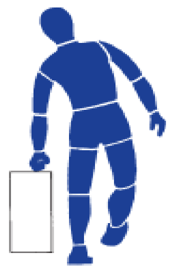
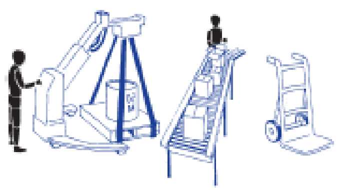
4. Explain ergonomic principles that contribute to good workplace design(or) work station.

The goal for the design of workplaces is to design for as many people as possible and to have an understanding of the Ergonomic principles of posture and movement which play a central role in the provision of a safe, healthy and comfortable work environment. Posture and movement at work will be dictated by the task and the workplace, the body's muscles, ligaments and joints are involved in adopting posture, carrying out a movement and applying a force. The muscles provide the force necessary to adopt a posture or make a movement. Poor posture and movement can contribute to local mechanical stress on the muscles, ligaments and joints, resulting in complaints of the neck, back, shoulder, wrist and other parts of the musculoskeletal system.

Ergonomic principles provide possibilities for optimizing tasks in the workplace These principles are summarizedbelow.

Ergonomic Principle		Description
Joints must be in a neutral position		In the neutral position the muscles and ligaments, which span the joints, are stretched to the least possible extent
Keep work close to the body		If the work is too far from the body, the arms will be outstretched and the trunk bent over forwards

Ergonomic Principle		Description
Avoid bending forward		The upper part of the body of an adult weighs about 40kg on average. The further the trunk is bent forwards, the harder it is for the muscles and ligaments of the back to maintain the upper body in balance
A twisted trunk strains the back		Twisted postures of the trunk cause undesirable stress to the spine

Alternate posture as well as movements		No posture or movement should be maintained for a long period of time. Prolonged postures and repetitive movements are tiring.
Avoid excessive reaches		It is necessary to limit the extent of forward and sideways reaches to avoid having to bend over or twist the trunk
Avoid carrying out tasks above shoulder level		The hands and elbows should be well below shoulder level when carrying out a task
Limit the weight of a load that is lifted		There are guidance weight limits for both males and females detailed in Figure 2 of this document
Use mechanical aids		Many lifting accessories are available to help lift and move loads
Avoid carrying loads with one hand		When only one hand is used to carry a load, the body is subject to mechanical stress
Use transport accessories		There are a large number of accessories such as roller conveyors, conveyor belts, trolleys and mobile raising platforms, which eliminate or reduce manual handling.

Advantages of ergonomics

- **Increased savings**
  - Fewer injuries, fewer workers' compensation claims.
  - More productive and sustainable employees.
- **Fewer employees experiencing pain**
  - Implementing ergonomic improvements can reduce the risk factors that lead to discomfort.
- **Increased productivity**
  - Ergonomic improvements can reduce the primary risk factorsfor MSDs, so workers are more efficient, productive, and have greater job satisfaction.
- **Increased morale**
  - Attention to ergonomics can make employees feel valued because they know their employer is making their workplace safer.
- **Reduced absenteeism**
  - Ergonomics leads to healthy and pain-free workers who are more likely to be engaged and productive.

Direct costs are those directly associated with the claim and include:

- Medical treatment
- Prescription costs
- Insurance premiums

Indirect costs associated with the injury can include:

- Overtime due to staff coverage during absence of injured worker
- Accommodation for modified duty
- Increased absenteeism
- Decreased morale
- Legal and investigation costs
- Replacement worker costs
- Orientation and training costs
- Advertising and recruiting if employee doesn't return to work
- Presenteeism: when an employee comes back to work too early and is less productive than in a healthy state.

Primary risk factors for developing an MSD

- Force
- Heaving lifting
- Push or pull
- Carrying
- Awkward or prolonged postures
- Repetitive activities
- Overhead work
- Contact stress
- Extreme temperatures

The work likely to result in an injurywhen

- It performed frequently
- It performed for a long period of time
- The work is intense
- There is a combination of several risk factors.

Risk reduction techniques

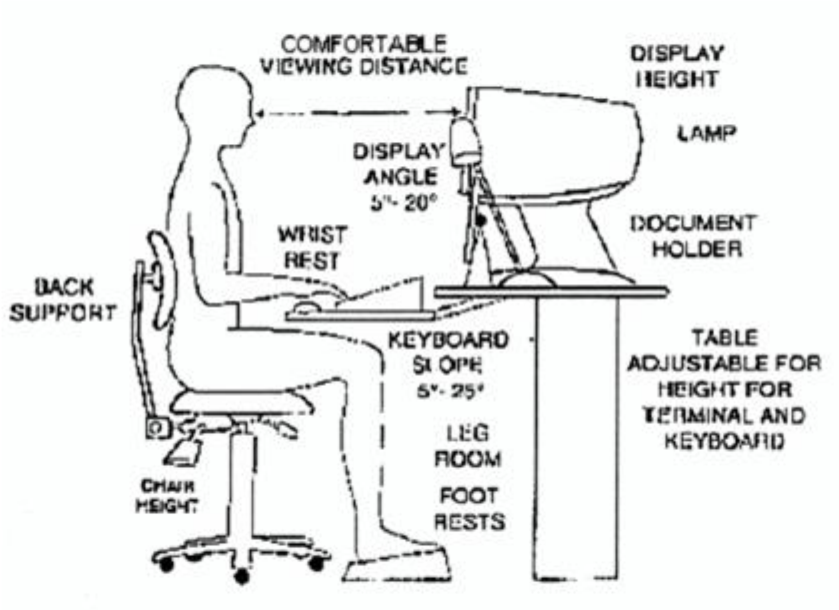
- **Job rotation**
  - Cross train workers so they can rotate jobs throughout the day.
  - Change tasks often within your own job (Example: type for an hour and then file for an hour).
- **Job hazard analysis**
  - Break each job up into smaller or different tasks.
  - Determine the risk factors for each task.
  - Determine how each task affects risk factors for the total job.
- **Select appropriate tools**
  - By attaching a handle extension, bending can be eliminated from many jobs.
- **Participatory ergonomics**
  - Enlist workers to brainstorm better ways to do their work.
  - Have trained workers teach new staff.

Ergonomics training

• Educate staff on the risk factors for musculoskeletal disorders, and how ergonomics can make their work easier, more efficient, and safer.

• Train staff to identify job tasks that may present a risk and determine better ways to complete those tasks.

Workstation Design



- **Adjust your furniture to fit you!**
  Chairs should provide adjustable height and arms in addition to comfortable lumbarsupport. They should also have five legs for stability and be made of fabric thatbreathes for comfort. Chairs are made in different sizes; your chair should be theright size for you.
- **Place Keyboards to provide comfortable (neutral) posture.**
  Place keyboard at approximately seated elbow height (25-32" from floor), shouldersshould be relaxed. Work with wrists straight **- NOT BENT**.
- **Bring your work within reach.**
  Bring your mouse, telephone, pens, notepads, and other frequently used items withinreach. Place the mouse close to the keyboard to avoid reaching for it.
- **Adjust the monitor to provide the most comfortable viewing position,**
  Position the top of the screen at or just below eye level. Maintain a comfortable eyeto monitor distance, generally an arm's length (appro.24").

- **Comprehensive ergonomics program**

  A comprehensive ergonomics program can save your company money. It must include several elements.

  - **Worker involvement** – workers must be involved in all aspects of the ergonomics program.
  - **Management commitment** – leaders must make employee health and safety a priority.
  - **Training** – employees need to be trained to understand ergonomics — why it's important and what is expected of them.
  - **Sustainability** – your program should become part of your safety committee/safety meetings.
  - **Evaluation** – maintain company statistics on annual MDS claims, direct and indirect costs, and number and outcomes of completed job analysis. This will help build your case when you present an issue to management and staff.

5.Describe sports biomechanics

Sport biomechanics studies the effects of forces and motion on sport performance.By understanding and applying mechanical concepts, sports biomechanics assess the most optimal way to move the body in order to achieve maximal performance, whilst minimizing risk of injury.

Mechanical Principles
The Rules of Sport Skill Technique

Mechanical principles of physics can be applied to sports. Using these rules as guides, athletes can achieve excellent technique to gain the greatest mechanical advantage. Newton's Laws of Motion are the foundation for these mechanical principles, which must be applied in concert with othertrainingprinciplesto achieve higher performance levels.

These mechanical principles of physics form a valuable guide for developing the optimum skating technique. However, there are many interpretations on how to apply the physics principles to a training program and effective teaching methods.

Principles from the Law of Inertia: -

Newton's First Law states that an object will remain at rest or in uniform motion in a straight line unless acted upon by an external force.

- Achieving skilled movements requires the effectively combination of linear and angular motion. For example, the hooking action of the edge/turn to convert the linear motion into an angular motion that establishes the spin's center.

- Two or more motions must be executed continuously in sequence. For example, in a jump the skater must spring into the air, complete the require number of revolutions, and check the rotation to achieve a controlled landing on the correct edge.

- There must be a balance of mass and/or velocity between partners and members of a team. For example, each individual must alter their direction and force to stabilize their combined movements or centrifugal force will become uncontrollable.

- Control momentum efficiently for each body part to coordinate the entire body as a single unit. For example, changing positions from upright/layback to camel and/or sit spin positions.

Principles from the Law of Acceleration:-

Acceleration is a very important ability for a figure skater to possess.

- Acceleration/velocity is proportional to the force appliedagainst the ice. A skater who can increase his/her force applied to the ice increases their acceleration by an equal amount.

- The maximum acceleration is achieved when all body forces are coordinated to achieve thrust in either the forward or backward direction. Body actions that do not contribute to the forward or backward motion should be minimized to prevent wasted energy and/or detract from productive creation of power.

- Lengthening the radius of our arms and/or free leg slows the body rotation; shortening the radius increases rotation. For example, a skater will achieve their maximum spinning rotation when they pull in their arms and free leg tight to the body. A came; spin can never approach the speed achieved in a scratch spin because the radius is longer in a camel spin.

- A skater establishes the path in the air at take off. The axis of the core body may wobble or tilt on its axis which can adversely affects the skater's ability to complete the rotation in a vertical position, and land the jump on one foot in a controlled position.

The Principles of Counterforce:-

A stable surface maximizes the potential counterforce that can be generated when force is applied against it. The less stable the surface, the less counterforce is generated. For example, a skidded edge does not produce the same spring force into the air for the skater as if a clean takeoff had been achieved. The friction of the skid absorbs energy that is not transferred into the force that propels the skater into air.

- To achieve maximum jumping height, it is necessary to push directly down upon take off. The direction of counterforce is directly opposite that of the applied force, and the applied force is most effective when it is perpendicular to the supporting surface because skidding the edge is minimized.

- Maximization of total force. The combination of thrusting from the jump foot and the free leg kick in the axel produce the total force into the air.

Advantages of Sport Biomechanics

When coaches understand how forces work on muscles and affect motion in sports, they have a clear advantage over those who lack this knowledge and its applications. Athletes who know the basic concepts have a rationale for learning the correct way to execute skills. Knowing the reason behind learning a challenging technique gives them more motivation to master it.

The key to success is finding effective instructional cues that help the athlete achieve correct mechanical technique. Coaches with a command of mental training tools and sports training principles can help athletes make amazing things happen on the field.

Anatomy and physiology lay the foundation for biomechanics and kinesiology, areas of study about human movement. With a command of these areas, coaches can:

- Analyze sport movements,
- Select the best training exercises,
- Reduce or prevent injuries,
- Design or choose the sport equipment that best matches athletes' personal needs.
- Maximize economy and efficiency of movements.

The laws and principles of sport biomechanics are particularly helpful for designing training activities that match the mechanical demands of sports. For example, these principles help answer questions about the similarities between weight training exercises and sport movements and provide a rationale for which exercises a coach might include in the training program.

Enhancing Sport Instructional Technique

Biomechanics is the study of the effects of forces on a sport performance. Using laws and principles grounded in physics that apply to human movement, athletes and coaches can make sound decisions to develop efficient sport techniques.

Sports Biomechanics combines Biomechanics and Kinesiology into a unified field of science that applies the laws of mechanics and physics to human performance.

A greater understanding of performance in athletic events is achieved through modeling, simulation, and measurement. It is necessary to have a good understanding of physical principles that are applied to most sporting events –

- Motion
- Resistance
- Momentum
- Friction.

The emphasis of Sports Biomechanics is on the broad biomechanical spectrum of human performance including, but not limited to:

- Technique,
- Skill acquisition,
- Training, Coaching, Teaching,
- Exercises - Strength & Conditioning,
- Equipment,
- Modeling,
- Simulation,
- Measurement,
- Injury prevention,
- Rehabilitation

The communication of knowledge between scientists, coaches, clinicians, teachers, and participants contributes to the improvement of human performance and to reduce the incidence of injury in practices and competitions.

When coaches understand how forces work in sports and how athletes can leverage these forces, they have a clear advantage over those who lack these tools. Coaches with a command of both mental trainingtools and sports training principles can make amazing things happen on the field.

Biomechanics and kinesiology, areas of study about human movement, can help coaches:

- Analyze sport movements
- Identify and faithfully execute the best training exercises
- Reduce or prevent the likelihood of injuries occurring
- Choose equipment that is appropriately sized for the athlete

- Choose the athletic shoe that fits properly and provides the necessary support and flexibility as required for the specific sport
- Choose the boot and blades that properly fit and support the skater or skier; replace prior to equipment degradation/failure

6. Write a brief notes on Injury mechanics or explain about injury mechanics.(A/M 2017)

Injury causation is a central issue in many personal injury claims. Over the last decade, some courts have limited accident reconstruction engineers from opining about injuries and medical doctors from opining about biomechanics.

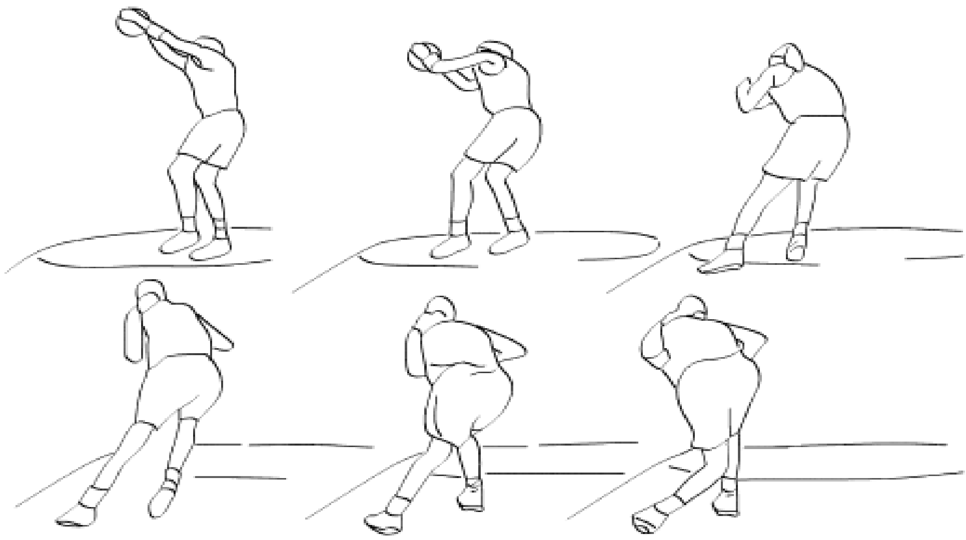
Who then can address issues of injury biomechanics?

The injury biomechanist: a professional who has training in both engineering mechanics and injury. The injury biomechanist is therefore qualified to calculate the forces applied to the body and compare these forces to the tolerance for specific injuries. Based on this comparison, an injury biomechanist can help prove or refute the causal relationship between the diagnosed injuries and the events alleged to have caused them.

Most sports medicine professionals must deduce the cause of injuries from the history presented by patients or clients. Occasionally athletic trainers may be at a practice or competition where they witness an injury. Knowledge of the biomechanical causes of certain injuries can assist an athletic trainer in these situations, in that diagnosis of the particular tissues injured is facilitated. Imagine you are an athletic trainer walking behind the basket during a basketball game. You look onto the court and see one of your athletes getting injured as she makes arebound. What kind of injury do you think occurred? What about the movement gave you the clues that certain tissues would be at risk of overload?

The athlete depicted in the following figure likely sprained several knee ligaments. Landingfrom a jump is a high-load event for the lower extremity, where muscle activity must be built up prior to landing. It is likely the awkward landing position, insufficient pre-impact muscle activity, and twisting (internal tibial rotation) contributed to the injury. It is also likely that the anterior (ACL) and posterior (PCL) cruciate ligaments were sprained. The valgus deformation of the lower leg would also suggest potential insult to the tibial (medial) collateral ligament. Female athletes are more likely to experience a non-contact ACL injury than males and the majority of ACL injuries are non-contact injuries. There are good recent reviews of knee ligament injury mechanisms.

You rush to the athlete with these injuries in mind. Unfortunately, any of these sprains are quite painful. Care must be taken to comfort the athlete, treat pain and inflammation, and prevent motion that would stress the injured ligaments. Joint tests and diagnostic imaging will eventually be used to diagnosis the exact injury.



A basketball player injuring her knee during a rebound.

Biomechanical analyses of injury

Biomechanical analyses of injury examine the causal relationship between a specific event and a specific set of diagnosed injuries. A summary of the injuries is drawn from the medical records and reports, and in the case of car crashes, data regarding the crash direction and severity is drawn from the collision reconstruction report. Thus biomechanical analyses rely on the evidence of other experts, and accurate reconstructions and diagnoses are needed for sound biomechanical analyses.

The biomechanical analysis itself consists of two main steps: mechanism and magnitude. To assess injury mechanism, the direction and location of the forces applied to the body are first determined for each event in question. This information is then compared to the direction and location of the forces required to cause each injury. If the direction and location of the required and applied forces do not match, then a mechanism for the diagnosed injury does not exist and the injury was not caused by the event in question. If, however, the direction and location of the applied and required forces do match, then a mechanism for the diagnosed injury exists and the analysis proceeds to the second step.

To perform the second step of the biomechanical analysis, the magnitude of the forces applied to or through the injured area is calculated for each event in question. The magnitude of the applied force is called the exposure and the threshold force above which an injury occurs is called the tolerance. The tolerance values are drawn from scientific studies published in the peer-reviewed literature. If the exposure is greater than or equal to the tolerance, then the injury is consistent with the event. Alternatively, if the exposure is less than the tolerance, then the injury is not related to the event in question.

Although the mechanism and magnitude analyses are relatively simple in theory, there are numerous factors that can complicate an injury biomechanics analysis. First, the medical diagnosis is sometimes unclear, particularly for soft-tissue injuries where the specific tissue injury responsible for the plaintiff's symptoms is often not identified. Second, the forces applied to the occupant may be difficult to calculate, either because of the nature of the event (e. g., rollover collisions) or the lack of scientific data. Third, the tolerance values for some diagnosed injuries or conditions (e. g., fibromyalgia or carpal tunnel syndrome) and the tolerance values for individuals with some pre-existing conditions are not known. Finally, there is considerable variation in the tolerance values for some injuries. Thus the quality of the diagnosis and the state of the science for a specific injury play a large role in the quality of the answer an injury biomechanist can provide regarding injury causation.

Biomechanics in action

Biomechanical analyses can be useful if it is unclear whether an injury is related to an event, or if the severity of an injury seems inconsistent with the exposure. Biomechanical analyses can be particularly effective from a defence perspective when alternate events or occupational exposures are compared to the exposure alleged to have caused the injury. A few examples of cases that benefit from a biomechanical analysis are given below:

The right-front passenger of a pickup truck that underwent a severe frontal collision sustained a left femur fracture and right knee and hip fractures. Heavy knee loading against the dash ahead of the passenger seat caused these fractures and indicated the passenger was not wearing his seatbelt. Seatbelt use would not have prevented knee contact with the dash, but the biomechanical analysis showed that the knee loads with a seatbelt would have been well below the fracture tolerance data in the scientific literature. Thus the passenger's fractures would have been prevented with seatbelt use.

A lumbar disc herniation was alleged to occur during a rear-end collision with a speed change of about six to eight kilometres per hour. The medical records revealed multiple episodes of low back pain with radiation into the left leg following lifting activates both before and after the collision. The forces across the L4/5 vertebral joint were calculated for the collision and lifting tasks. The biomechanical analysis showed the published tolerance for this injury was greater than the force applied during the collision, but about equal to the force applied multiple times during the lifting tasks.

A bicyclist participating in a road race struck an unpadded lamp standard and sustained a fatal neck injury. A biomechanical analysis showed that for the speed at which the bicyclist struck the lamp standard, the neck loads were sufficient to cause the fracture even if the lamp standard had been padded. It was further shown that padding can increase the potential for pocketing or trapping the head, which increases rather than reduces the likelihood of neck injury.

A man carrying a cardboard box down wet stairs outside a residential apartment building slipped fell down the remaining stairs and suffered a fracture/dislocation of his left ankle. Friction tests performed under both wet and dry conditions showed the available coefficient of friction was adequate for normal stair descent. The type of ankle fracture/dislocation was more consistent with slipping off the stair nosing due to poor foot placement than slipping on the stair tread because of low friction. Since the stair otherwise complied with local building codes, the injury was attributed to a misstep rather than a stair design or construction problem.

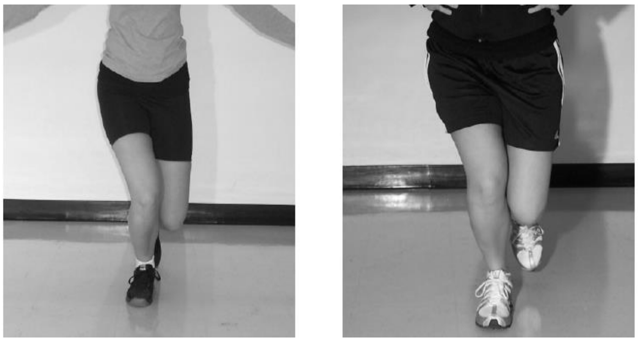
Other types of motor vehicle cases include seatbelt use and effectiveness, driver identification, motorcycle or bicycle helmet use and effectiveness analyses, and whiplash injury potential. In slip, trip and fall incidents, injury analyses can be used to define the plaintiff's pre-event behaviour, quantify the level of friction needed for specific tasks, and quantify reaction times and recovery strategies to unexpected mishaps. Injuries caused during sport or recreational activities can also be analyzed to assess the use and effectiveness of safety equipment (helmets, padding, etc.) and determine whether arena facilities or product failures contributed to the injuries. Miscellaneous cases involving assaults, police shootings, gait analyses and the limits of human performance can also benefit from the principles used for injury biomechanics analyses.

INJURY PREVENTION

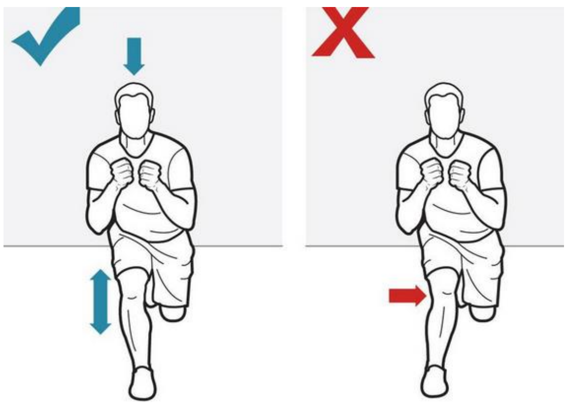
One of the most common injuries in sports, a non-contact sprain of the ACL. The large numbers of injuries to young female athletes has resulted in considerable research on how these injuries occur in landing, jumping, and cutting. Manybiomechanical factors have been hypothesized to be related to increased risk of ACL injuries in sport: peak vertical ground reaction force, knee flexion angle at landing, hamstring strength, and balance. A large prospective study of the biomechanics of landing in female adolescent athletes who then participatedin high-risk sports has recently identified several variables that are associated with risk of ACL injury.

The variables that were associated with girls that became injured were greater knee abduction angle (lower leg valgus), and greater ground reaction force and knee abduction moment. It is possible that as girls enter adolescence the increased risk of ACL injuries comes from dynamic valgus loading at the knee that results from a combination of factors. With adolescence in females the limbs get longer and hips widen, if strength at the hip and knee, coordination, and balance do not keep up with these maturational changes it is likely thatrisk of ACL injury could be increased.

While sports medicine professionals have qualitatively evaluated the strength and balance of patients in single leg stance and squats for many years, recently have proposed that simple two-dimensionalmeasurements of frontal plane motion of the lower extremity in single leg squats might be a useful clinical tool for identifying athletes that may be at a higher risk for ACL injury.While this test is not as dynamic as landing, it is likely a safer screening procedure that also can be qualitatively evaluated. If screening suggests an athlete may be at risk (poor control of knee in the frontal plane), research has shown that preventative conditioning programs can decrease the risk of ACL injuries.



In the above figure, the position of the lower extremity at the bottom of a single leg squat for two young athletes. If you were an athletic trainer or physical therapist screening these athletes before a competitive season, which athlete would you be most concerned about for a higher risk of ACL injury?



Right: Single leg squat in good form. The model is flexing his hip, knee, and ankle equally and keeping his knee forward and his leg perpendicular with the ground. This will strengthen leg, promote good ski technique, and teach him to avoid a dangerous position.

Wrong: Single leg squat in poor form. The model is letting pelvis drop and his standing knee fall inwards. This is a component of the "phantom foot" position and should be avoided.

7. Differentiate between intrinsic and extrinsic factors related to injury development.(N/D 2016)

Intrinsic and extrinsic risk factors

Intrinsic factors are variables that you are able to control to prevent yourself from an injury whereas extrinsic factors are variables that you are unable to control to prevent yourself from an injury.

Variables fit under the intrinsic factors category

There are six different variables that fit into the intrinsic factors category. They are:

– Flexibility and joint laxity

– Nutrition

– Leg length discrepancies

– Fitness levels

– Age

– Weight/size

Variables fit under the extrinsic factors category

There are five main categories of extrinsic risk factors. They are:

– Coaching

– Incorrect techniques

– Environmental factors

– Clothing, footwear and equipment

– Safety hazards

Both the intrinsic and extrinsic risk factors have a relation to sport because both can cause injuries.

Intrinsic:

– Flexibility and joint laxity can cause a sporting injury by trying to over stretch a muscle or trying to perform a move which requires the joints that are not attached together very well.

– Nutrition can cause a sporting inury by a player not having enough energy and maybe making silly mistakes that could lead to eitherthem or someone else picking up an injury, all down to them not having the correct nutrition.

– Leg length discrepancy can cause a sporting injury one leg is longer than the other, this could change our posture but it could also change the forces going through the muscles and joints, by there being different forces someone could go in for a fifty-fifty collision in football and come out a lot worse as they have a unusual posture and have different forces between their muscles and joints.

– Fitness levels can cause a sporting injury because if someone has low fitness levels, they will fatigue a lot quicker than if they had higher fitness levels, with them becoming tired very quickly, they will make more mistakes which could lead to an injury.

– Age could cause a sporting injury as an individual gets older due to their bone strength, you wouldn't have a seventy year old playing rugby against a load of middle aged men because the chances of the older individual coming out of the game with an acute injury due to the force of impact being too much for their bones to handle.

– Weight/size can cause a sporting injury because if an individual is overweight they have additional stress put on their muscles, tendons and organs, therefore even small movements can create large forces. With little movement creating a large force a player could get seriously injury when going in for a fifty-fifty challenge in a football game, the player who weighs less will not have as much force to be put in as the other player and will come out with an injury, the other player could also come out with an injury as they have maybe put too much force into the tackle.

Extrinsic

– Coaching is linked with incorrect techniques as you require a coach who knows about the sport meaning that they can teach you the correct techniques, because if you had a coach who has no or minimal knowledge of the sport then they could teach you the wrong techniques from a young age which could then become difficult to adjust to the correct technique.

– Incorrect techniques can cause sporting injuries and not only could you get an acute injury but you could also be building up on a chronic injury. An example of how you could suffer from an acute injury by not having the correct technique would be in rugby, players are always diving in at each other, making tackles and getting hurt, if you have the incorrect technique and tackle with your head too high or too low, you could suffer from a very serious injury due to you having the wrong technique. An example of how an individual may build on the development of a chronic injury would be that a football has the wrong technique on how to pass a ball, in a football match you pass the ball quite a lot and with them having a repetitive action of the incorrect technique they could be causing local tissue damage, which could be a severe injury when the injury has fully developed into its worse state for the individual (when the individual notices that he is sufferring from a tissue injury).

– Environmental factors can cause sports injuries as an individual could only ever train in the sun and on a hard pitch but say it came to game day and it is raining with a moggy pitch, or snowing, this individual is not going to be use to the conditions meaning that they could cause the individaul to act differently, which could could injury, or the individual could perform the same as they usually do which could as well as positive have a negative impact on the individuals performance and them causing a sporting injury for themselves.

– Clothing, footwear and equipment can cause sporting injuries, you must be wearing the correct clothing and footwear to have the minimum risk of injury. An example of wearing incorrect footwear would be when a female (or male) wears high heels to go into the gym to do dead-lifts. By them doing the dead lifts they have very little balance due to the heels and are very likely to lose balance at any point, maybe break the heels and have a severe injury of maybe dislocating or breaking a bone, just because they were not wearing the correct footwear.

Protective clothing is used to protect you and aid is decreasing the risk of injury, meaning that it should be worn! An example of protective clothing would be a helmet in racing sports, the helmet is there to help reduce the risk of causing brain damage and/or damaging the skull in the event of a crash.

Equipment should always be checked before participating in a sport, an example of needing to check the equipment could be in football, the goals must be properly looked at by the ref and he must then determine whether they are safe and properly fixed. If the goal posts were not checked before a game, they could fall down and land on either the goalkeeper or any other person that may be near to them at the time, causing a serious injury which could have easily been prevented.

– Safety hazards are placed to help try and reduce the risk of injury, a risk assessment must be taken before activity on a playing field because there could be anything on the pitch that could cause injury to an individual, if the risk assessment is not taken then as already stated, a player could go in for a slide tackle and yet come off with a very large, deep cut down their leg as they could have gone sliding across grass or anything else with a sharp edge, this is why a risk assessment must be done and safety hazards must be put into place.

8. List the limitations of finite element modeling. (N/D 2015)

The following are the steps adopted for analyzing a structural engineering problem by the finite element method.

1. Discretization of the domain

The continuum is divided into a number of finite elements by imaginary linesor surfaces. The interconnected elements may have different sizes and shapes. Thechoice of the simple elements or higher order element straight or curved, it's shape,refinement are to be decided before the mathematics formulation starts.

2. Identification of variables

The elements are assumed to be connected at their intersecting points referredto as nodal points. At each node, generalized displacements are the unknown degreesof freedom. They are dependent on the problem at hand. For example in a plane stressproblem the unknowns are two linear translations at each nodal point.

3. Choice of approximating functions

Once the variables and local coordinate system have been chosen. The nextstep is the choice of displacement function. In fact it is the displacement function thatis the starting point of the mathematical analysis. This function represents thevariation of the displacements within the element. The function can be approximated65in a number of ways. The displacement function may be approximated in the form ofa linear function or a higher order function. The shape of element or the geometrymay also be approximated. The co ordinates of corner nodes define the element shapeaccurately if the element is actually made of straight line or plates.

4. Formation of the element stiffness matrix

After the continuum is discritised with desired element shapes, the elementstiffness matrix is formulated. This can be done in a number of ways. Basically it is aminimization procedure whatever may be the approach adopted. For certain elements,the form involves a great deal of sophistication. With the exception of a few simpleelements, the element stiffness matrix for majority of elements is not available inexplicit form. As such they require numerical integration for their evaluation.The geometry of the element is defined in reference to a global frame. In manyproblems such as those of rectangular plates, the global and local axis systems arecoincident and for them no further calculation is needed at the element level beyondcomputation of element stiffness matrix in local coordinates. Coordinatestransformation must be done for all elements where in is needed.

5. Formulation of the overall stiffness matrix

After the element stiffness matrices in the global coordinates are formed, theyare assembled to form the overall stiffness matrix. The assembly is done through thenodes, which are common to adjacent elements. At the nodes, the continuity of thedisplacement function and possibly their derivatives are established. The overallstiffness matrix is symmetric and banded.

6. Incorporation of boundary conditions

The boundary restraint conditions are to be imposed in the stiffness matrix.There are various techniques available to satisfy the boundary conditions. In some of66these approaches, the size of the stiffness matrix may be reduced or condensed in itsfinal form. To ease the computer programming aspect and to elegantly incorporate theboundary conditions, the size of the overall stiffness matrix is kept the same.

7. Formulation of element load matrix

The loading forms an essential parameter in many structural engineeringproblems. The loading inside the element is transferred at the nodal points andconsistent element load matrix is formed. Sometimes, based on the typicality ofproblem, the load matrix may be simplified.

8. Formation of the overall load matrix

Like the overall stiffness matrix, the element loading matrices are assembledto form the overall loading matrix. This matrix has one column per loading case and itis either a column vector or a rectangular matrix depending on the number of loadingconditions.

9. Solution of simultaneous conditions

All the equations required for the solution of the problem are now developed.In the displacement method, the unknowns are the nodal displacements. The gausselimination and Cholesky's factorization are the most commonly used procedures forthe solution of simultaneous equations. These methods are well suited to a small ormoderate number of equations. For large sized problems, a frontal technique is one ofthe methods of obtaining solution. For systems of large order, Gauss-Seidel or Jacobiiterations are more suited.

10. Calculation of stress or stress-resultants

In the previous step, nodal displacements are calculated and these values areutilized for the calculation of stresses or stress-resultants. This may be done for allelements of the continuum or it may be limited only to some predetermined elements.Results may be obtained by graphical means. It may be desirable to plot thecontour of the deformed shape of the continuum. The contour of the principal stressesmay be one of the sought after items for certain category of problems.

9. Describe the methods used to achieve the goals of sport and exercise in biomechanics. (M/J 2016)

- **The major goal of biomechanics of sport and physical exercise is to improve performance in given sport or physical exercise.**
  In a wider context the goal of biomechanics of sport and physical exercise is also to increase physical fitness. For instance the correct biomechanics of running allows athletes to carry out regular physical exercise for long enough periods of time without being seriously limited by injuries and their consequences.
- **The secondary goal of sport biomechanics is to provide recommendations for injury prevention and rehabilitation.**
  The secondary goal of sport biomechanics is strongly related to the main goal because a healthy athlete will perform better than an athlete plagued with frequent injuries.

Performance Improvement

Technique Improvement

In many sporting events technique is the major factor of performance. Martens (2004) defines sport technique as follows:

Sport technique is a physical action of an athlete which leads to the best possible execution of a physical motion, in conformity with a required task of a given sporting event.

Improvement of technique with the help of biomechanics can be used by teachers and coaches to correct motions of students or athletes. Moreover, research workers in the field of biomechanics may develop a new and more effective technique for better execution of a sport motion. In the former case teachers and coaches make use of the methods of qualitative biomechanics analysis in their every day practice to produce changes in the technique used by their charges. In the latter case research workers in the field of biomechanics use quantitative biomechanics methods to develop new techniques which can then be implemented into teaching and training processes.

For instance if a gymnastics coach sees that her charge has difficulties to turn a somersault she can come up with three recommendations to help the gymnast execute this exercise correctly: 1. to jump higher, 2. to fling arms with more energy before taking off, or 3. to curl up more tightly. All these recommendations can help to execute this task correctly and are based on the principles of biomechanics. If the gymnast jumps higher, she has more time to finish the turn during the flight phase. To curl up more tightly means to increase the speed of rotation while keeping the same angular momentum. To fling arms with more energy increases the angular momentum which helps the gymnast to rotate faster.

Equipment Improvement

Use of biomechanics can also lead to a better look and better functioning of sport equipment. For example ski boots can have a real impact on sport performance. Sophisticated sport equipment gives advantage to both elite and recreational athletes.

Artistic gymnastics offers good examples. An introduction of the new vaulting equipment (vaulting table) after the 2000 Olympics represents the most substantial transition in the development of gymnastics equipment in the last decades. New vaulting equipment allows gymnasts to produce bigger angular momentum and thus to execute more complex vaults with multiple rotations around horizontal and vertical axes (Farana and Vaverka, 2010).

Researchers have recently also developed a new swimming suit which helped swimmers at the Sydney Olympics in 2000 better several world records because it has a favourable influence on the draft force and buoyancy of water that is acting against swimmers. This swimming suit had such an influence on sport performance in swimming, in fact, that its use was later banned.

Training Improvement

Biomechanics can help improve training of athletes in two ways:

- **By the analysis of mechanical values a coach defines such training conditions that may lead to threshold stimuli.**

We can use as an example the research project by Jandacka and Uchytil (2011) who carried out mechanical analysis of bench press with various loads in elite footballers. They discovered that the use of a load equalling to 30 – 50% of the load which the footballers were only able to lift once leads to maximal produced mechanical power output. The recommendation resulting from this research project was as follows: Soccer players should train maximal strength during the preparatory period for their competitive season along with training for speed and endurance. When athletes maximally develop muscular power toward the end of a season when the most important competitions are scheduled, dynamic effort strength training with loads from 30 to 50% of 1RM BP should be used. During the competitive season, loads of 50% of 1RM BP should be used to maintain muscular power over a wide load range.

- **By the analysis of technical imperfections of a given athlete the coach/teacher identifies the type of training needed for this athlete to improve.**

An athlete is limited by strength or endurance of certain muscle groups, by speed of motion, or by specific aspects of motion technique. Sometimes the limits are quite obvious. For example a gymnast executing the crucifix on the gymnastic rings must have very strong shoulder adductors. In the case of certain sport skills the required abilities to execute a motor task are not easy to detect and quantitative biomechanics analysis must be used.

Injury Prevention

By injury prevention it is meant an attempt to prevent or to limit the seriousness of injuries before they are actually incurred.

The concept of injury prevention is part of public health and its goal is to improve the general health of the population and thus to increase the quality of life. Biomechanics is a tool that can be used in sport medicine to identify forces and mechanical energy that cause injuries. It helps to understand how injuries originate, how to avoid them during sport performance, and how to identify exercise suitable for injury prevention and rehabilitation. Biomechanics offers possibilities to create alternative techniques of executing specific movements, using new equipment, and carrying out more effective training methods, which also contributes to injury prevention.

Good examples of how biomechanics helps reduce the prevalence of injuries can be found in volleyball. Zahradník and Jandacka (2011) examined whether it is possible to adapt the landing after a volleyball blocking to reduce impact reaction forces acting on knee joints. They found that it is better for volleyball players to make one step back after blocking as opposed to staying on the landing spot and absorbing the relevant forces there.

Another interesting result of biomechanics analysis with the purpose of injury prevention was the causation study of the so-called iliotibial band syndrome. Research workers Hamill, Miller, Noehren, Davis (2008) allege that the lateral knee pain, typical for many distance runners who train regularly, represents roughly 12% of all injuries in runners. They also discovered that this syndrome may be caused by increased hip abduction and knee internal rotation during the stance phase, which causes strain in iliotibial band. From a large prospective study, female runners who incurred iliotibial band syndrome during the study were compared to a control group who incurred no injuries. Strain, strain rate and duration of impingement were determined from a musculoskeletal model of the lower extremity. This study indicated that a major factor in the development of iliotibial band syndrome was the strain rate. Therefore, Hamill et al. (2008) suggested that strain rate, rather than the magnitude of strain, could be a causative factor in developing the iliotibial band syndrome.

Injury prevention and rehabilitation are currently among very important goals of research in the field of biomechanics of sport and physical exercise.

Injury reduction through changes to equipment function

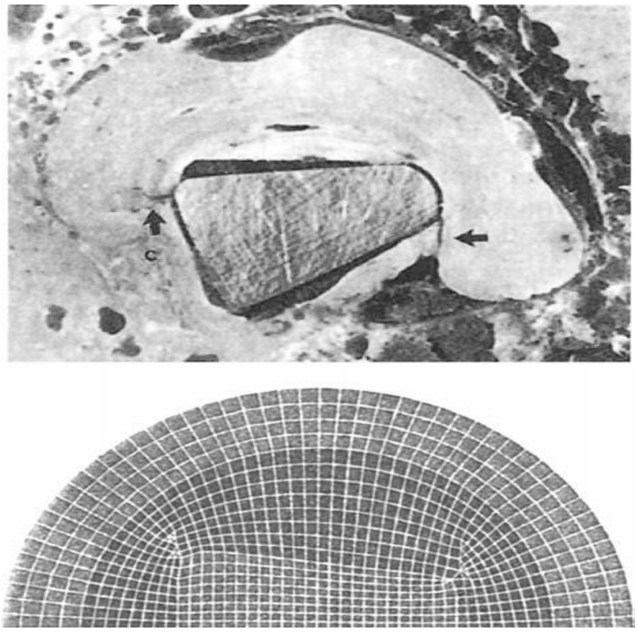
One of the examples of using the results of biomechanics research for improving the functioning of sport equipment can be found in running. The number of people who realize the importance of healthy life style is recently growing. Running, as an elementary human locomotion, is a legitimate part of healthy lifestyle. But the growing numbers of people engaged in running also brought higher prevalence of injuries. Running shoes at the beginning of the 1970s were too stiff for inexperienced runners. Among the injuries with growing prevalence were stress fractures and shin bone pain. Shoe manufacturers therefore started to market shoes with soft soles. However, soft soles did not offer good stability and motor control. Runners started to suffer from ankle, knee and hip injuries. Biomechanics research has made it possible to manufacture running shoes which reduce impact forces and, at the same time, offer good stability and motor control. With the help of biomechanics it is even possible to recommend custom made shoes for individual athletes. Prevalence of injuries in running has decreased again.

Isn't human body itself the best equipment for running? People who wear shoes from very early age mostly touch the ground first with their rear foot when they walk. Lieberman et al. (2010) studied the style of running in Kenyans who never wore shoes and assert that in barefoot running people naturally touch the ground first with their forefoot. This produces slower loading rate in foot compared to running in shoes and touching the ground first with rear foot. Grand reaction forces during running may cause chronic injuries that runners often suffer from.

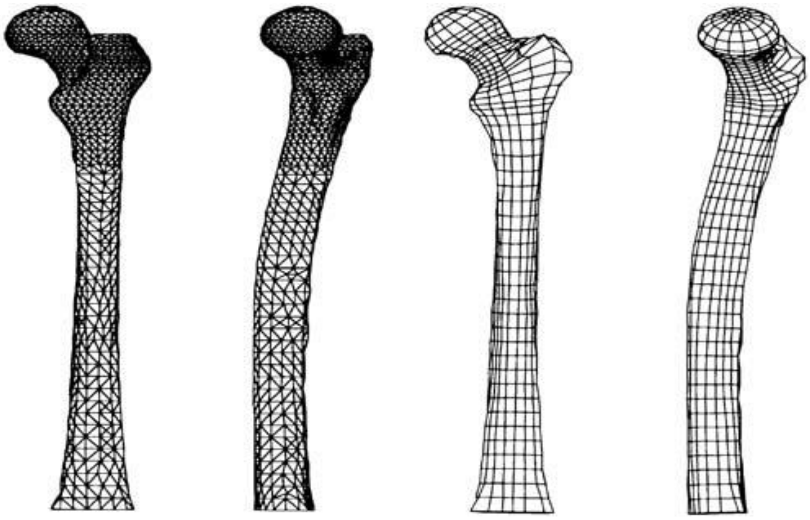
10.Explain overview of the finite element method

The essential steps in implementing the FEM follow:

(i) The region of interest (continuum) is discredited, that is, subdivided into a smaller number of regions called elements, interconnected at nodal points. Nodes may also be placed in the interior of an element. In one dimension, the elements are line segments; in two dimensions, they are usually triangles or quadrilaterals, in three dimensions, they can be rectangular prisms (hexahedra) or triangular prisms (tetrahedral), for example Elements may be quite general with the possibility of non-planar faces and curvilinear sides or edges.

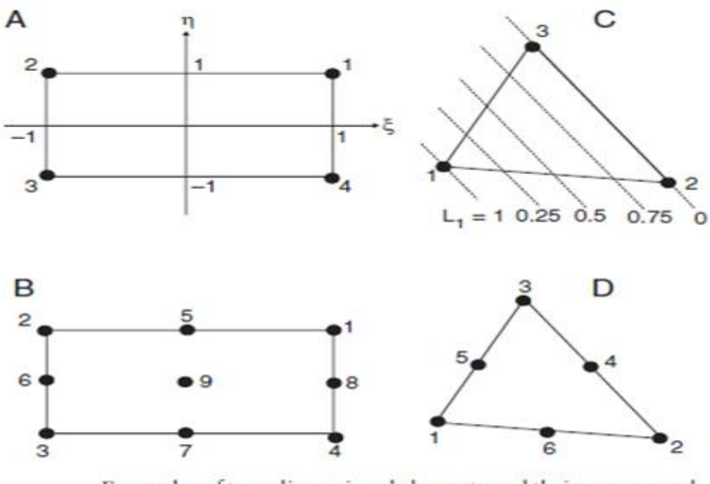


Cross-section of an autopsy-retrieved femur showing a cracked mantle (arrows).Mixed planar quadrilateral/triangle FE representation.



3D FE representations of the human femur.Tetrahedral elements and hexahedral elements.

(ii) The unknown variables within the continuum (e.g., displacement, stress, or velocity components) are defined within each element by suitable interpolating functions. Interpolating functions are traditionally piecewise polynomials and are also known as basis or shape functions. The order of the interpolating functions (i.e., first, second, or third order) is usually used to fix the number of nodes in the elements.



Examples of two-dimensional elements and their corresponding local coordinate systems [embedded in (A) and (C)]. For the rectangles, the local coordinates (ξ, η) are referred to a cartesian system with $-1 \le \xi, \eta \le 1$; for the triangles, the local coordinates (L_1, L_2, L_3) are area coordinates satisfying $0 \le L_1, L_2, L_3 \le 1$. Elements with linear interpolating functions (first order) are shown in (A) and (C). Quadratic elements (second-order interpolating functions) are shown in (B) and (D).

(iii) The equations that define the behavior of the unknown variable, such as the equations of motion or the relationships between stress and strain or strain and displacement, are formulated for each element in the form of matrices. These element matrices are then assembled into a global system of equations for the entire discretized domain. This system is defined by a coefficient matrix, an unknown vector of nodal values, and a known "right-hand side" (RHS) vector. Boundary conditions in derivative form would already be included in the RHS vector at this stage, but those that set the unknown function to a known value at the boundary have to be incorporated into the system matrix and RHS vector by overwriting relevant rows and columns. Since the RHS vector contains information about the boundary conditions, it is sometimes called the "external load vector."

(iv) The final step in FEA involves solving the global system of equations for the unknown vector. In theory, this can be achieved by premultiplying the RHS vector by the inverse of the coefficient matrix. The result is the discrete (pointwise) solution to the original problem. If the problem is linear and isotropic, the elements of the matrix are constants and the required matrix inversion can be done. If the defining equations are nonlinear or the material is anisotropic, the coefficient matrix itself will be a function of the unknown variables and matrix inversion is not straightforward. Some kind of linearization is necessary before the matrix can be inverted (e.g., successive approximation or Newton's methods). In practice, the global system matrix, whether linear or nonlinear, is seldom inverted directly, usually because it is too large. Some indirect method of solving the system of equations is preferred [i.e., lowerupper (LU) decomposition, Gaussian elimination.

The evaluation of element matrices, their assembly into the global system, and the possible linearization and eventual solution of the global system is a task that is always passed on to a high-speed computer. This usually requires complex computer programs written in a high-level language, such as FORTRAN. Indeed, it is the advent of high-speed computers and workstations and the continuous improvements in processor speed, memory management, and disk storage that have enabled large-scale FE problems to be tackled with relative ease.

The modern-day FEA toolbox also includes facilities for data pre- and post-processing. Data preprocessing usually involves input formatting and grid definition, the latter of which may require some ingenuity, because mesh design may affect the convergence and accuracy of the numerical solution.

The result is usually a dramatic improvement in convergence, accuracy, and computational efficiency. Post processing of data involves the evaluation of ad hoc variables such as strains, strain rates, stresses; generating plots such as simple xy-plots, contour plots, and particle paths; and solution visualization and animation. All of the additional information facilitates the understanding and interpretation of the results.

The importance of checking and validating FE solutions cannot be overemphasized. The most basic validation involves a “patch test” in which a few elements (i.e., a patch of the material) are analyzed to verify the formulation of interpolating functions and the consistency of the code itself. Second, a very simple problem with known analytical solution is simulated with a coarse grid to verify that the code reproduces the known solution with acceptable accuracy.

For example, parabolic flow in a tube can be simulated with a very coarse grid and the result quickly compared against the analytical solution. We caution, however, that reproducing the solution in a simpler problem does not guarantee that the code will work in more realistic and complicated cases. It is also recommended that numerical solutions be obtained from at least three meshes with increasing degrees of mesh refinement such solutions should converge with mesh refinement. Comparison of numerical results to experimental data should always be made where possible. Last, especially in the absence of analytical solutions or experimental data, numerical solutions should be compared across different numerical methods or across different numerical codes if the same method is used. There is no gold standard for the number of validation tests that is required for any particular problem. The greater the variety of test problems and checks, the greater the degree of confidence one can have in the results of the finite element method.