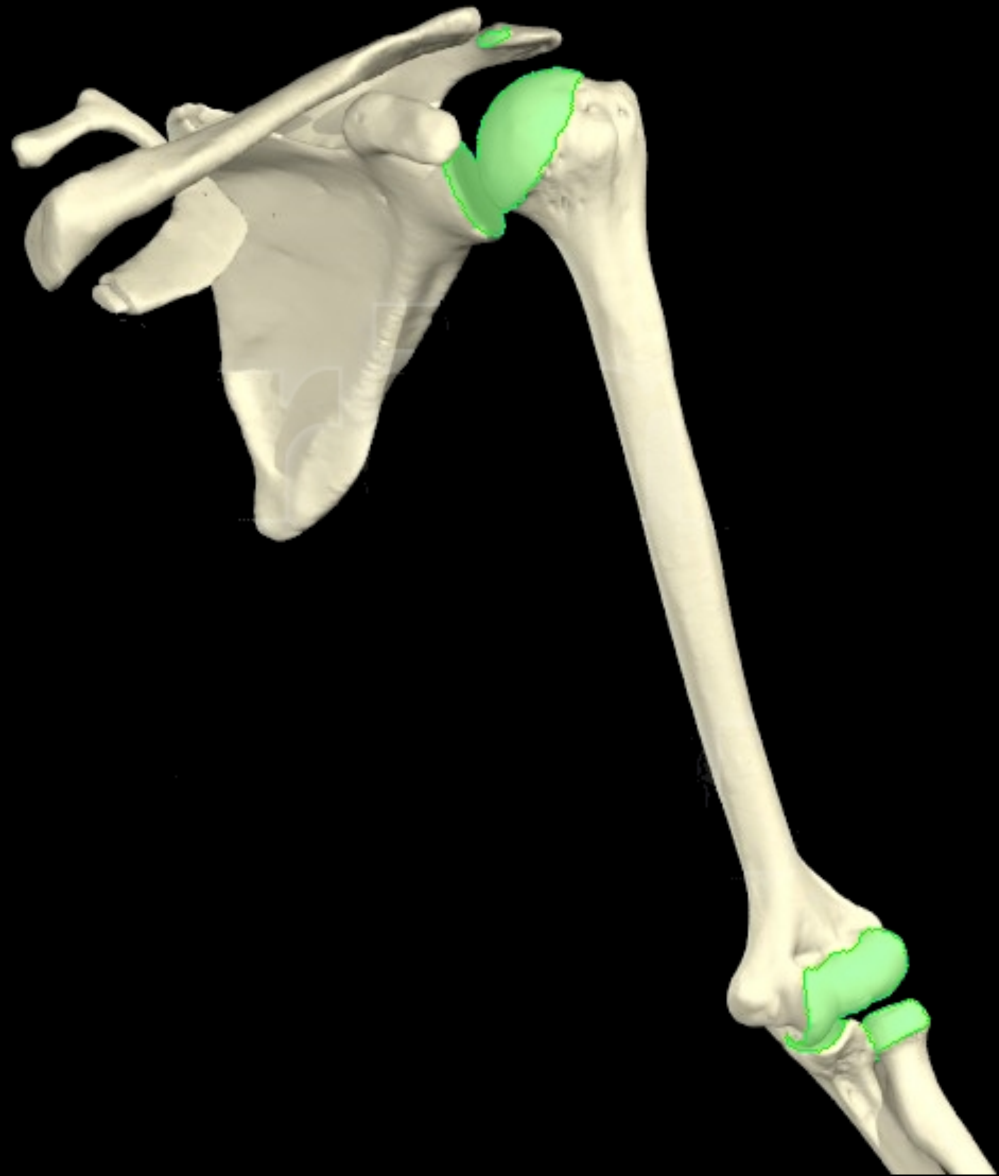
Biomechanics of Articular Cartilage

Christos K. Yiannakopoulos

Articular Cartilage

- Important to understand
  - Structure and mechanical properties of normal cartilage
  - Manner by which biochemical and structural factors contribute to the material properties of cartilage
  - Manner by which changes in tissue composition affect the mechanical properties of cartilage

Articular cartilage covers bone surfaces within the joint capsule



Functions of Articular Cartilage

- Distributes joint loads over a wide area, decreasing the stresses sustained by the contacting joint surfaces
- Allows relative movement of the opposing joint surfaces with minimal friction and wear
- Minimizes peak stresses on subchondral bone

Functions of Articular Cartilage

- Surfaces roll or slide during motion
- Provides a friction-reducing, weight-bearing surface with a friction coefficient of 0.0025
- Functions within a contact pressure range of 2-11 MPa

Functions of Articular Cartilage

despite common belief does not serve as a “shock absorber”

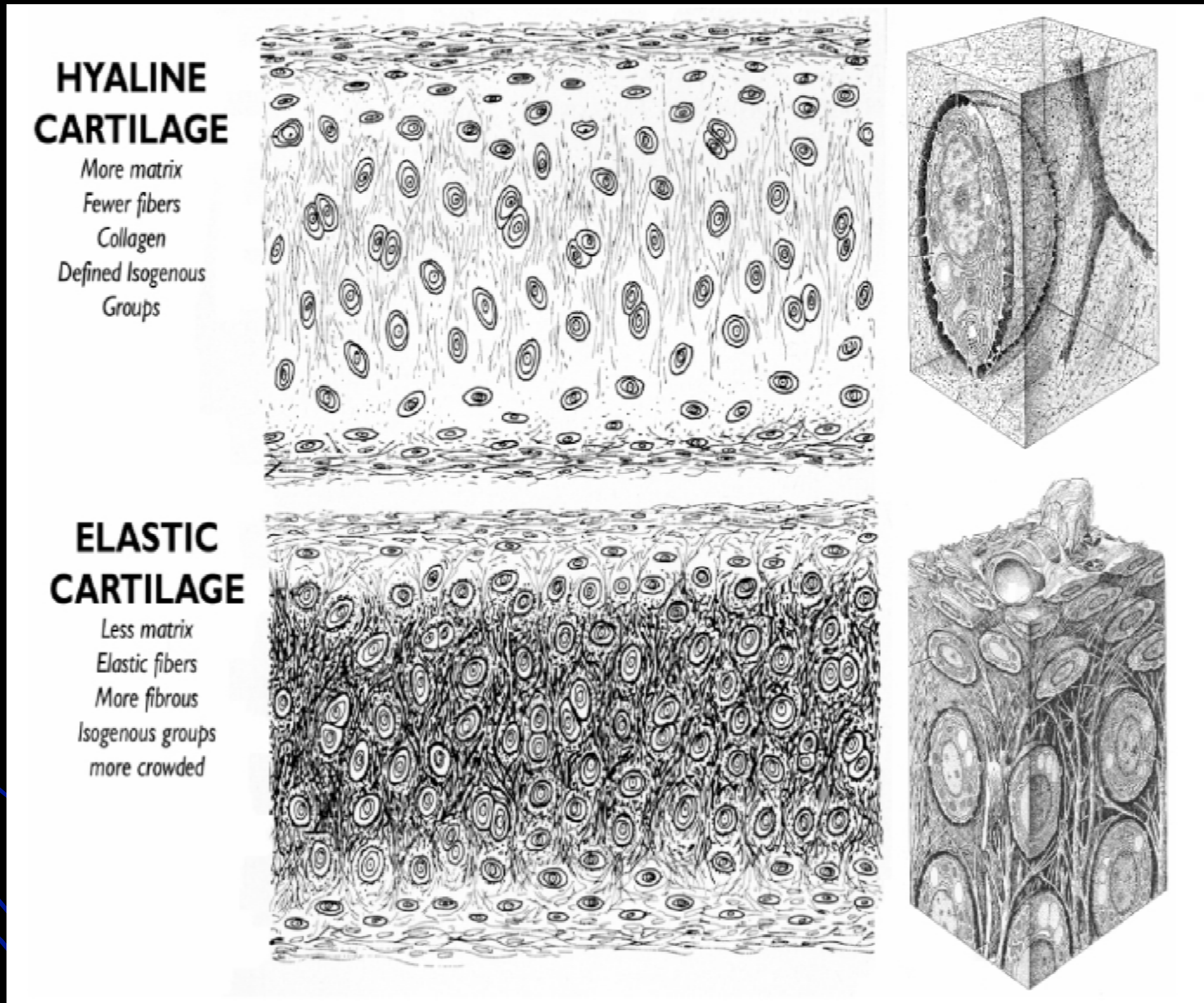
- very thin
- capacity negligible compared to muscles and bones

Normal Cartilage

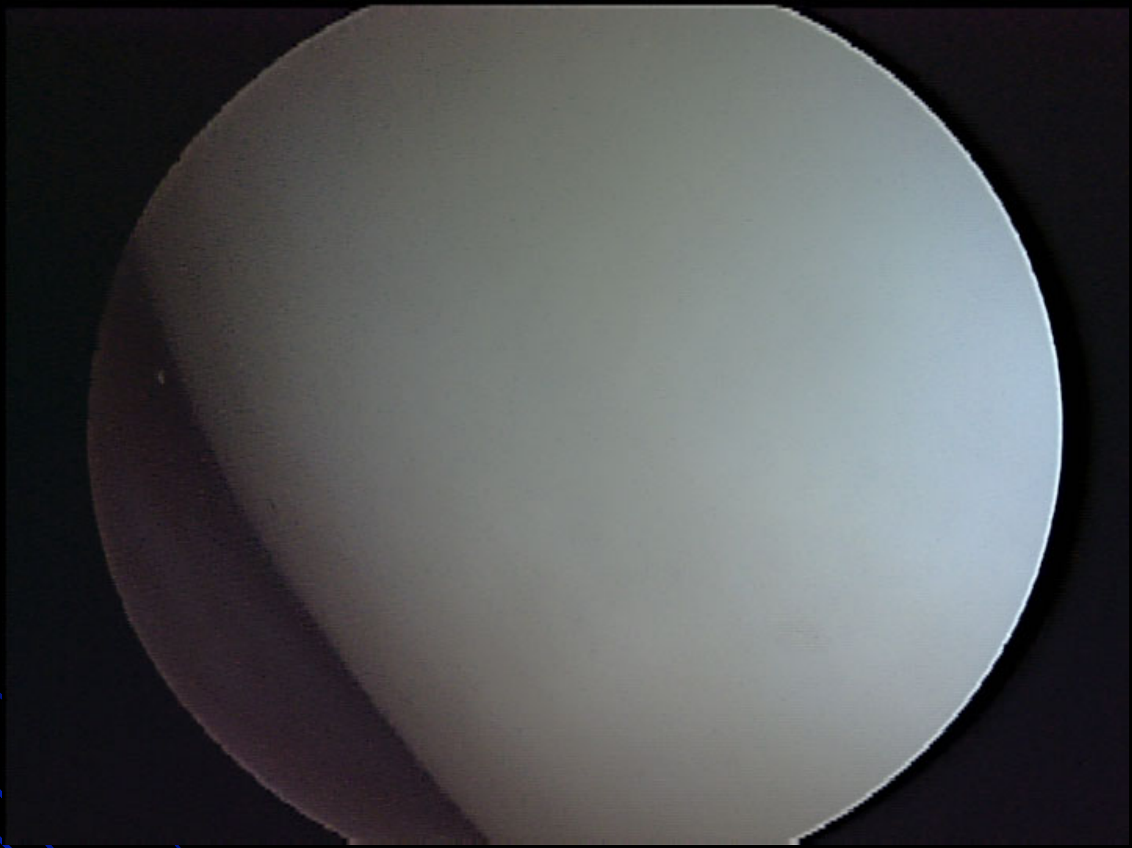
- Avascular, alymphatic and aneural tissue

Types of Cartilage

Type	Appearance	Location
Hyaline	Glassy, smooth	Covers long bones, growth plates
Fibro	Dense	Intervertebral disks, meniscus
Elastic	Yellow, opaque	Epiglottis, eustachian tube


HYALINE CARTILAGE
More matrix
Fewer fibers
Collagen
Defined Isogenous Groups
ELASTIC CARTILAGE
Less matrix
Elastic fibers
More fibrous
Isogenous groups more crowded

Articular Cartilage



The thickness of articular cartilage

- varies with the particular joint,
  and the location within the joint.
- ranges from 0.5 mm (rabbit knee)
  to 10.0 mm (pf groove of bovine knee).
- Humans: thickest over ends of femur and tibia
  (range 2-4 mm).

Hyaline cartilage

a multiphasic material with two major phases:

- fluid phase: water (68-85%) and electrolytes
- solid phase: type II collagen(10-20%), proteoglycans

and other glycoproteins (5-10%), chondrocytes

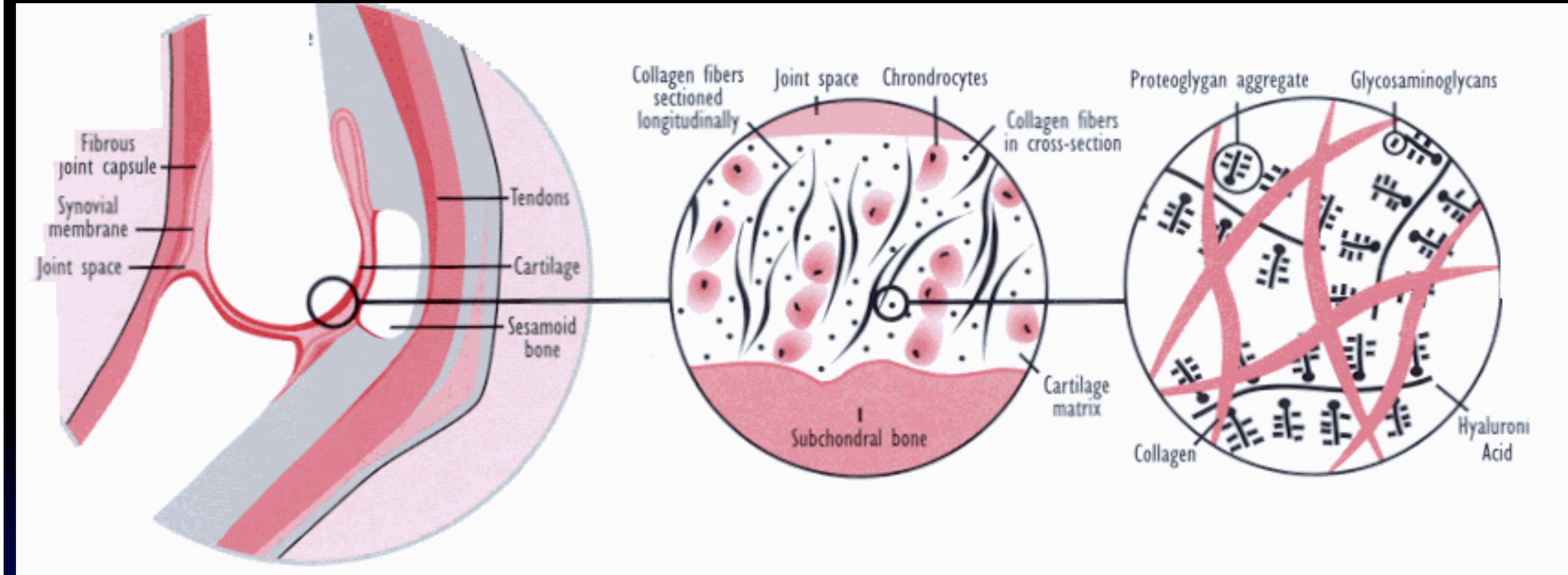
Biochemical Composition

composite material with widely differing properties.

- Chondrocytes (<10%)
- Extracellular matrix
  - Proteoglycan (monomer and aggregates) (3-10%)
  - Collagen (10-30%)
  - Water + Electrolyte (60-87%)

Tissue	Water	Collagen	Proteoglycans
Articular Cartilage	68-85%	10-20% (type I)	5-10%
Meniscus	60-70%	15-25% (type II)	1-2%

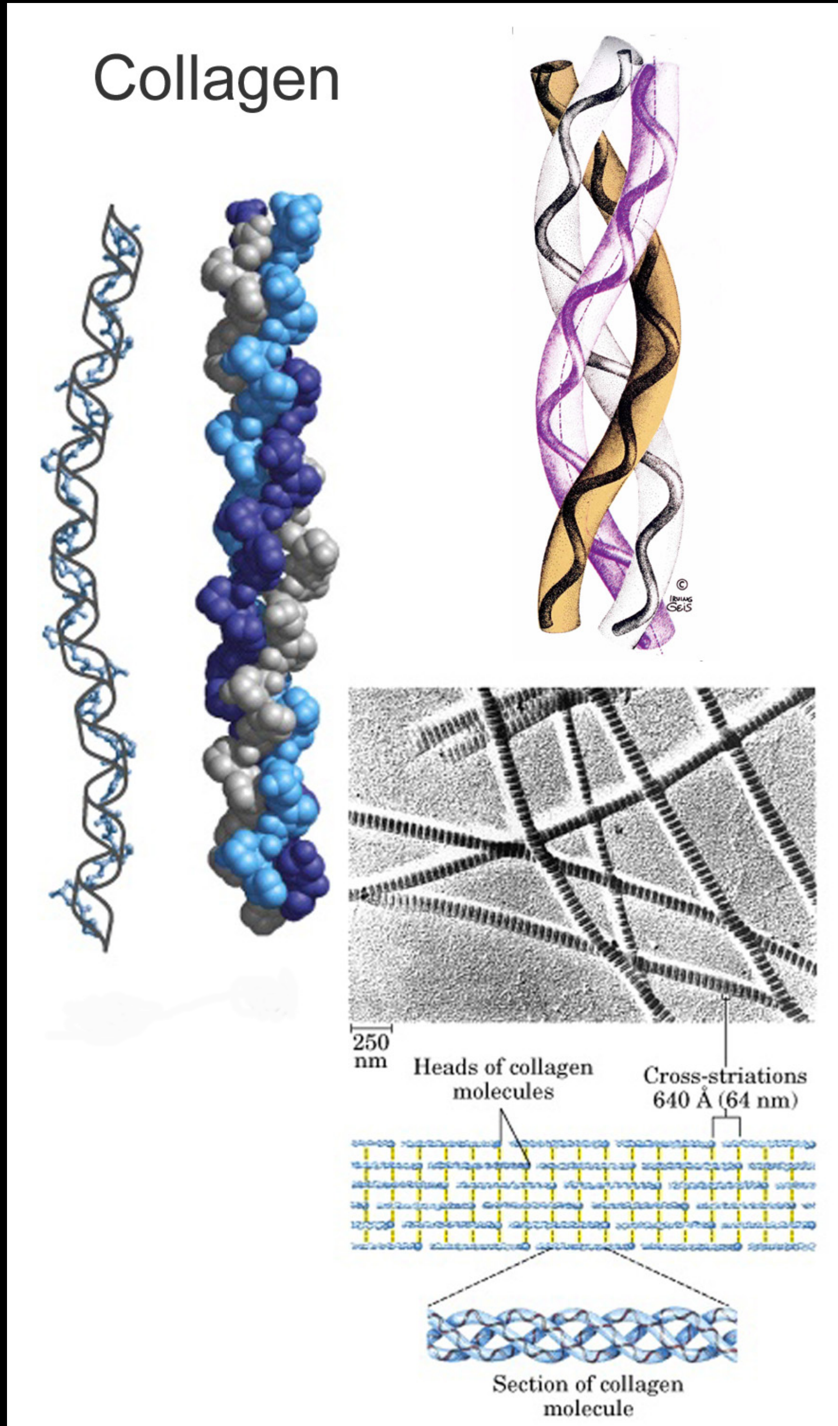
The material properties of articular cartilage
depend on its extracellular matrix,
but the existence and maintenance of the matrix
depend on the chondrocytes.


Fibrous
joint capsule
Synovial
membrane
Joint space
Tendons
Cartilage
Sesamoid
bone
Collagen fibers
sectioned
longitudinally
Joint space
Chrondrocytes
Collagen fibers
in cross-section
Cartilage
matrix
Subchondral bone
Proteoglygan aggregate
Glycosaminoglycans
Hyaluroni
Acid
Collagen

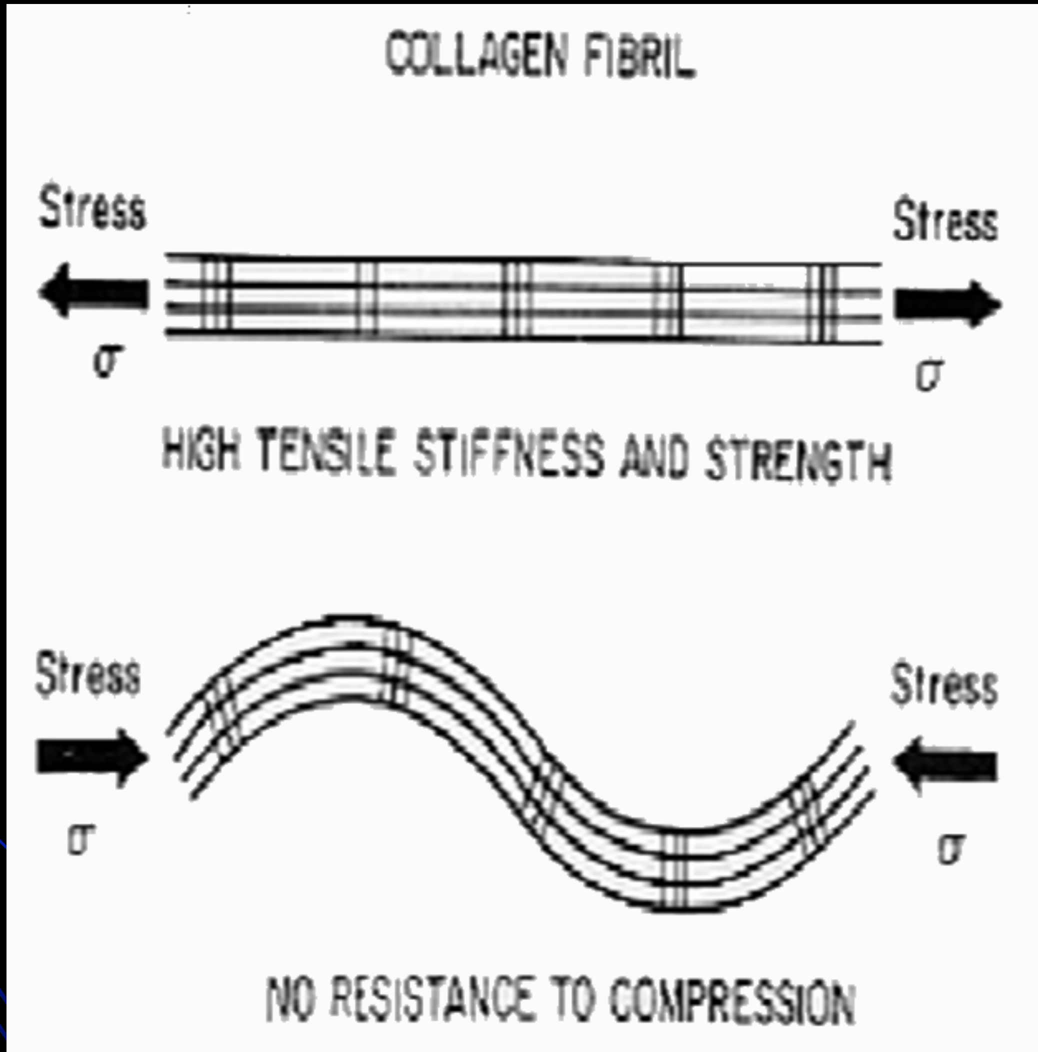
Collagen

Creates a framework that houses the other components of cartilage

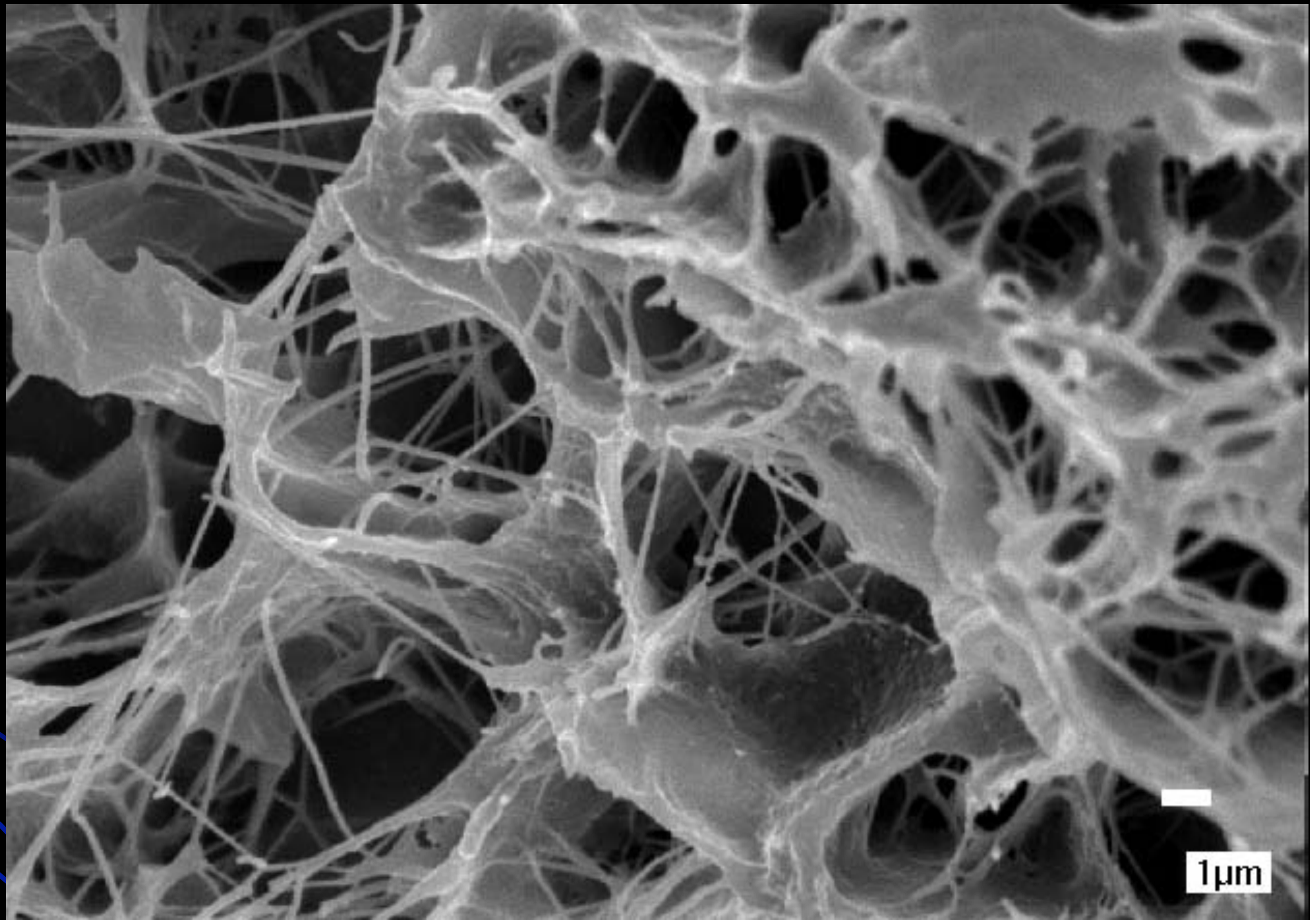
- Majority is Type II collagen
- Provides cartilage with its tensile strength



Collagen

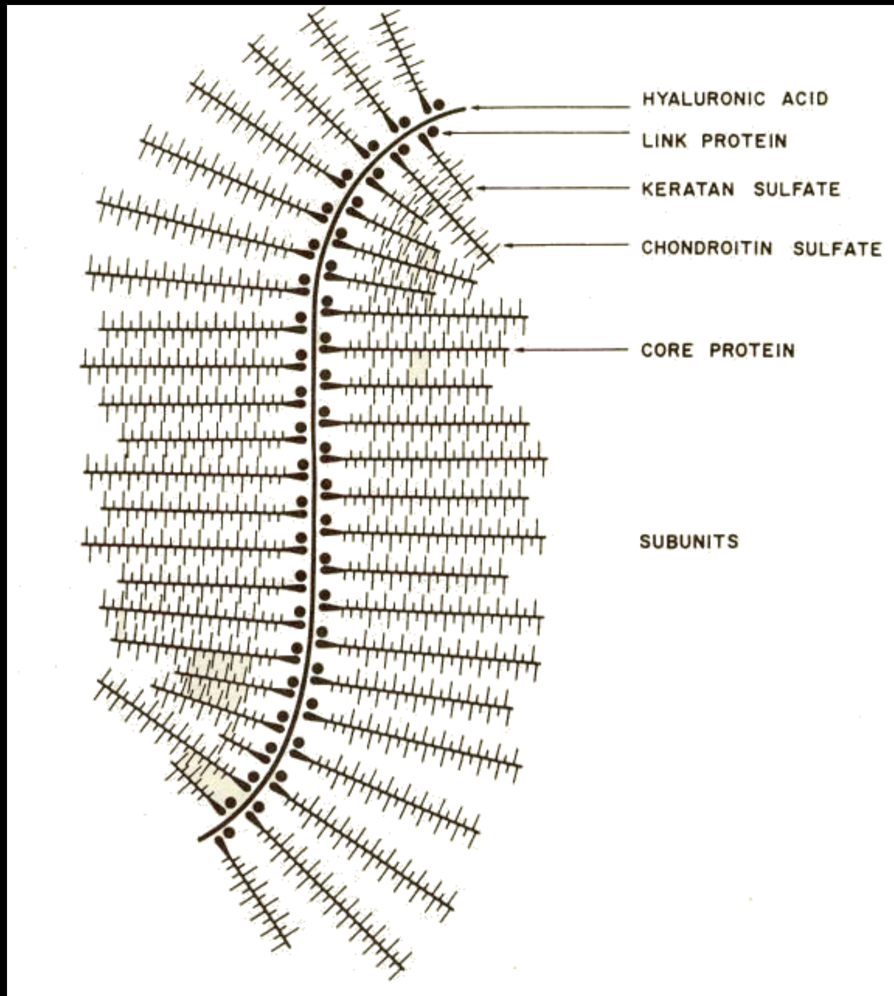

COLLAGEN FIBRIL
Stress
Stress
σ
σ
HIGH TENSILE STIFFNESS AND STRENGTH
Stress
Stress
σ
σ
NO RESISTANCE TO COMPRESSION

Collagen

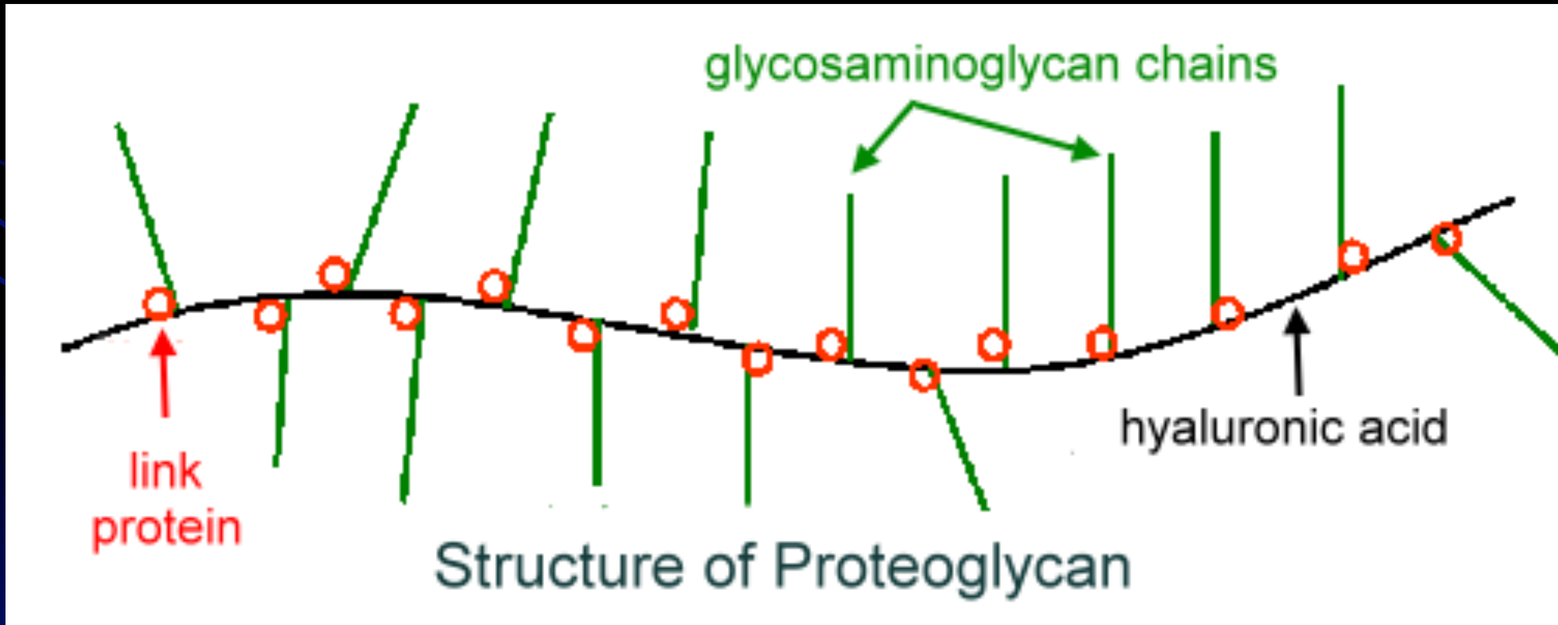


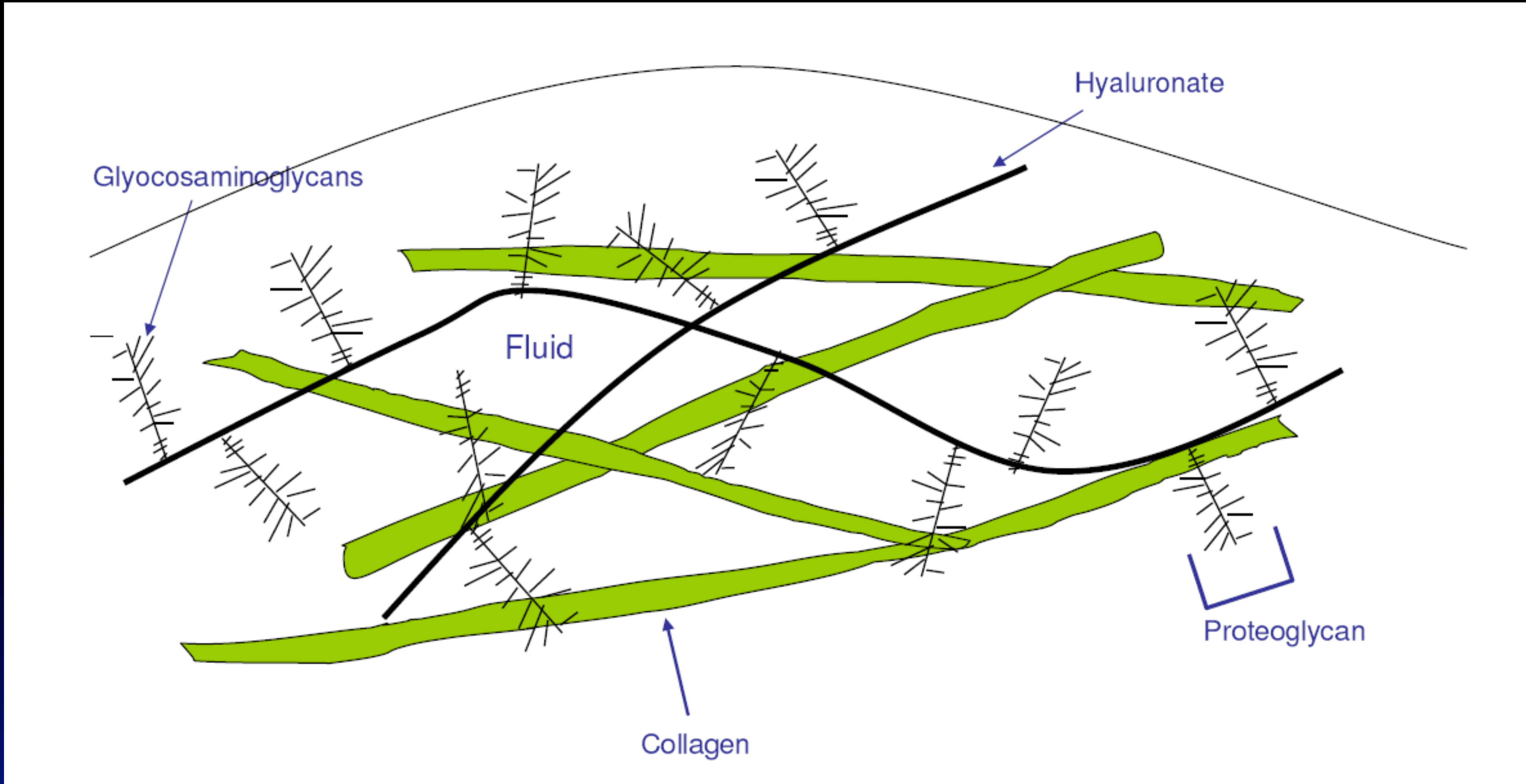
Structure of Proteoglycans (PGs)

Compose about 40% of the dry weight of the articular cartilage that lines joint surface

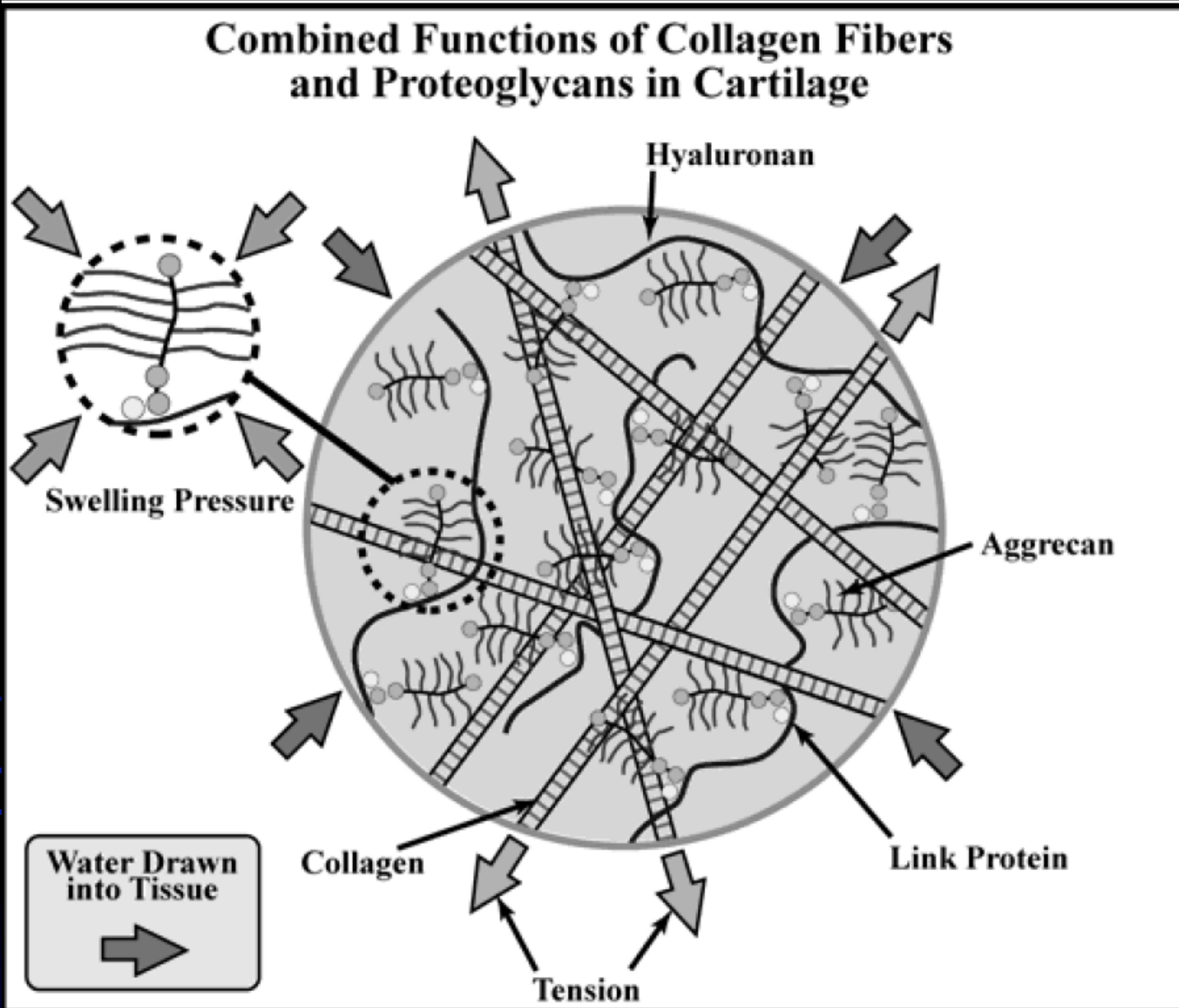


- Each subunit consists of a combination of protein and sugar: Long protein chain
- Sugars units attached densely in parallel
- Subunits are attached at right angles to a long filament
- Produce a macromolecule: the proteoglycan aggregate





Hyaluronate
Glyocosaminoglycans
Fluid
Proteoglycan
Collagen



Combined Functions of Collagen Fibers and Proteoglycans in Cartilage
Hyaluronan
Swelling Pressure
Aggrecan
Water Drawn into Tissue
Collagen
Link Protein
Tension

Pressure in Cartilage

	Max. Pressure
Light Jogging	7.7 MPa
Standing up	9.2 MPa
Stair climbing	10 MPa
One leg stance	6.7 MPa

- Proteoglycans are negatively charged proteins that give cartilage compressive stiffness.
- Collagen fibers are like biological ropes that give tensile strength.
- The interaction between proteoglycan and collagen causes there to be a “balance of forces” within the cartilage.

- Each sugar has one or two negative charges, so collectively there is an enormous repulsive force within each subunit and between neighboring subunits
- This causes the molecule to extend stiffly out in space
- This property gives articular cartilage its resiliency to compression
- The negative charges make the molecules extremely hydrophilic and cause water to be trapped within.
- It is used during biomechanical or lubricant activity.

Electrostatic attractions between the positive charges along the collagen molecules and the negative charges that exist along the proteoglycan molecules.

Hydrostatic forces are applied to cartilage as the fluid tries to move throughout the tissue.

The combined effect of all these interactions gives rise to the mechanical properties of the material.

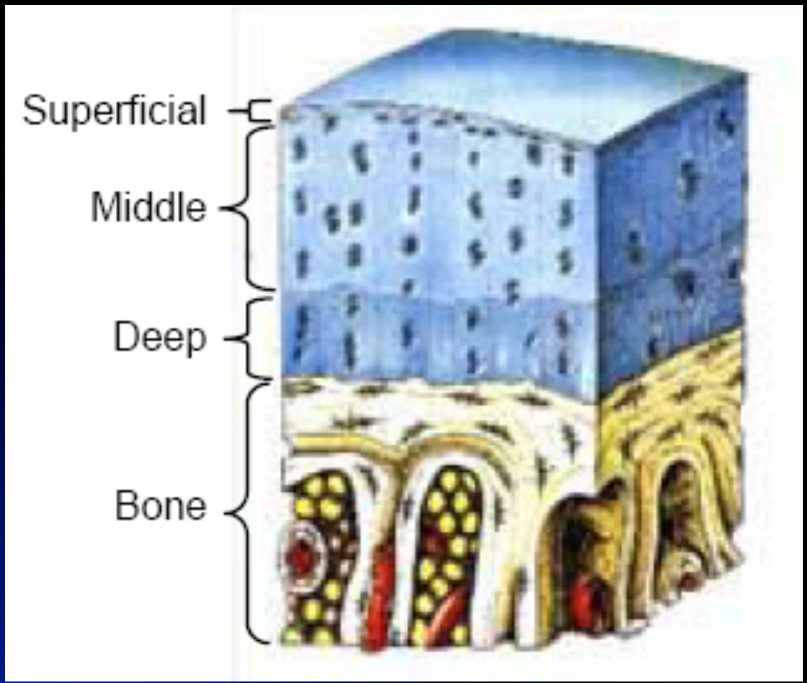
Cartilage Structure

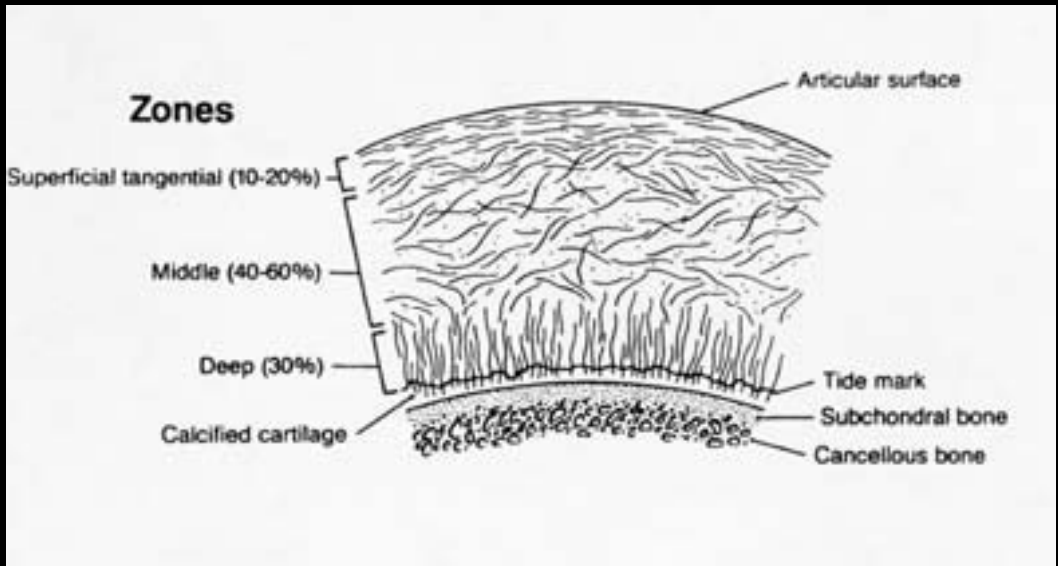
3 layers:

- superficial, 10-20%
- middle, 40-60%
- deep, 25-35%

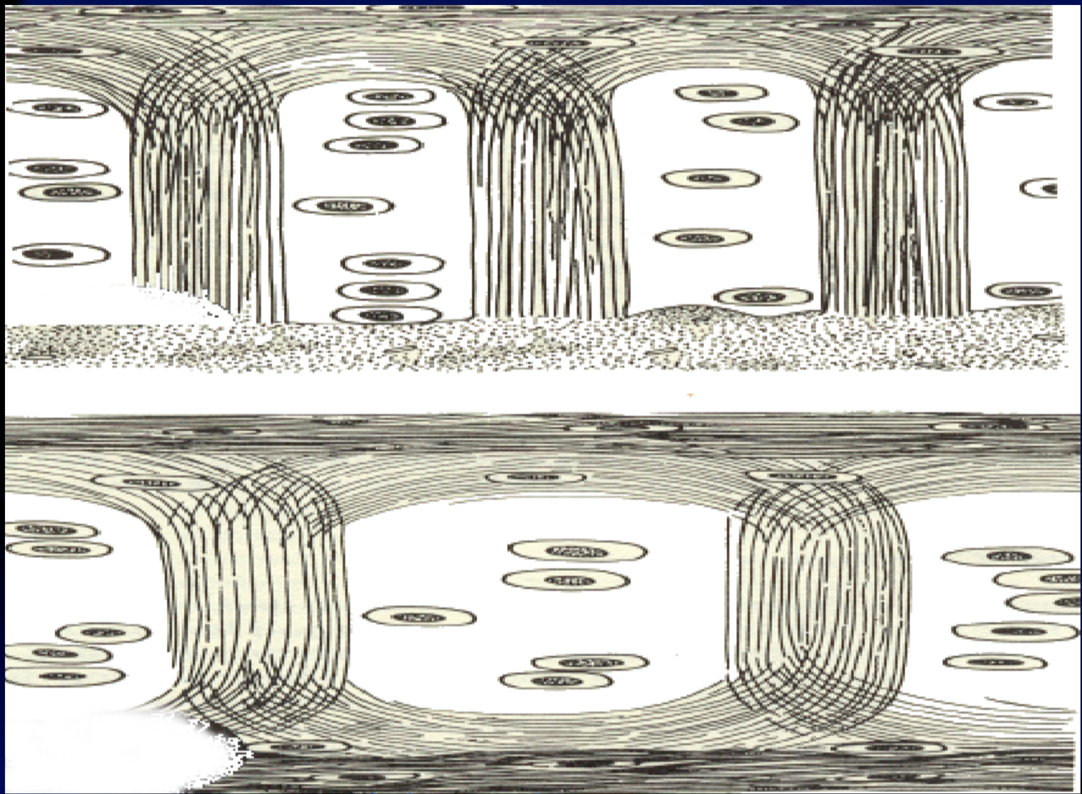
4 zones:

- superficial
- middle
- deep
- calcified



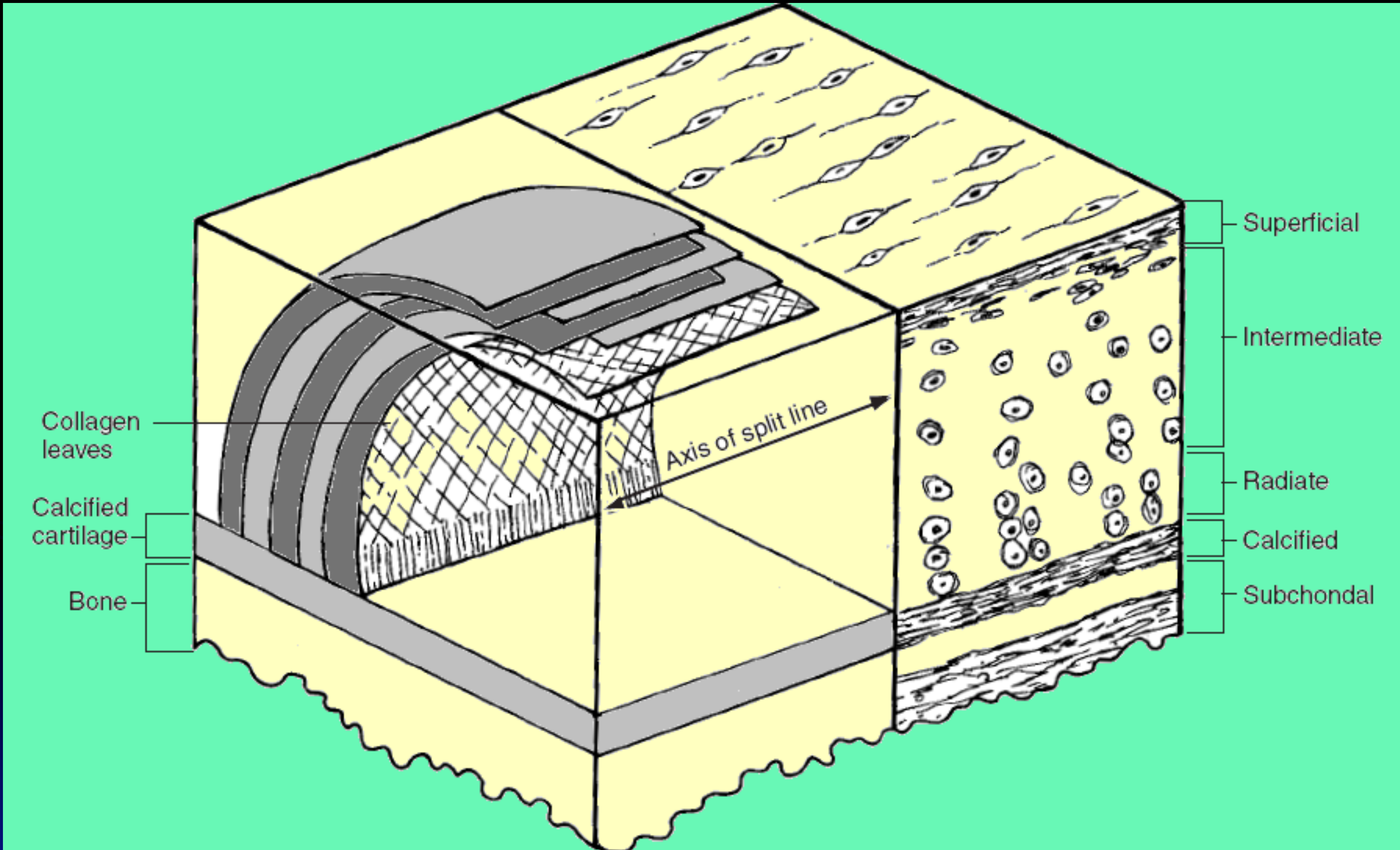


Collagen fiber arrangement: Cartilage

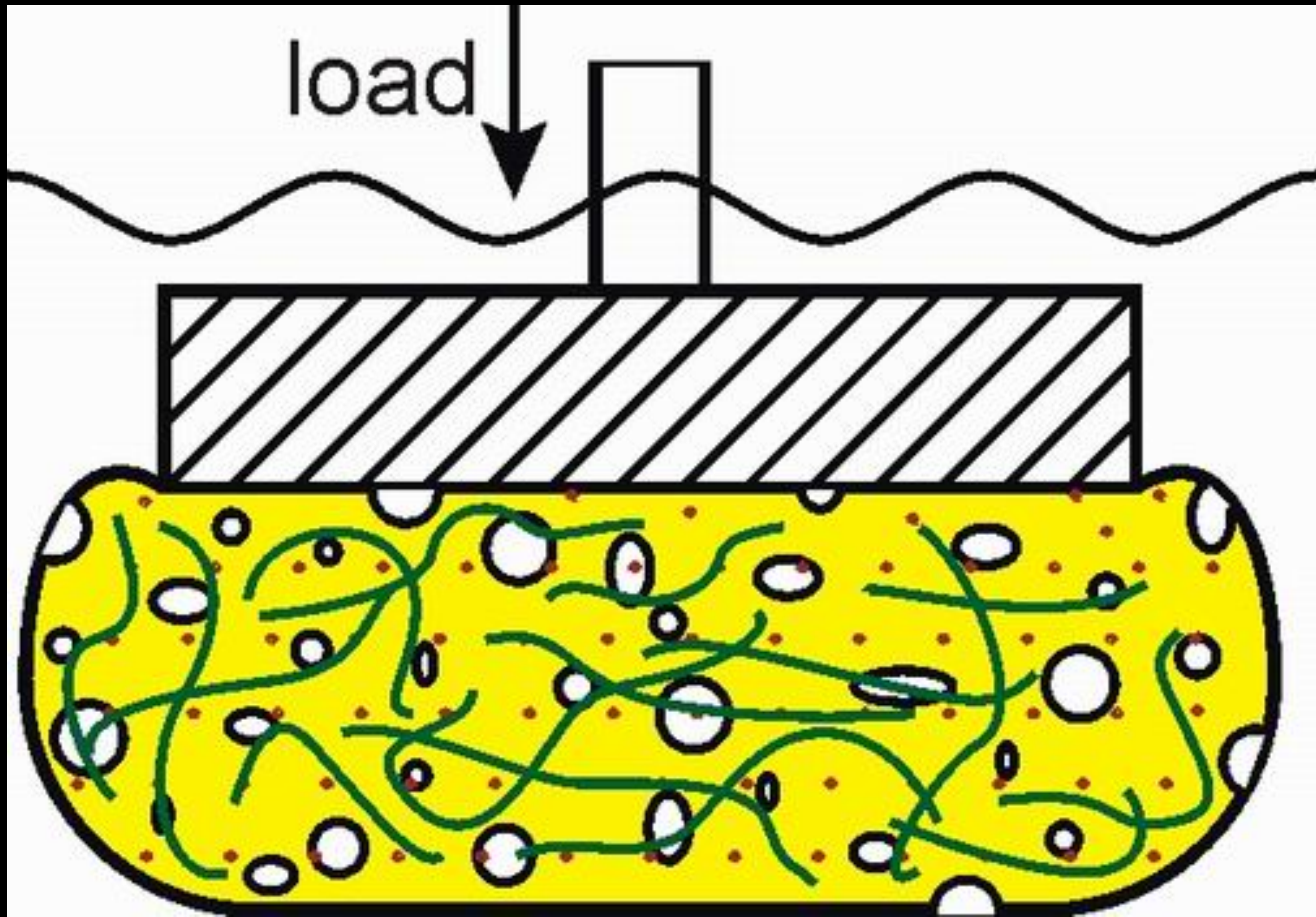


- Articular cartilage
  – single Gothic arches

- Hyaline cartilage
  – double Gothic arches



Collagen leaves
Calcified cartilage
Bone
Axis of split line
Superficial
Intermediate
Radiate
Calcified
Subchondal



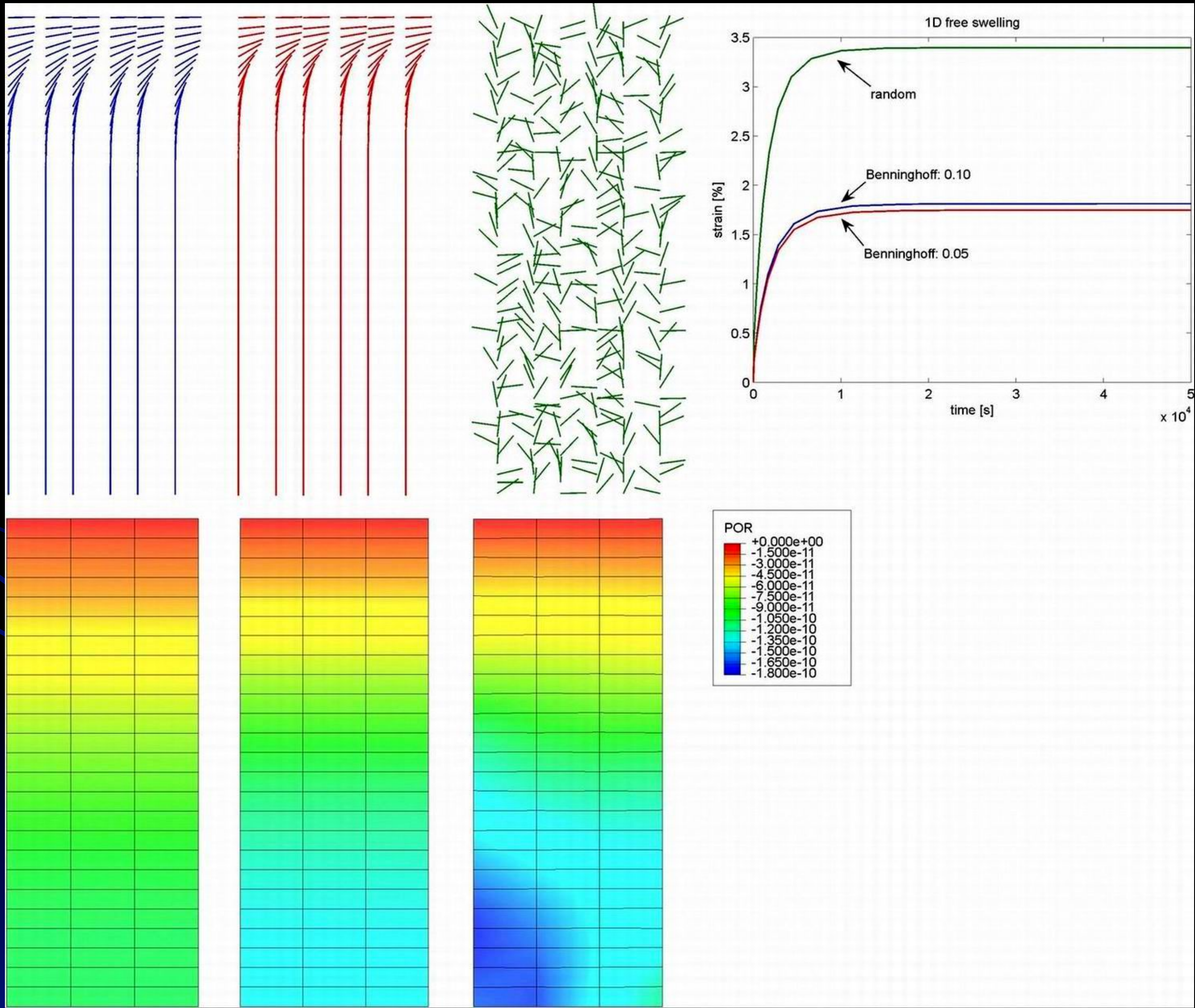
cartilage is a biphasic (sponge-like structure, containing fixed charges or salts (red, causing osmosis) and fibrils (green, adding stiffness in tension).

The distribution and arrangement of cartilage components is not uniform.

Each layer has different biochemical, structural, and cellular characteristics.

Influence of collagen fibril orientation on the free swelling behavior and osmotic pressure of articular cartilage

3 collagen fibril architectures

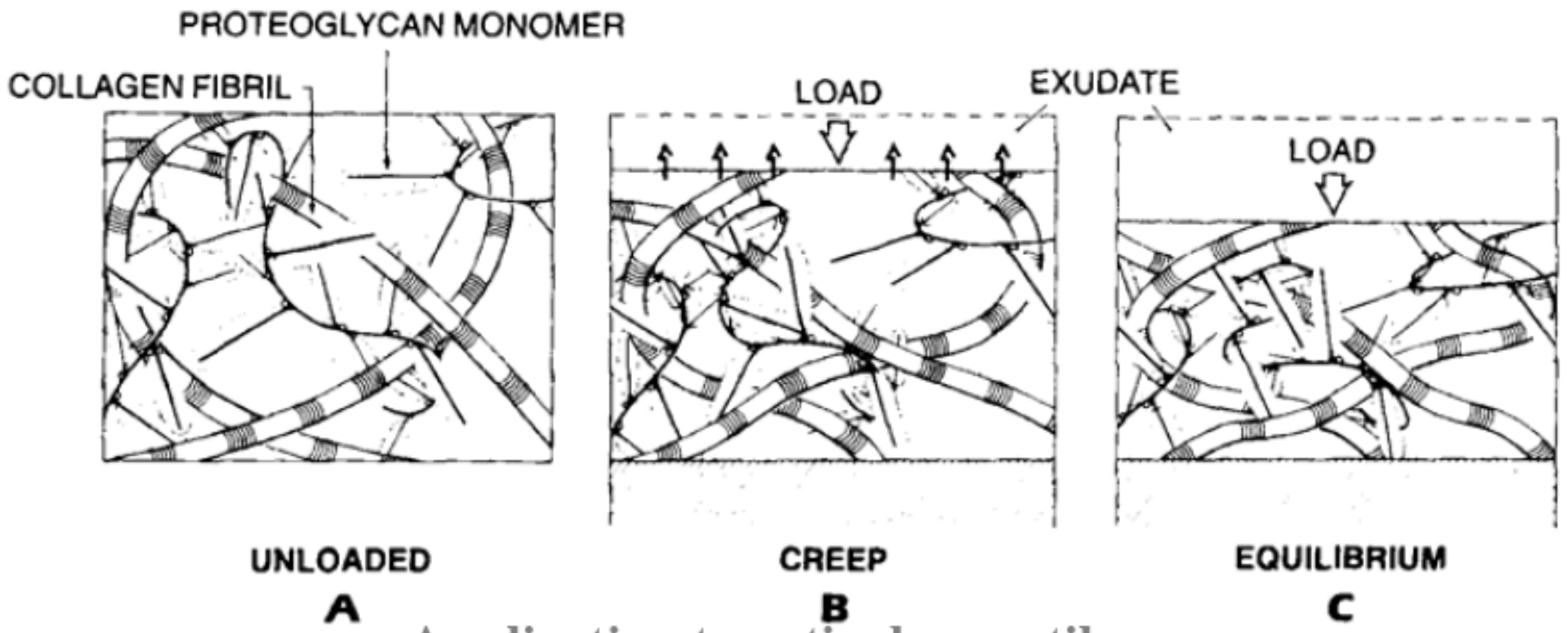


The simple homogeneous appearance of cartilage hides its highly ordered complex structure.

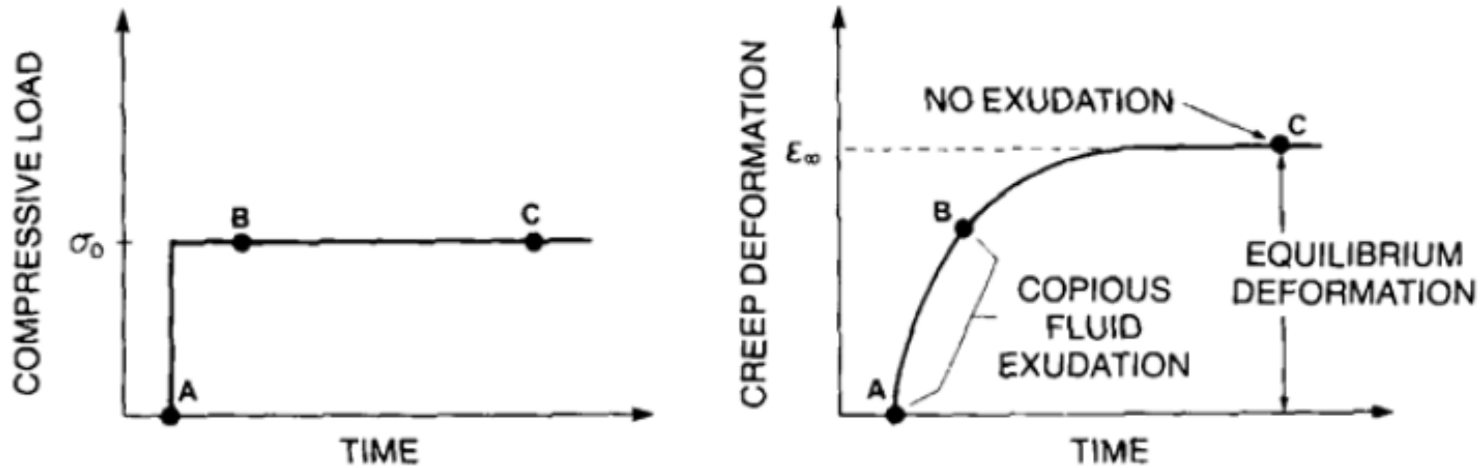
This structure apparently remains unchanged unless affected by disease or injury.

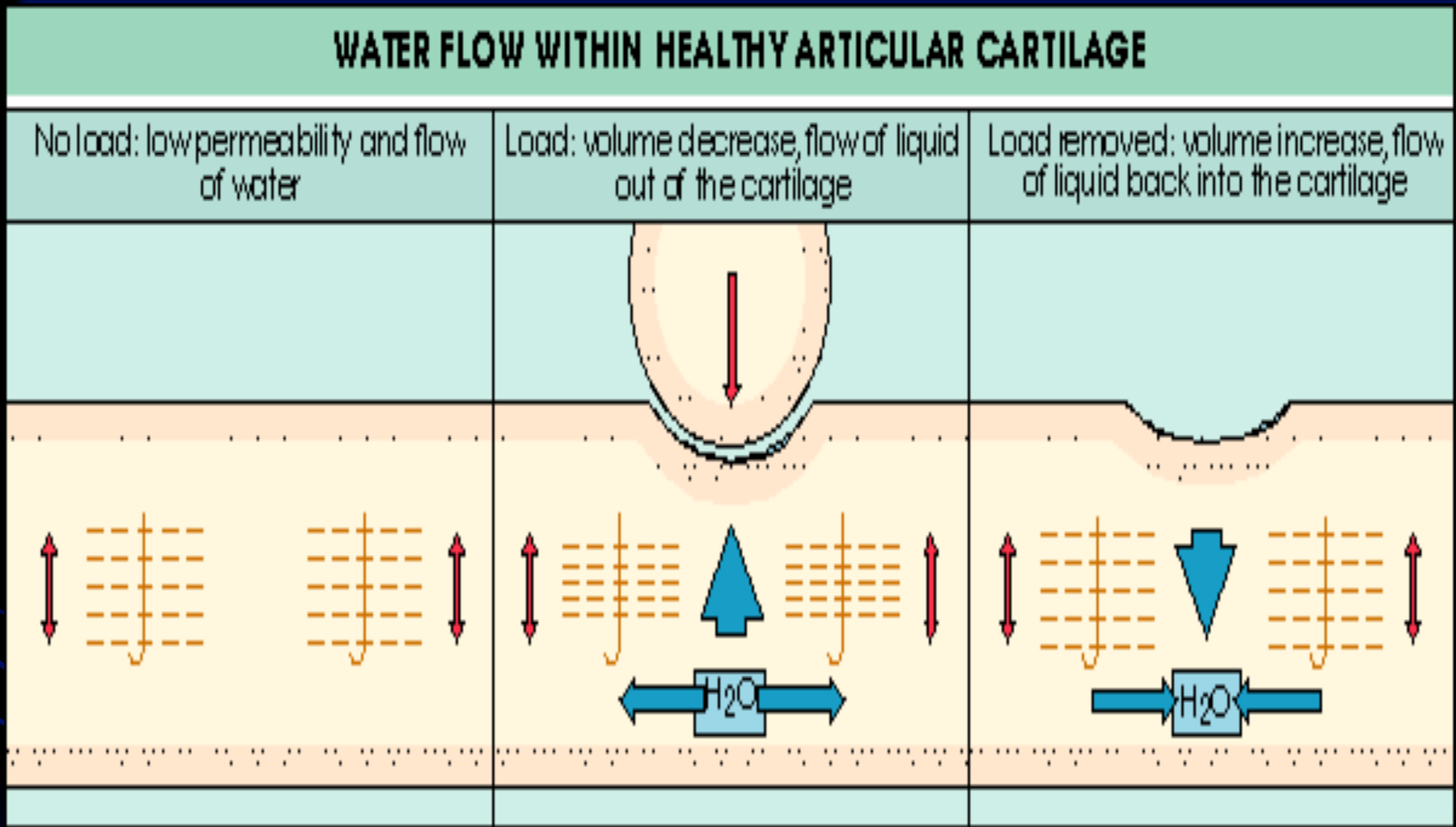
Compression of Normal Cartilage

- When cartilage compresses H2O and solutes are expressed until repulsive forces from PGs balance load applied
- On removing load, PGs rehydrate restoring shape of cartilage
- Loading and unloading important for the exchange of proteins in ECM and thus to chondrocytes



Application to articular cartilage




WATER FLOW WITHIN HEALTHY ARTICULAR CARTILAGE
No load: low permeability and flow of water
Load: volume decrease, flow of liquid out of the cartilage
Load removed: volume increase, flow of liquid back into the cartilage
H2O
H2O

Interstitial solute transport

- acquisition of nutrients
- rejection of wastes
- intercellular signalling
- matrix remodelling

Cartilage: Friction Coefficient

TABLE 7.4. Typical friction coefficients

Materials and conditions	μ
Rubber on concrete, wet or dry, static	1.0
Brake material on cast iron, clean and dry, static	0.4
Brake material on cast iron, lubricated with mineral oil, static	0.1
Graphite on steel, static	0.1
Hickory on dry snow, waxed, 4 m/s, −3°C, dynamic	0.18
Hickory on dry snow, unwaxed, 4 m/s, −3°C, dynamic	0.08
Ice on ice, 4 m/s, 0°C, dynamic	0.02
Articular cartilage in human joints, dynamic:	
Human knee; Charnley (1960)	0.005–0.02
Porcine shoulder; McCutchen (1962)	0.02–0.35
Canine ankle; Linn (1967, 1968)	0.005–0.01
Unsworth et al. (1975)	0.01–0.04
Malcom (1976)	0.002–0.03

Values from *CRC Handbook of Chemistry and Physics* (1984) and Mow and Soslowsky (1991).

Coefficient of Friction

The fraction of force transmitted across two bearing surfaces that must be used to initiate movement (μs -static friction) or keep the surfaces moving (μd -dynamic friction).

The static coefficient of friction between two surfaces is always greater than the dynamic coefficient of friction.

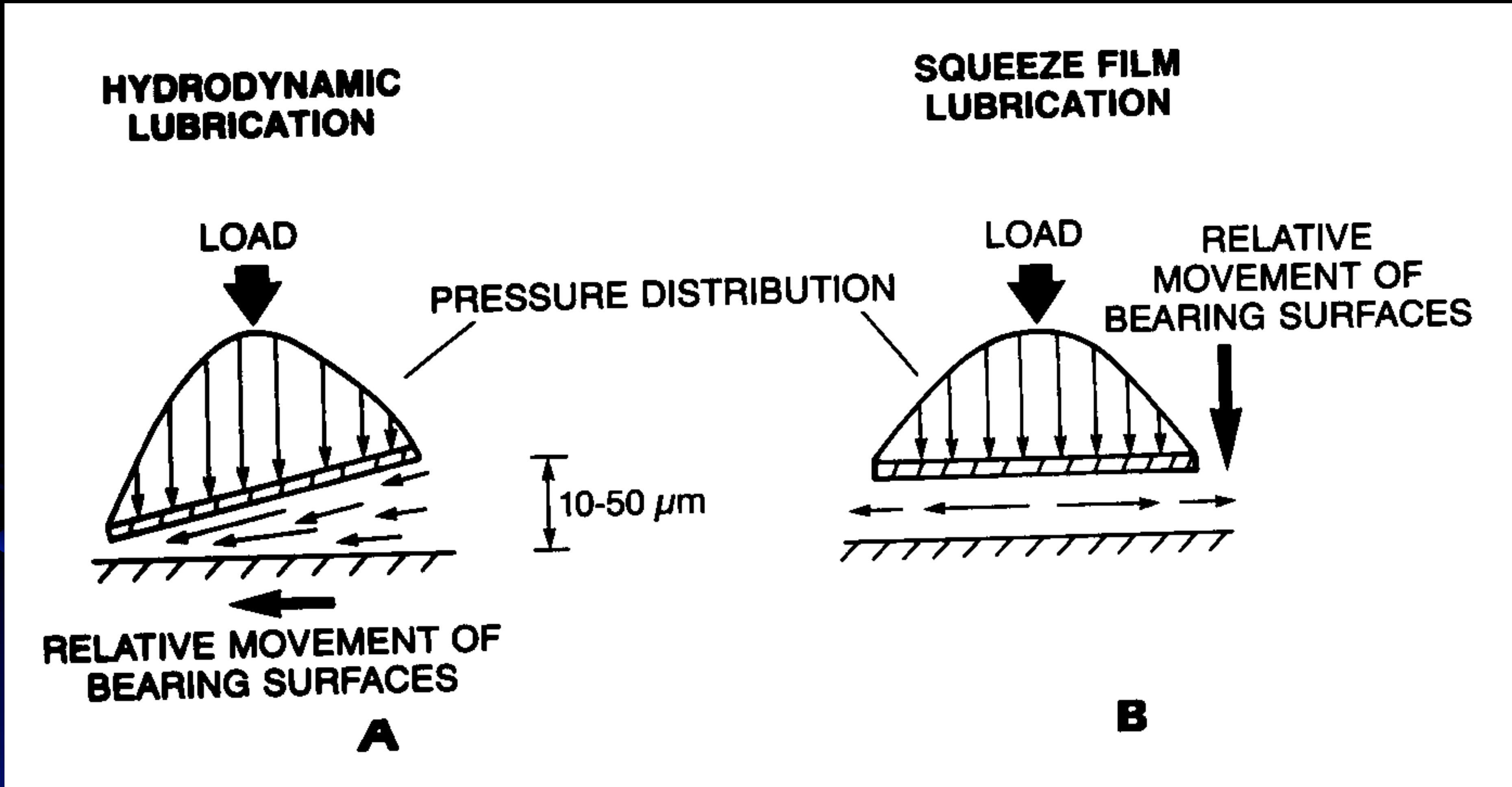
For a fluid film to lubricate moving surfaces effectively, it must be thicker than the roughness of the opposing surfaces.

The thickness of the film depends on the viscosity of the fluid, the shape of the gap between the parts, and their relative velocity, as well as the stiffness of the surfaces.

A low coefficient of friction can also be achieved without a fluid film through a mechanism known as boundary lubrication. In this case, molecules adhered to the surfaces are sheared rather than a fluid film.

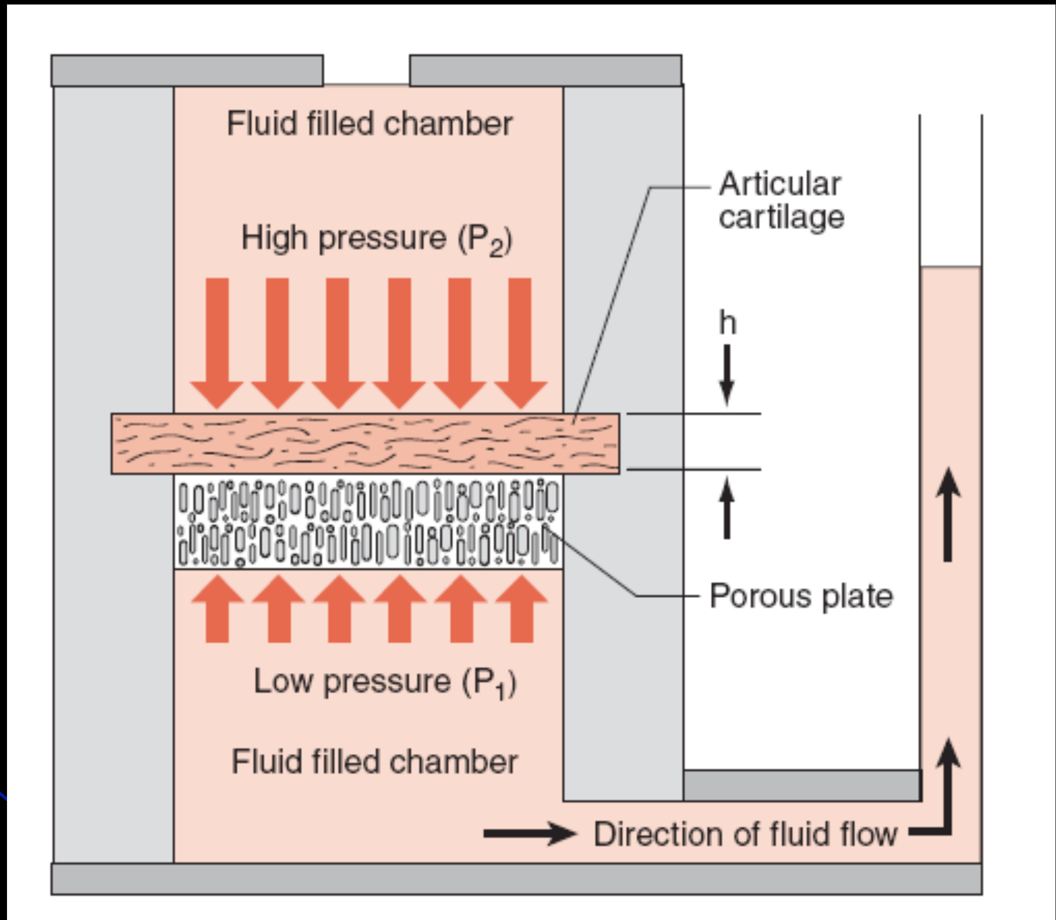
It now appears that a combination of boundary lubrication (at low loads) and fluid film lubrication (at high loads) is responsible for the low friction in synovial joints.

Lubrication Abilities of Articular Cartilage



Cartilage Permeability

The resistance to fluid flow through the cartilage matrix

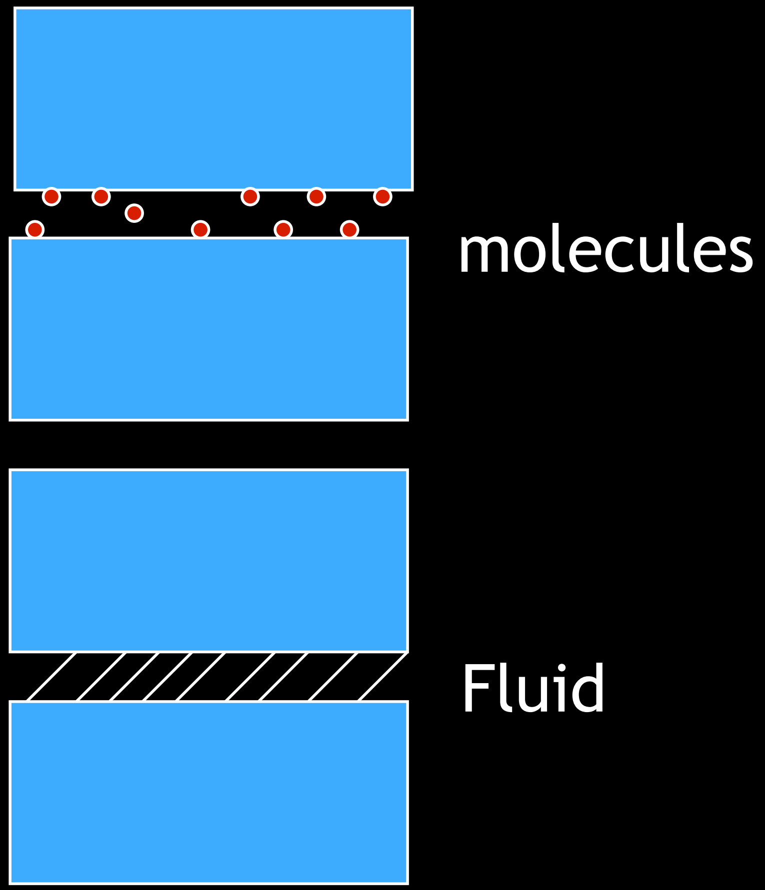


The average fluid velocity (vave) is proportional to the pressure gradient (Δp) and the constant of proportionality (k=*permeability)*.

Darcy's law, vave = kΔp

Articular Cartilage lubrication

- Synovial joints
  - Low coefficients of friction .01-.04
- Theories of lubrication
  - Boundary
  - Fluid film
    - hydrodynamic (non deformable)
    - elastohydrodynamic
  - Squeeze Film
    - right angle movement
    - short duration



Cartilage: Permeability

- *Biphasic system*:
  - liquid+solid
- Water molecules restrained by PGs
- Limited mobility due to pressure gradient.
- Darcy's Law:
  - $Q = k\, dP/dx$
    - Q : flow rate per unit area
    - dP/dx : pressure gradient
    - k : hydraulic permeability coefficient

Cartilage: Permeability

- permeability diminishes in the deepest layers
- permeability affected by:
  - composition of the liquid phase
  - deformation

The permeability of cartilage is 10_15 to 10_16 m4/Ns.

If a pressure difference of 210,000 Pa (automobile tire) is applied across a slice of cartilage 1 mm thick, the average fluid velocity will be only 1x10_8 m/s, which is about 100 million times slower than normal walking speed. Permeability is not constant through the tissue.

The permeability is highest near the joint surface (making fluid flow relatively easy) and lowest in the deep zone (making fluid flow relatively difficult).

Permeability also varies with deformation of the tissue. As cartilage is compressed, its permeability decreases.

As a joint is loaded, most of the fluid that crosses the articular surface comes from the cartilage closest to the joint surface. Under increasing load, fluid flow will decrease because of the decrease in permeability that accompanies compression.

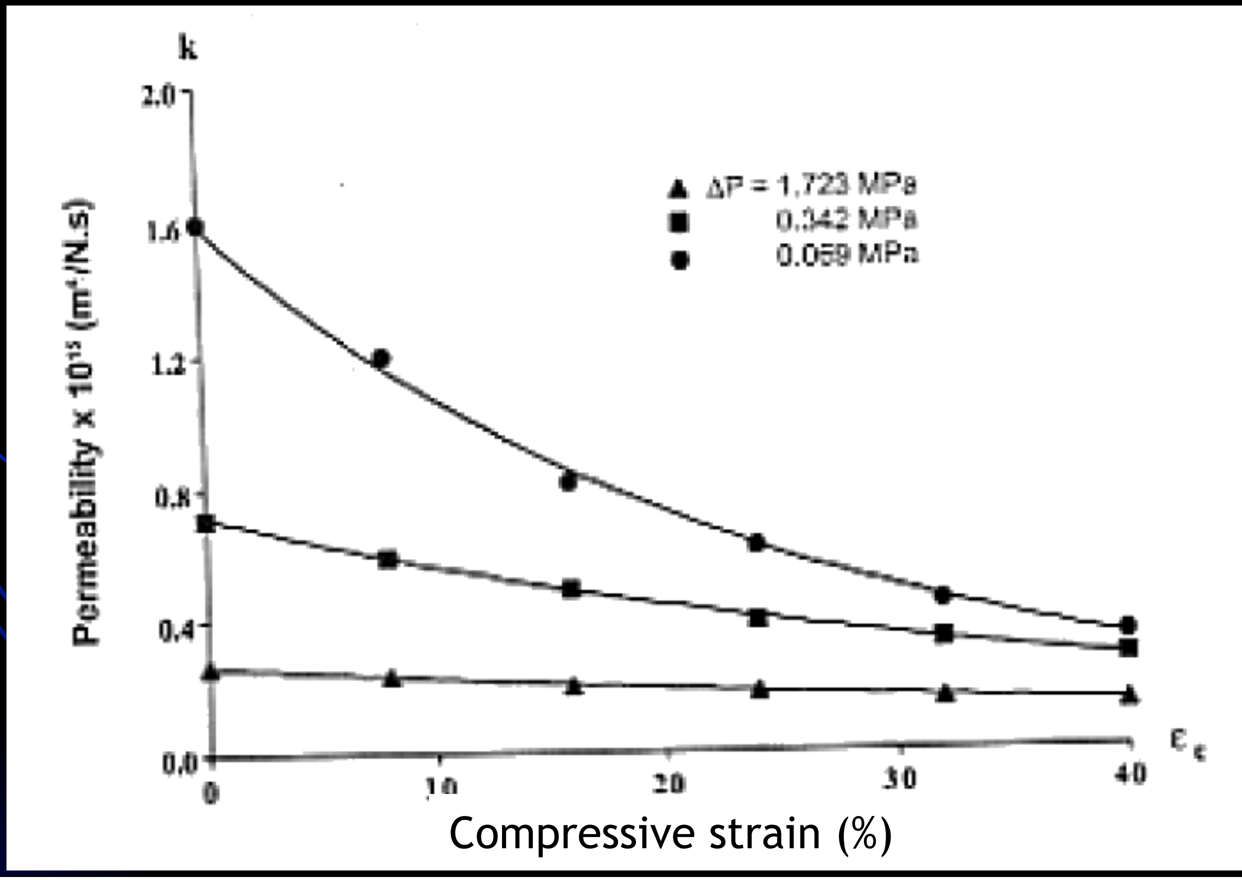
Permeability of Osteoarthritic Cartilage

- lower modulus and increased permeability
- greater and more-rapid deformation of the tissue than normal.

These changes may influence the synthetic activity of the chondrocytes, which are known to respond to their mechanical environment.

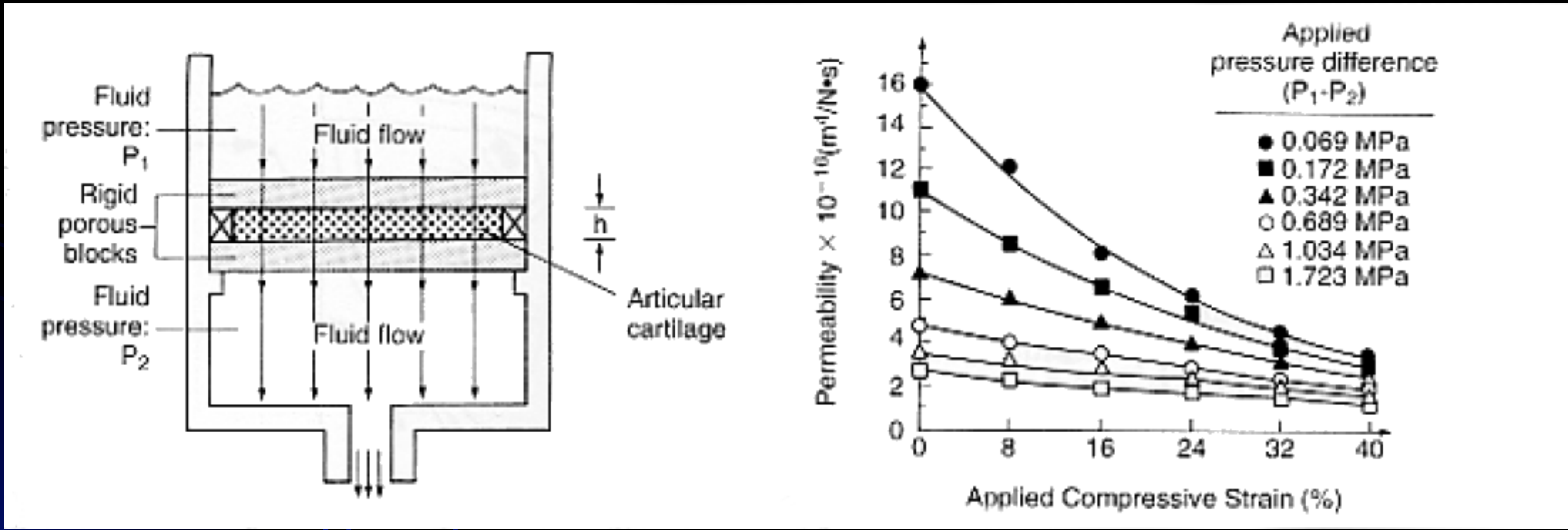
Permeability

- Articular cartilage shows nonlinear strain dependence and pressure dependence
- The decrease of permeability with compression acts to retard rapid loss of interstitial fluid during high joint loadings

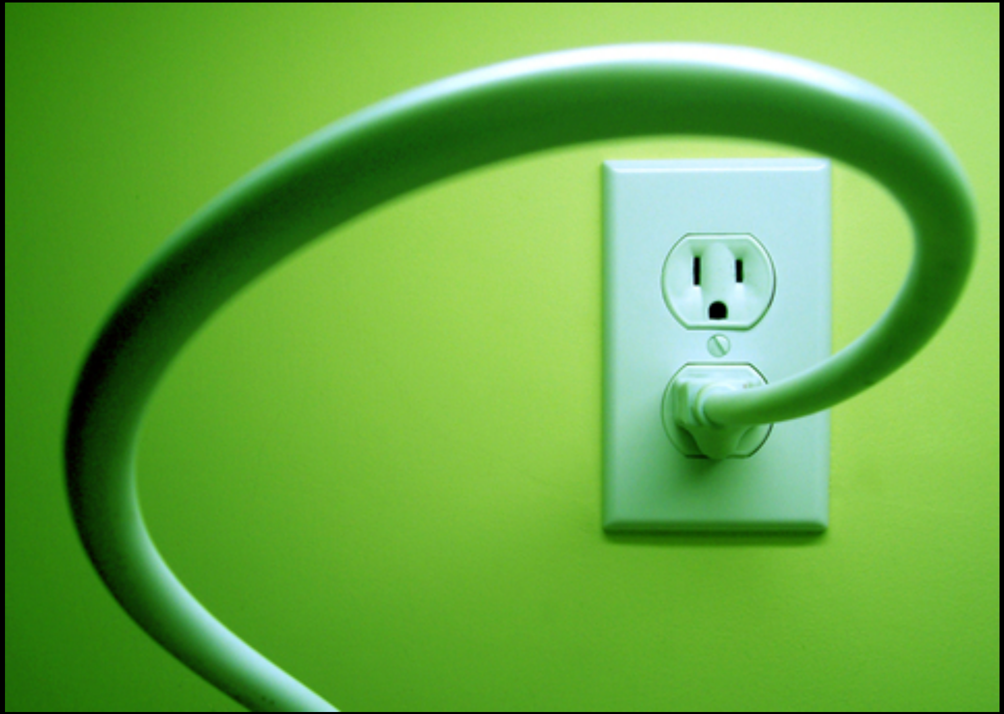


Permeability

Permeability decreases in an exponential manner as function of both increasing applied compressive strains and increasing applied pressure



Electric Phenomena in Cartilage



ECM of articular cartilage

A hydrated polyelectrolyte gel containing fixed negative charges on proteoglycans enmeshed within a collagen network, with corresponding positive charges in the matrix fluid.

- proteoglycans are polyanionic, negatively charged molecules (sulfate and carboxyl groups)

- the mutual repulsion of the negative charges causes an aggregated proteoglycan molecule to spread out and occupy a large volume.

- the volume occupied by proteoglycan aggregates is limited by the entangling collagen framework.

electrokinetic phenomena

relative motion between matrix fluid and solid is induce giving rise to electrokinetic phenomena:

- streaming potentials that accompany pressure-driven fluid flows

- electroosmotic fluid flows driven by electric current

Electrokinetic phenomena

- contribute to functions of many biological tissues.

- depend upon the concentration of glycosaminoglycans immobilized in the matrix.

Compression

Cartilage glycosaminoglycans change their orientations and relative positions such that matrix molecular architecture can acquire direction dependent features though none exist in unloaded tissue.

Such changes appear to underlie the emergence of anisotropy (direction-dependence) in the hydraulic permeability tensor of cartilage with increasing compression.

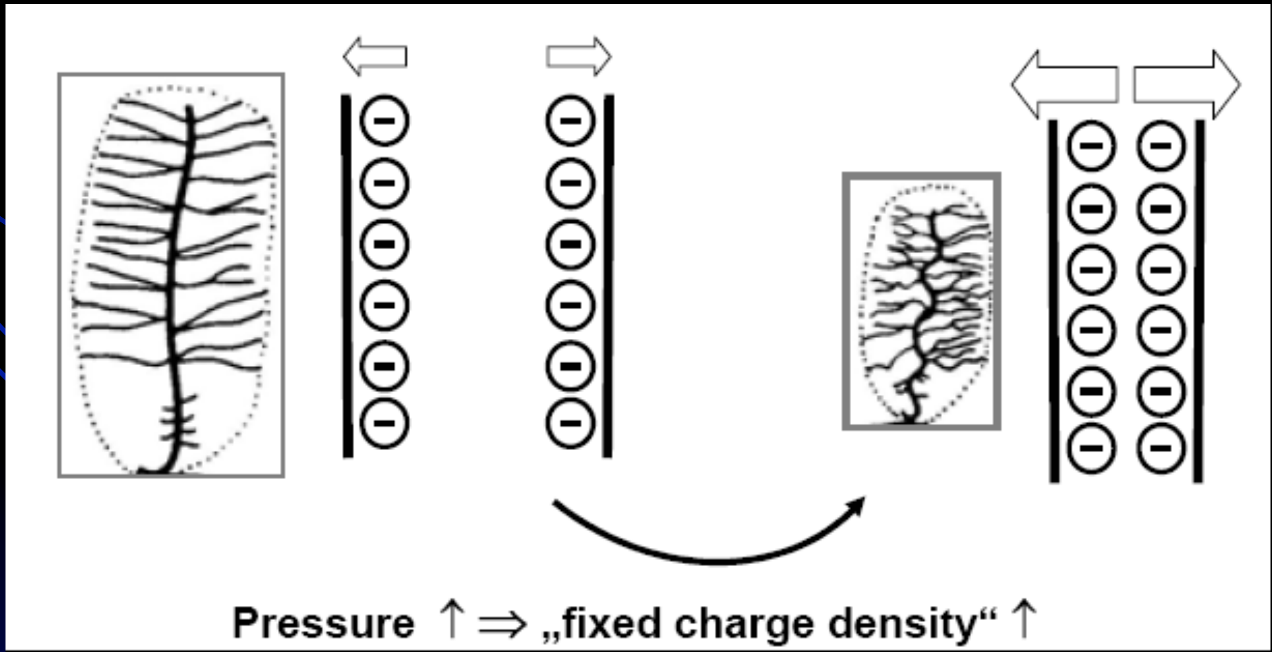
External loading

- decrease in the interstitial fluid
- increased proteoglycan concentration
- increase in the osmotic swelling pressure

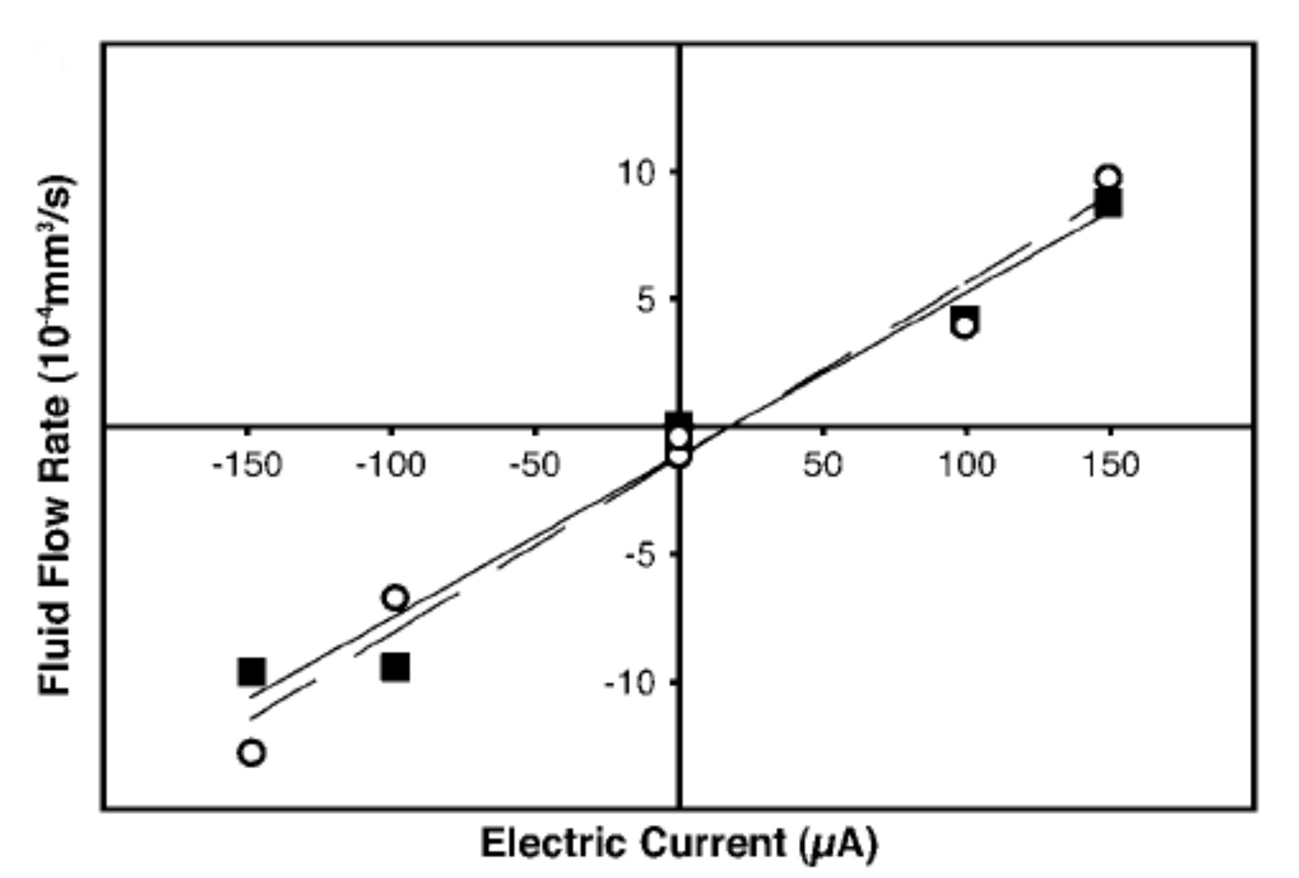
collagen fibers are strong in tension but can easily fold under compression.

Loading effects

- Bigger molecule
- more water
- less negative loading
- Smaller molecule
- less water in between
- more negative loading



Applied electric current and radial flow versus time for a cartilage disk under 29% static compression.



Basic Biomechanics

Structural vs. Material Properties

- A biological tissue is often described in terms of its structural and material properties.
- Structural properties characterize the tissue in its intact form.
- Material properties characterize the behavior of the material comprising the tissue and are independent of the size of the tissue.
- The material properties are usually expressed in terms of the stress-strain relationship of the material.
- Structural and material property curves are similar in appearance, differing only by a scaling factor.

Mechanical Properties of Orthopaedic Tissues

- Compliant: much softer than synthetic materials
- Anisotropic: properties depend on orientation
- Non-linear: stress-strain
- Viscoelastic: time dependent
- Poroelastic: fluid flow and permeability
- Remodeling: structural evolution

Anisotropy and Non-homogeneity

- Ideal materials are isotropic and homogeneous.

- A material is called isotropic when its properties are the same in each of three coordinate axes (x,y,z).

- Tensile and compressive properties may be different, but each respective property must be the same in three directions.

- A material is said to be homogeneous if it is made of the same material throughout.

- Biological tissues are anisotropic and non-homogeneous.

The mechanical properties of cartilage change with its fluid content, thus making it important to know the stress-strain history of the tissue to predict its load carrying capacity.

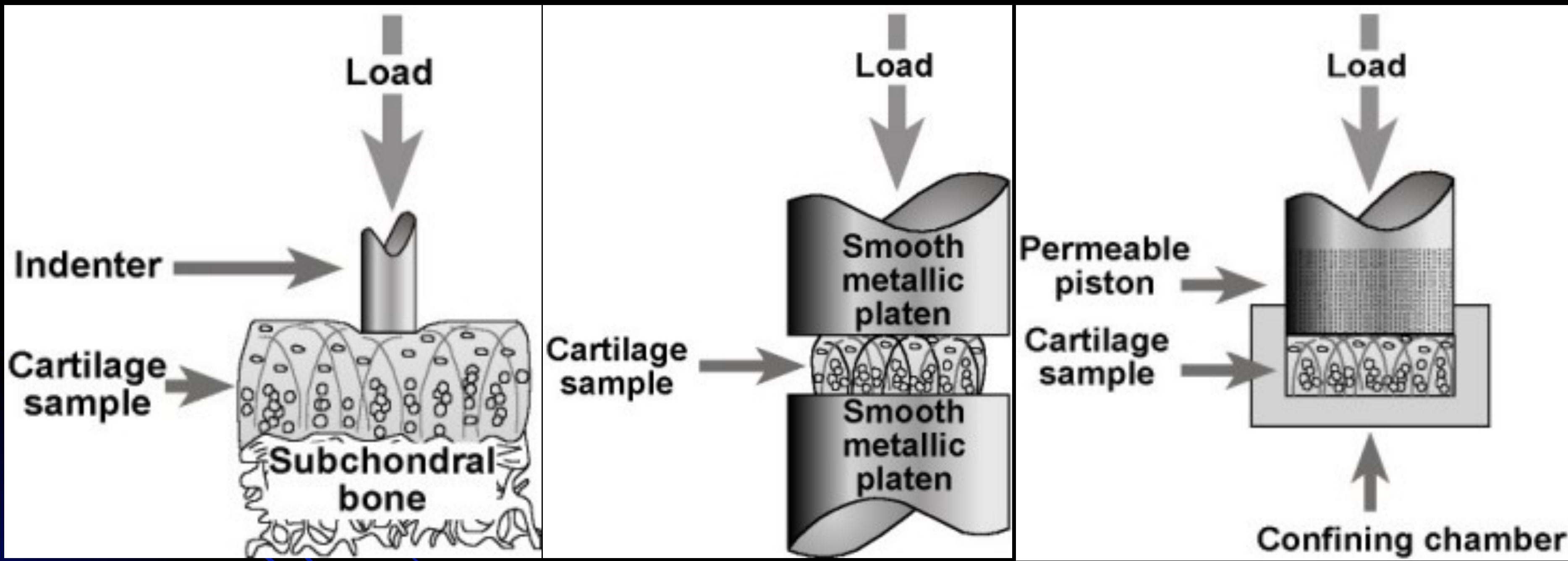
The material properties also change with pathology.

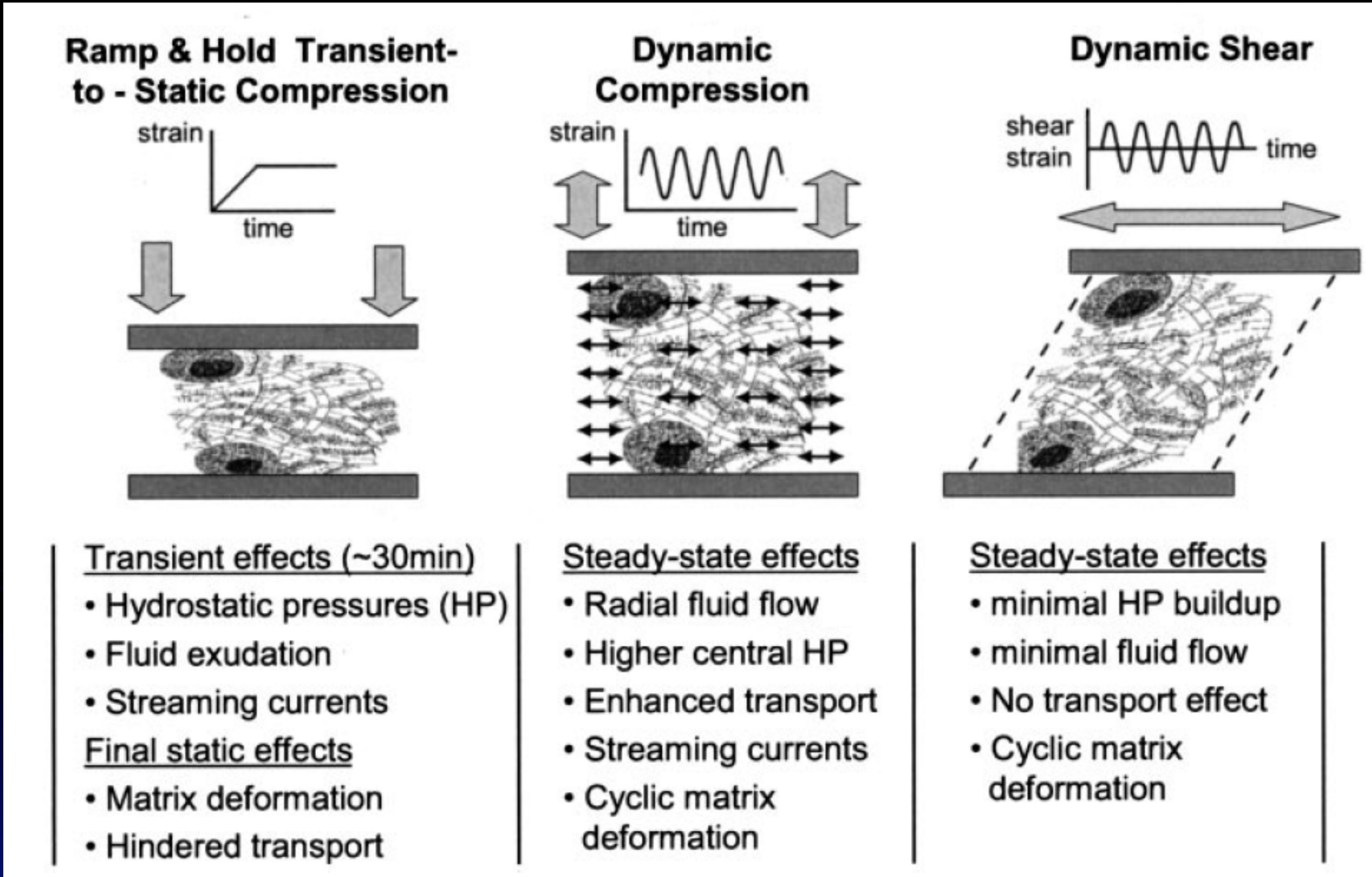
The compressive aggregate modulus for human articular cartilage correlates in an inverse manner with the water content and in a direct manner with proteoglycan content per wet weight.

There is no correlation with the collagen content thus suggesting that proteoglycans are responsible for the tissue's compressive stiffness.

3 modes of cartilage testing

indentation, unconfined and confined compression

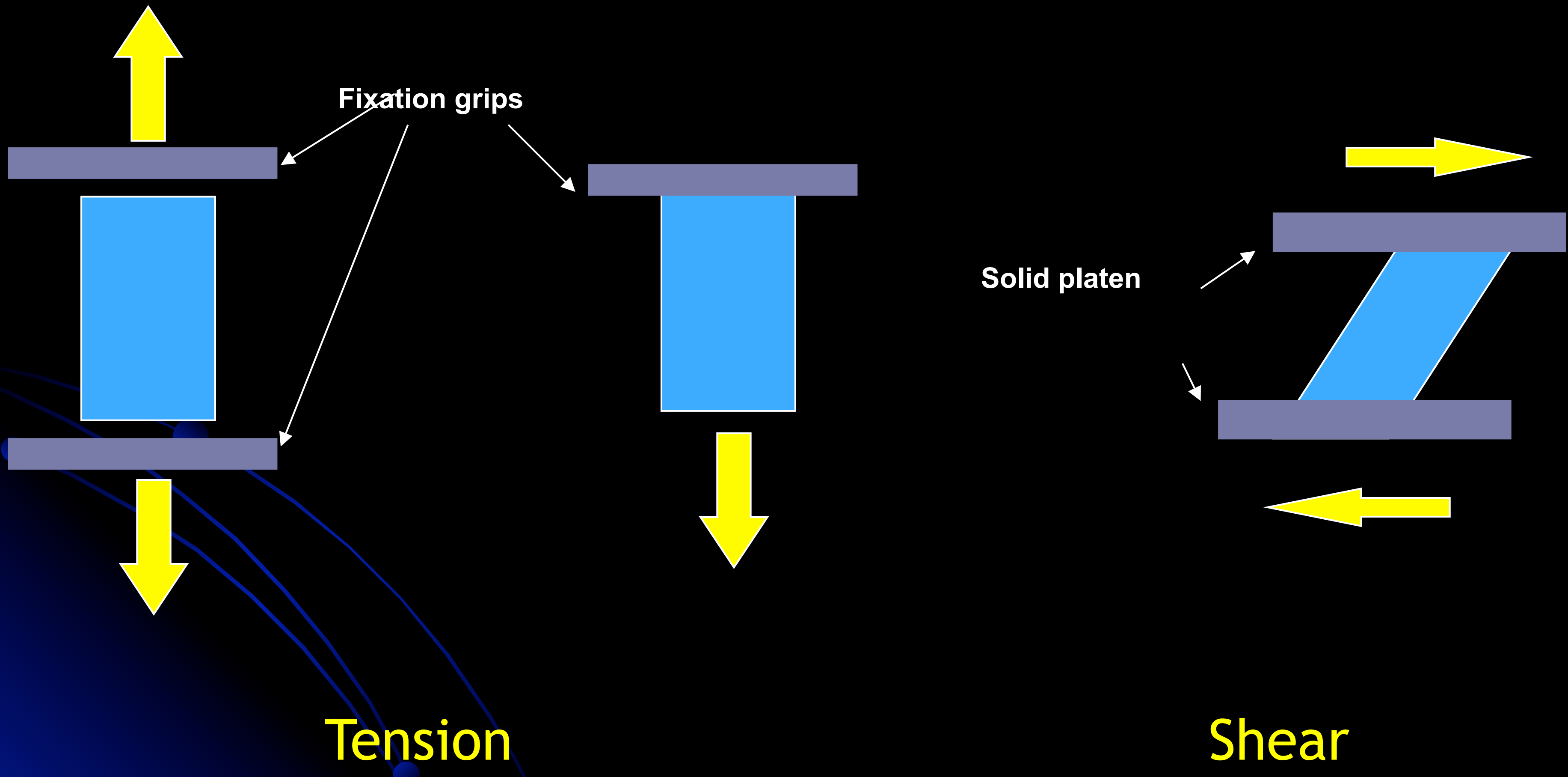




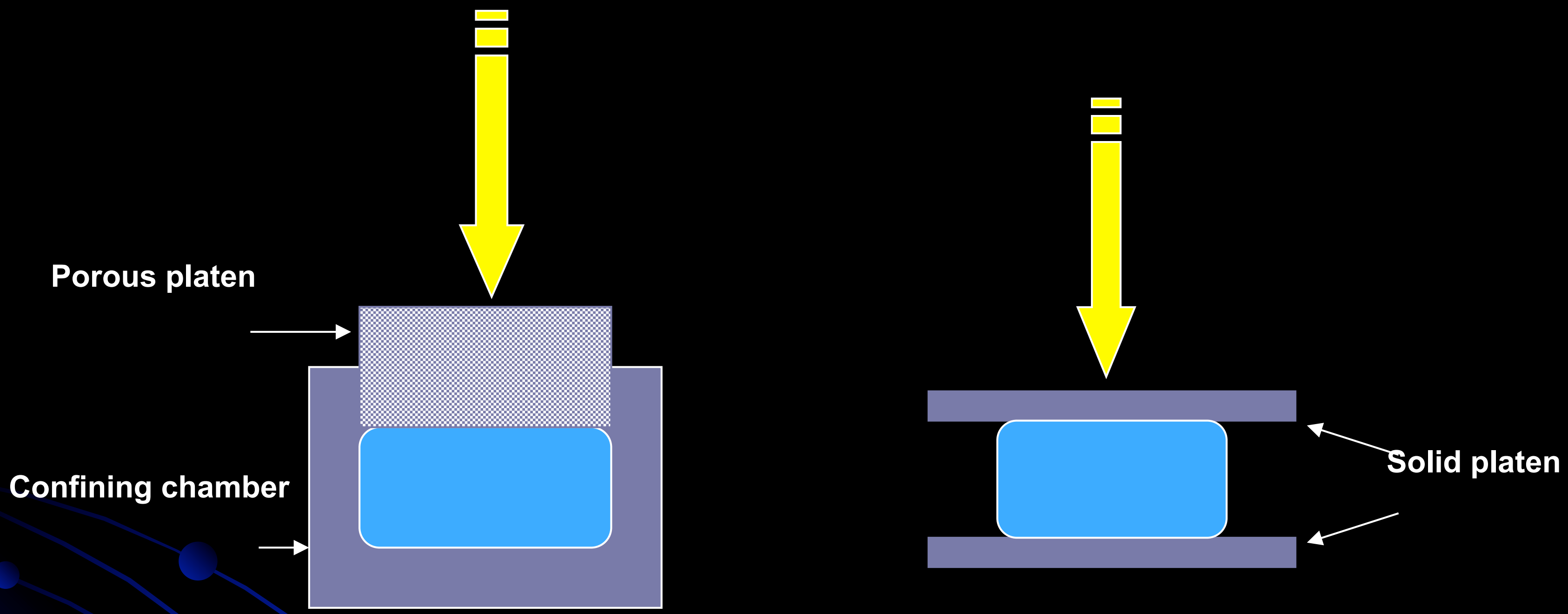
Normal joint motion *in vivo* produces a superposition of all these components of cartilage loading.

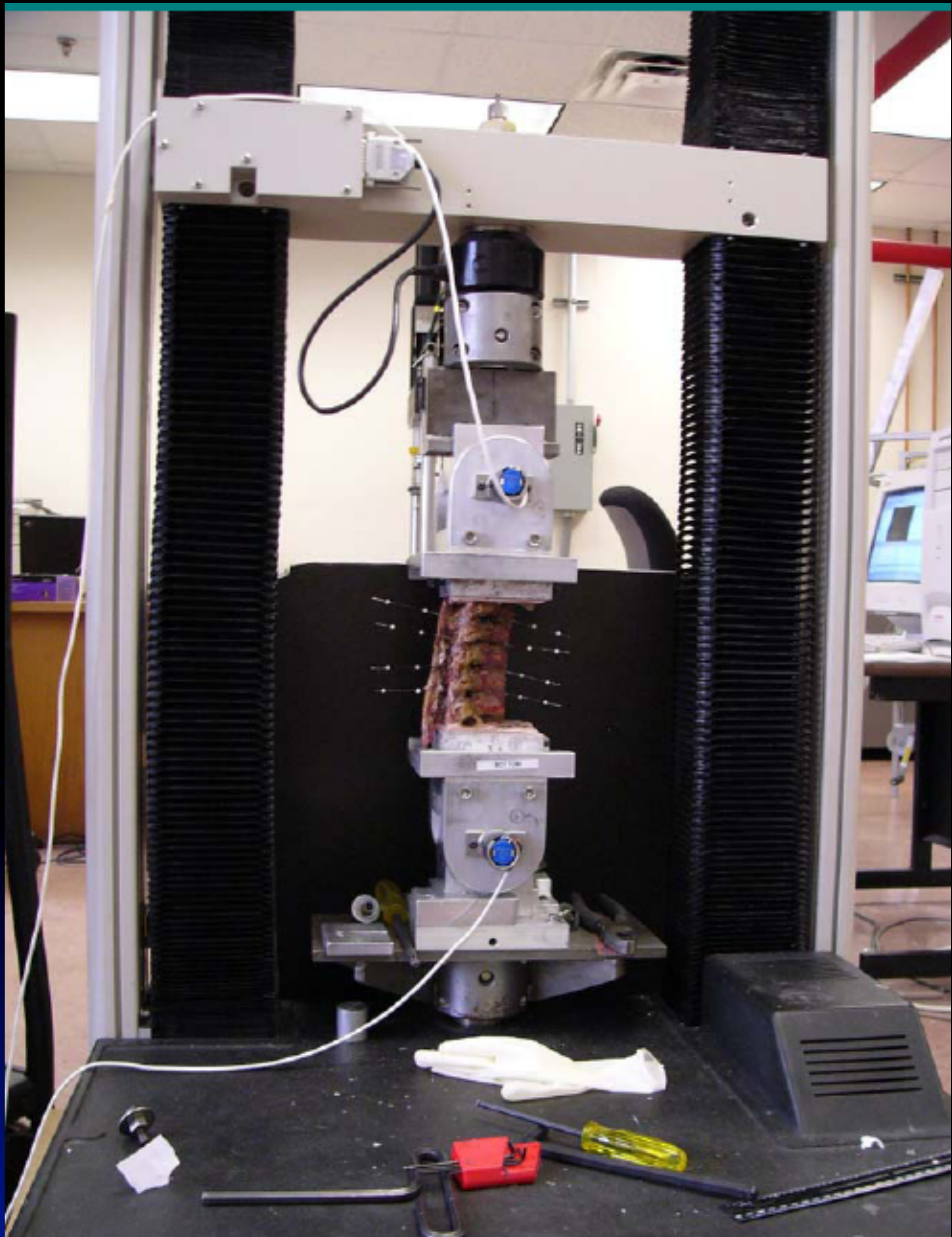
Mechanical Properties of Cartilage

Viscoelastic, nonlinear, inhomogeneous, anisotropic



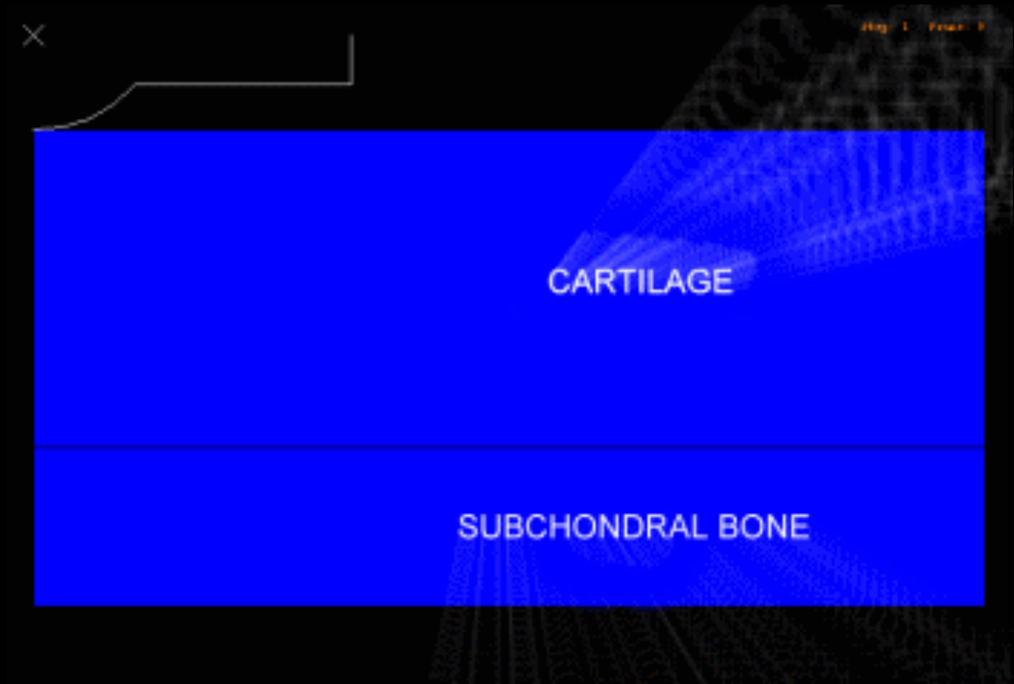
Compression






41269 if
TOP
BOTTOM

Impact load

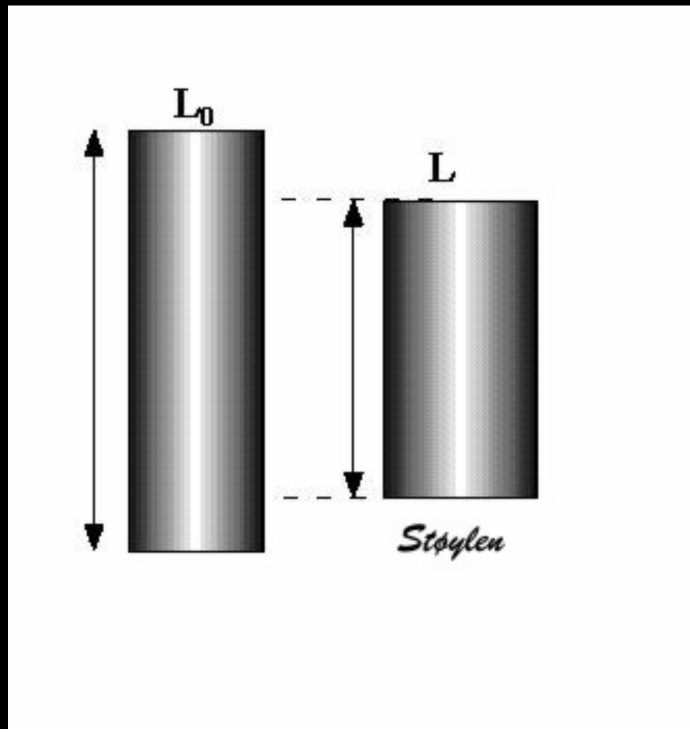


- Cartilage behaves as a single-phase, incompressible, elastic solid.

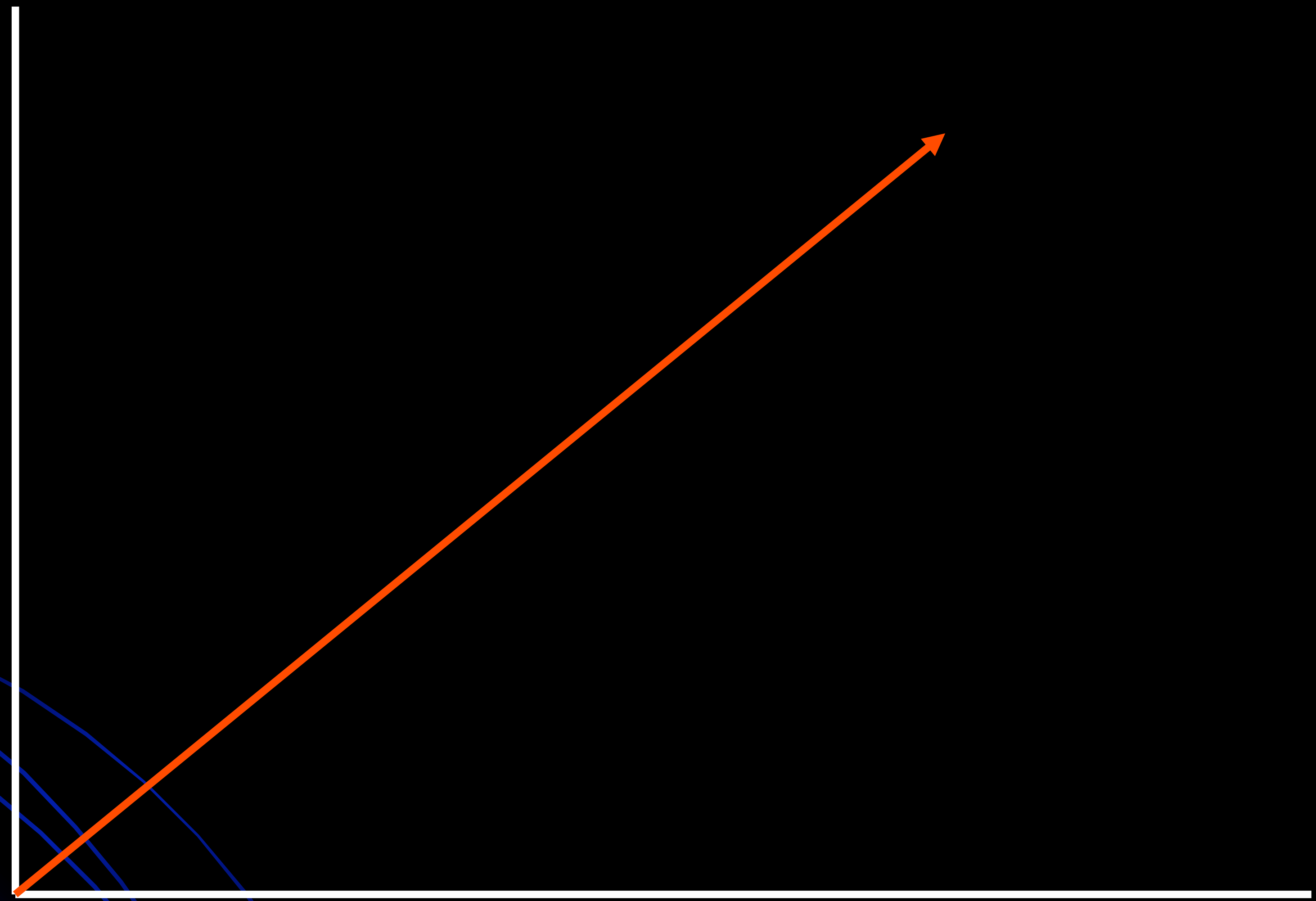
- There isn't time for the fluid to flow relative to the solid matrix under rapidly applied loads.

Force - Deformation

Stress - Strain

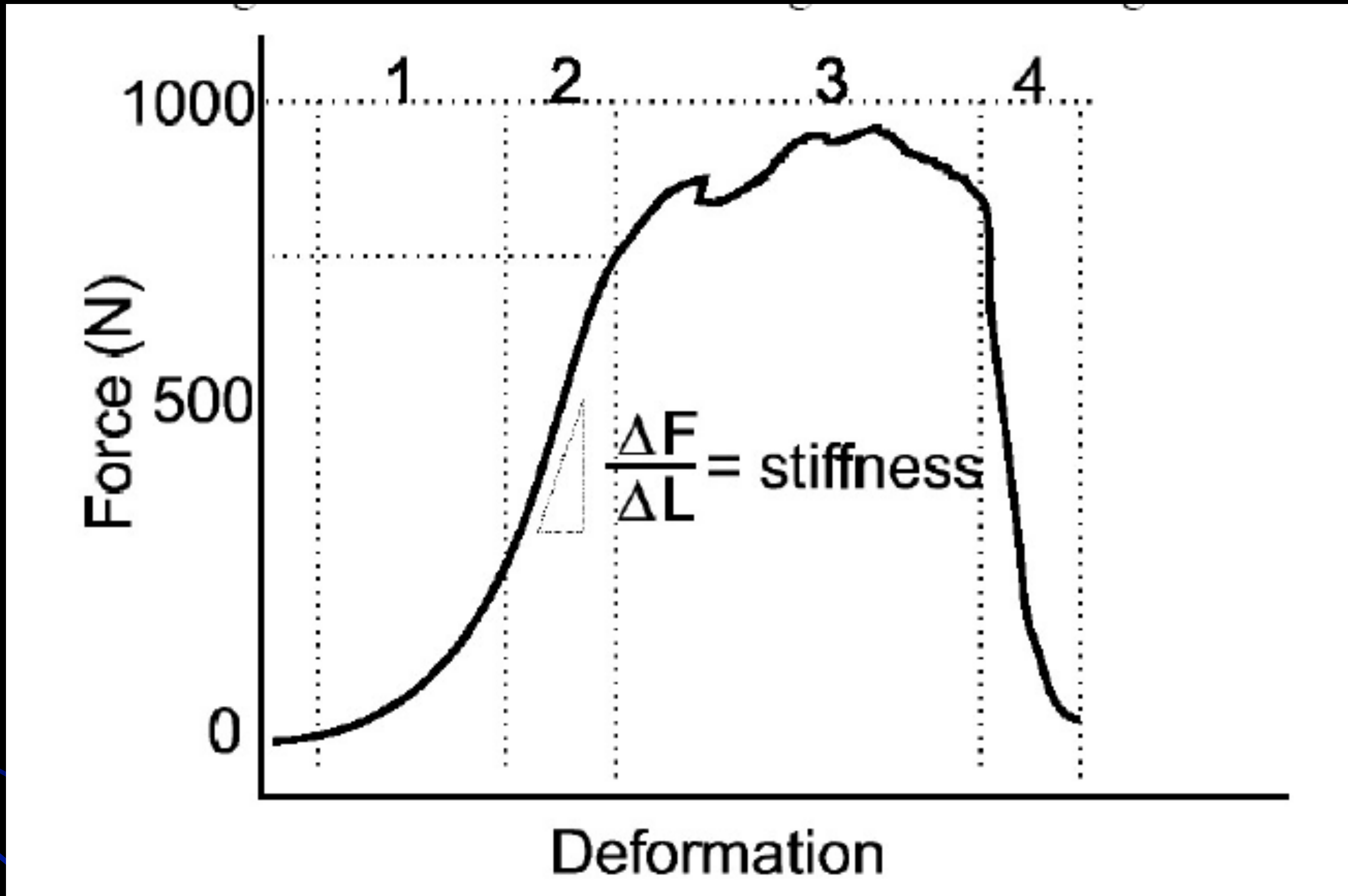


Hookean Behaviour



The stress in a material is directly proportional to the strain for all strains

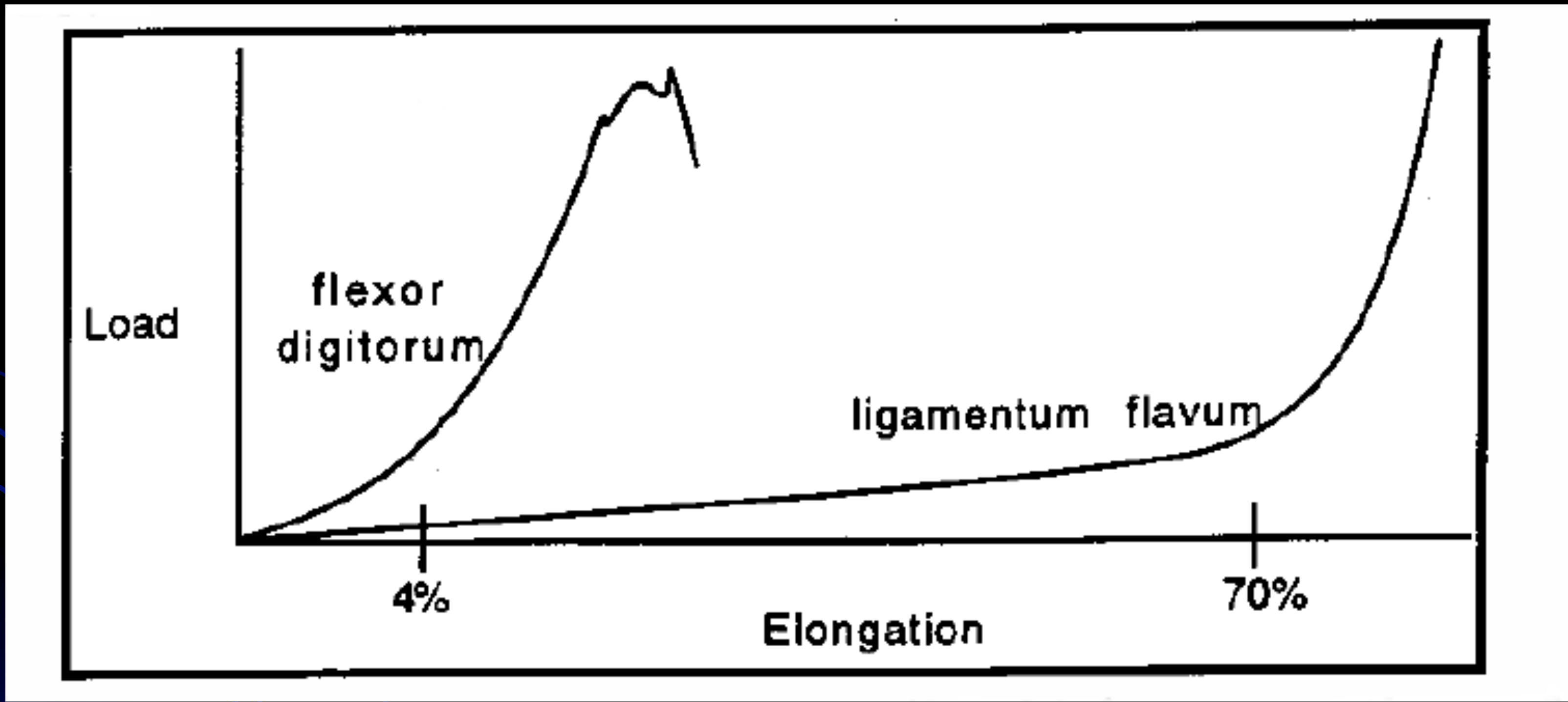
The 4 regions of a standard stress-strain curve obtained from a test with ligamentous tissue

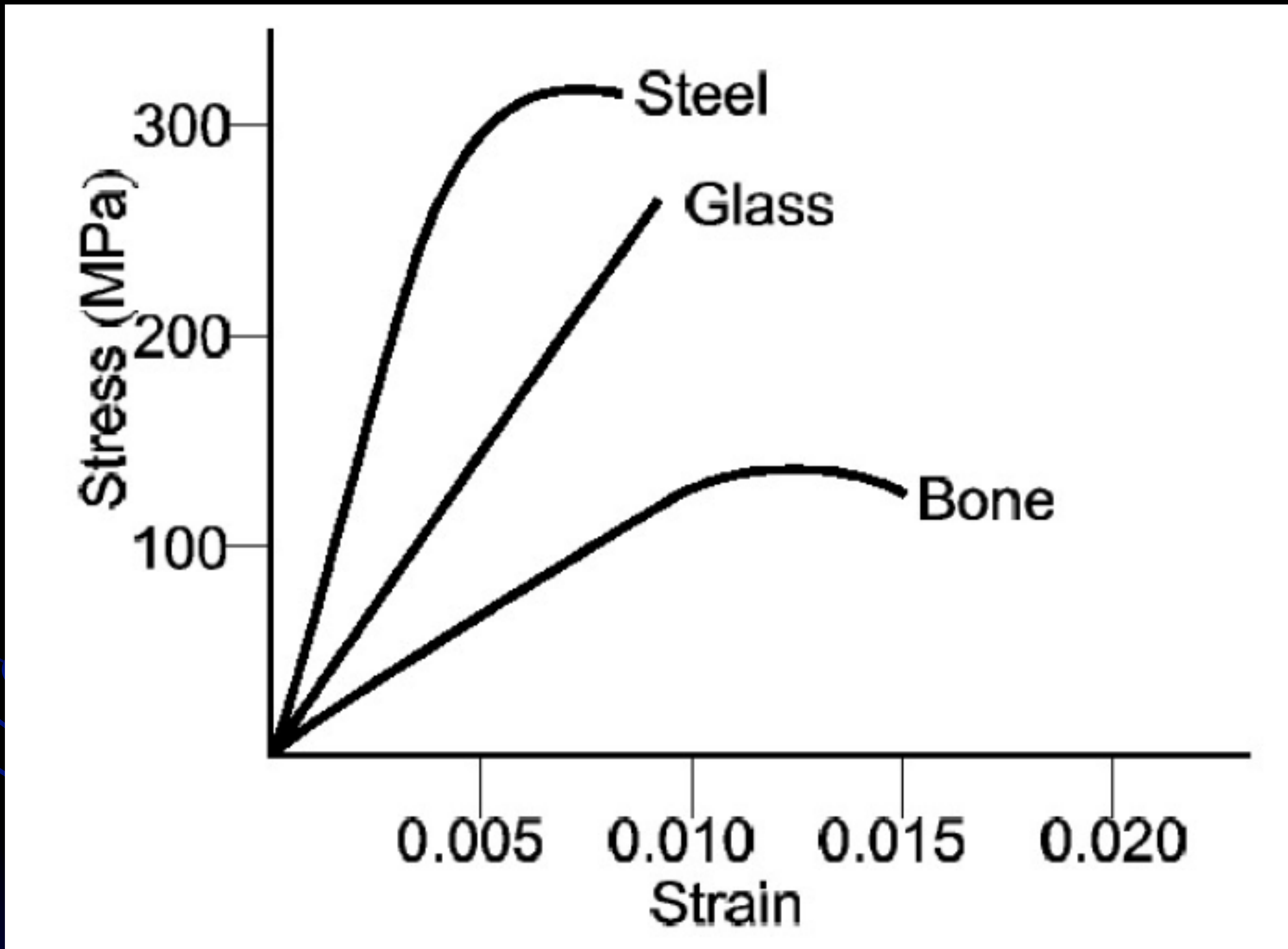


Region 1: Non-linear response due to the straightening of the crimp pattern.
Region 2: the crimp is lost and further deformation stretches the collagen fibers themselves.
Region 3: Microstructural damage occurs.
Region 4: Further stretching causes progressive fiber disruption and ultimately complete rupture.

Biomechanical Response

- Depends on the makeup of the tissue





Steel
Glass
Bone
300
200
100
Stress (MPa)
0.005
0.010
0.015
0.020
Strain

Stiffness

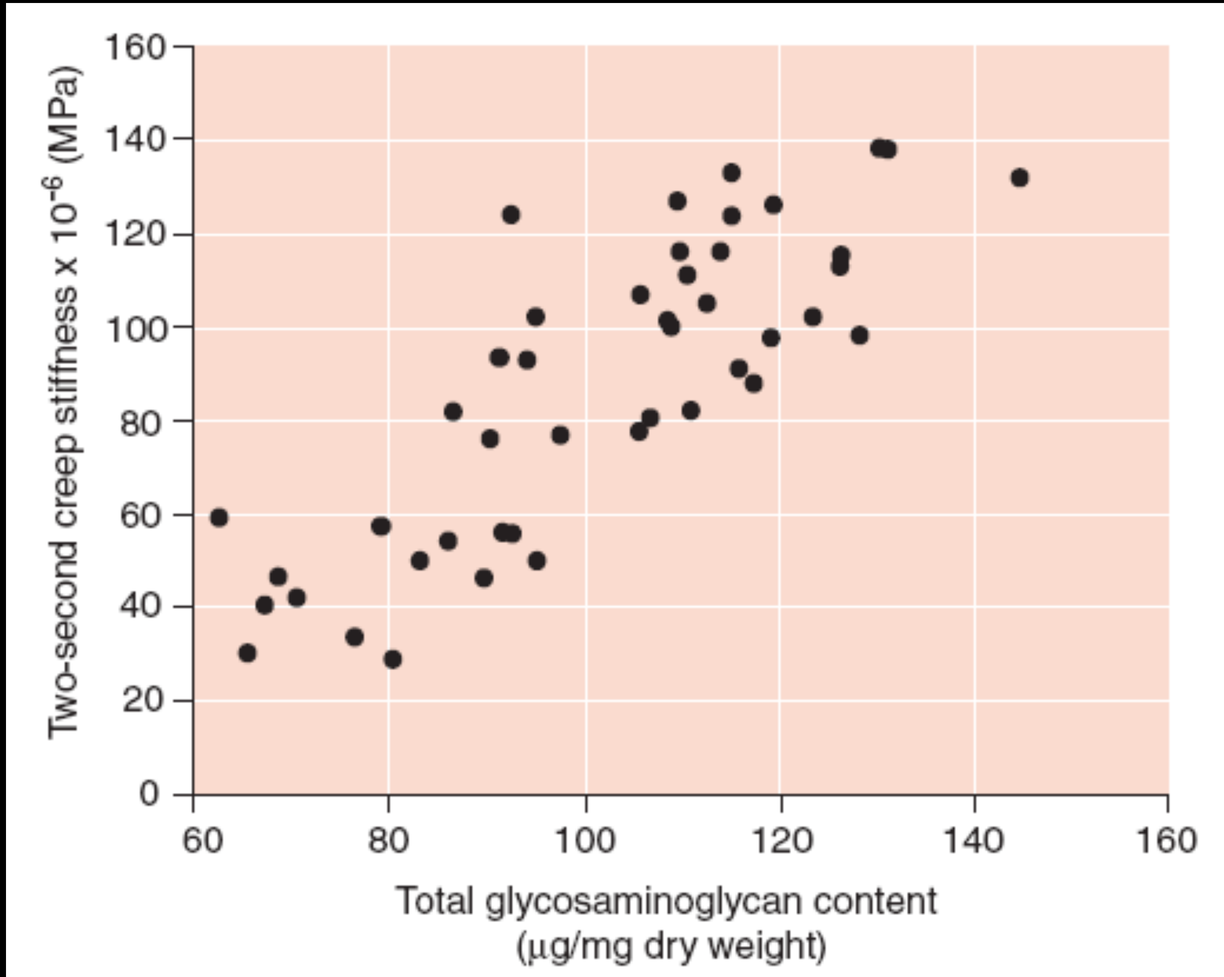
- the materials ability to resist deformation.

- the larger the stiffness, the greater the force required to cause a given deformation.

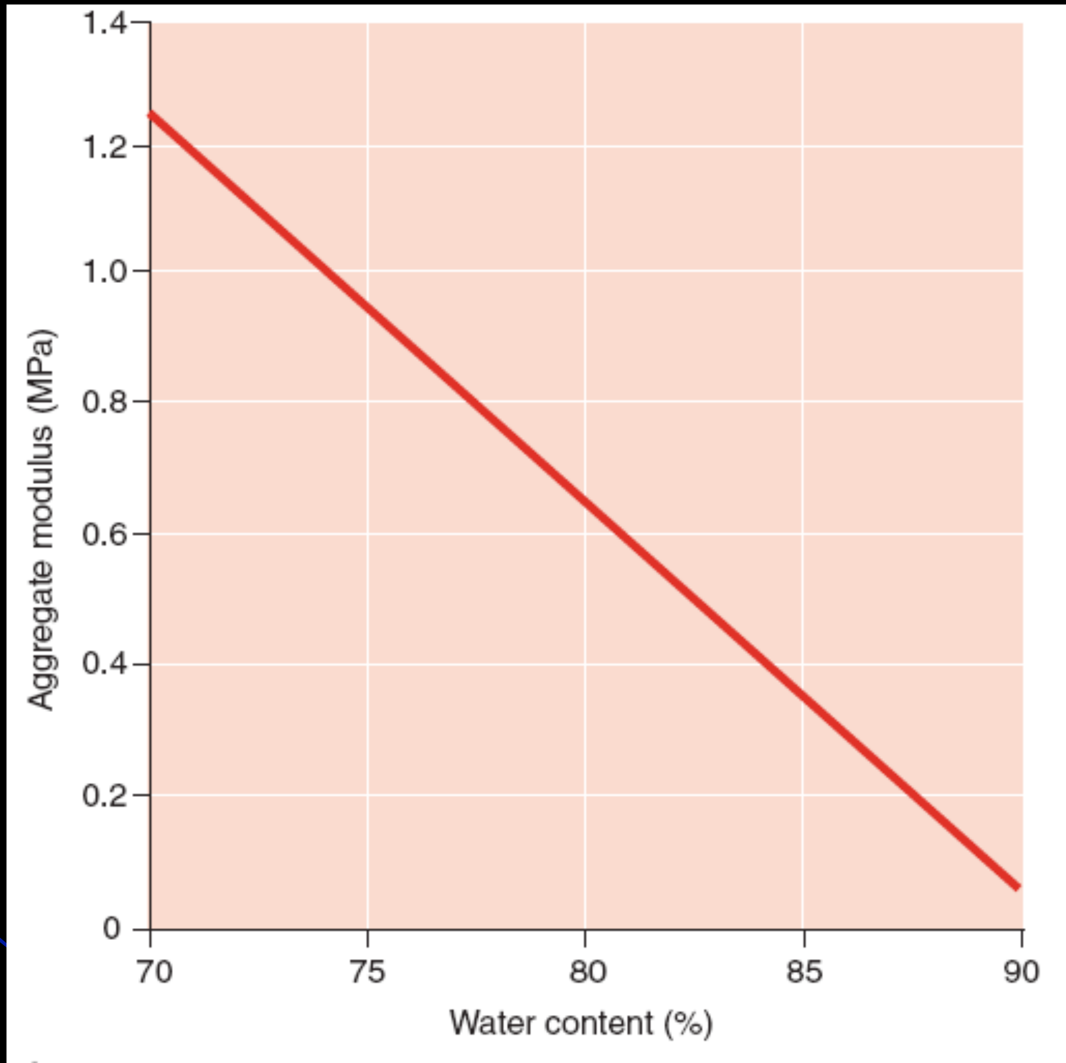
- commonly characterized by the slope of the linear region of a stress-strain curve, also referred to as Young's Modulus.

- to describe the slope of other regions of the stress-strain curve a Tangent Modulus is defined. If a Tangent Modulus is defined it should have associated with it a strain value or a range of strains.

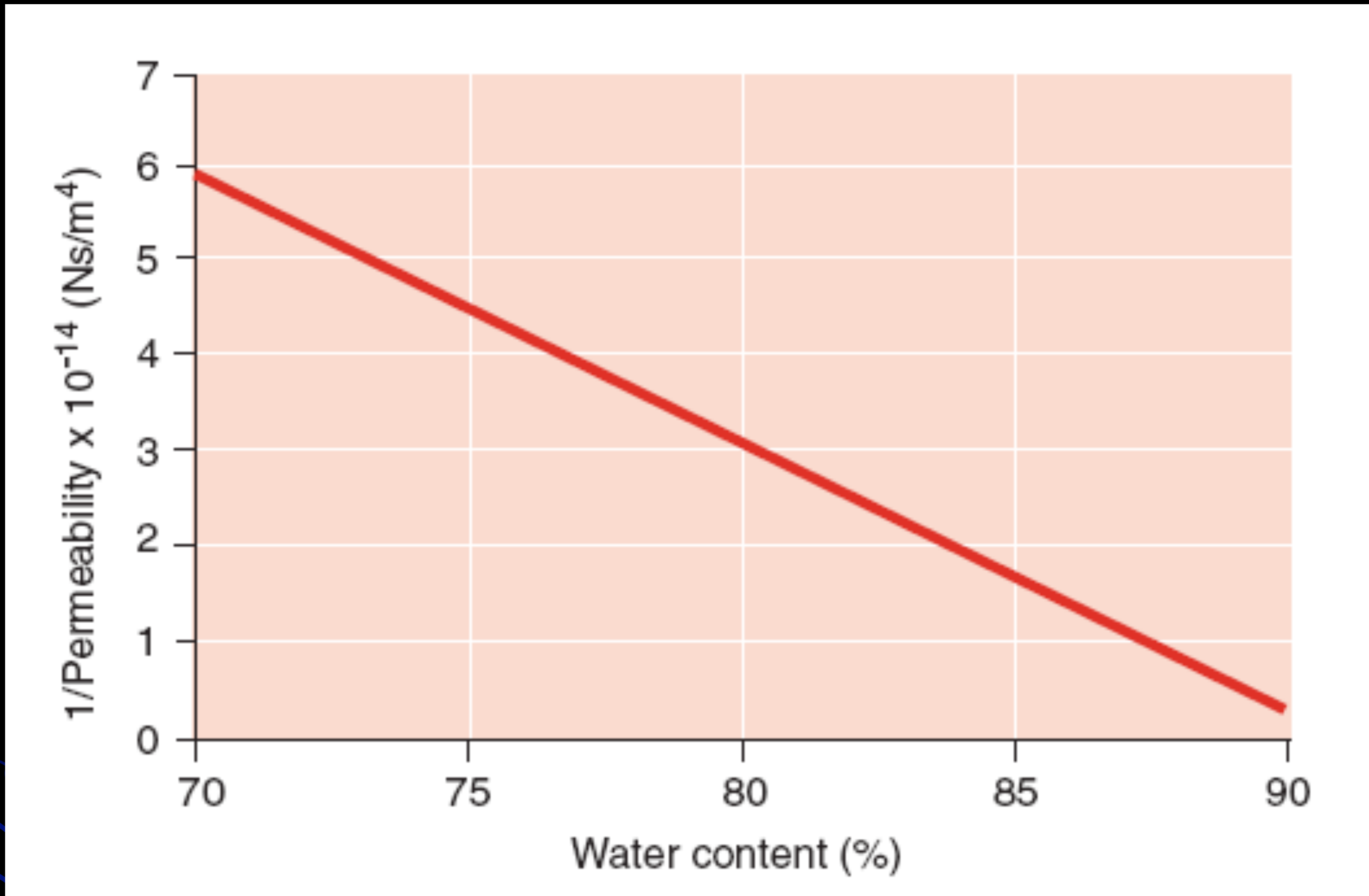
- there can be different modules depending on the loading conditions (e.g. shear modulus, compression modulus).



- The compressive stiffness of cartilage increases as a function of the total glycosaminoglycan content.
- There is no correlation of compressive stiffness with collagen content.

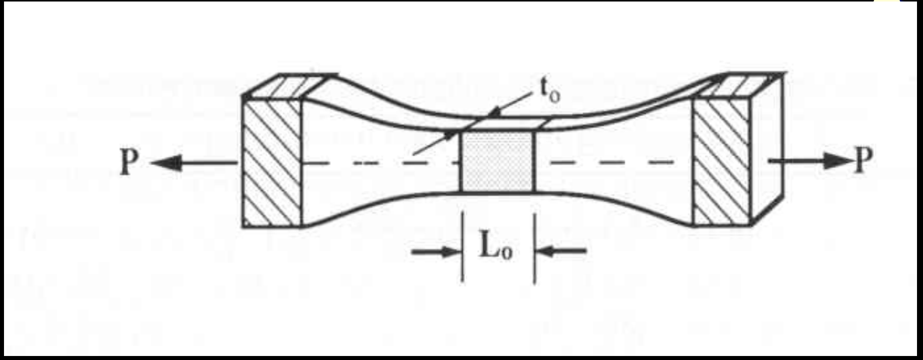


As the water content increases, cartilage becomes less stiff and more permeable



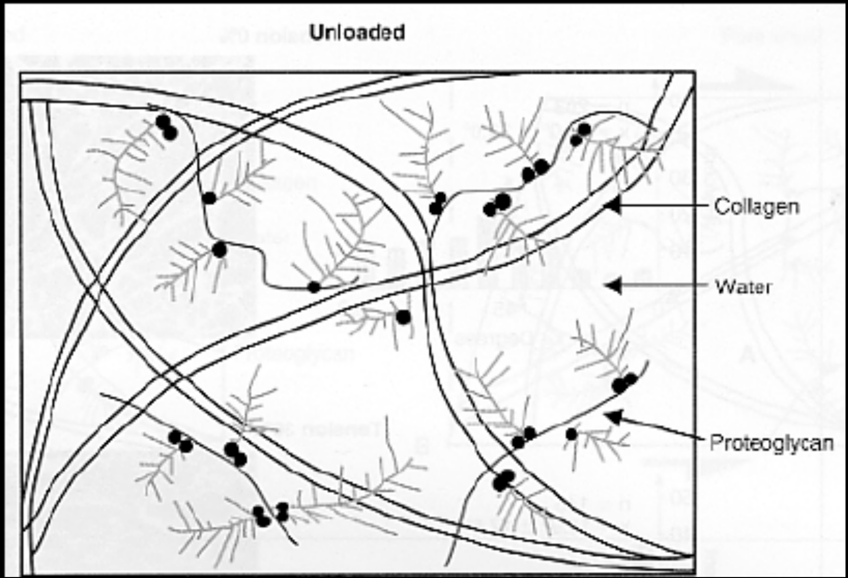
As the water content increases, the permeability increases.

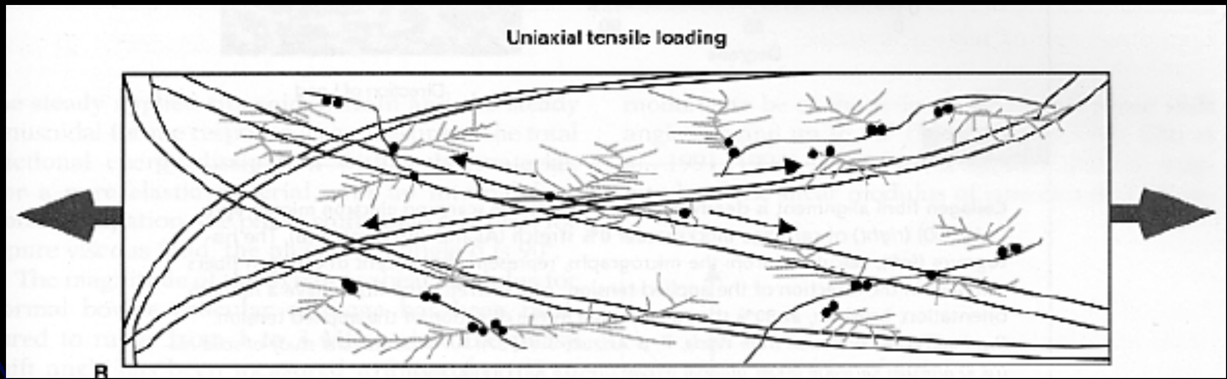
Tension



- Equilibrium Tensile Modulus (1-30 MPa)
  - Type of tissue
  - Age of animal
  - Type of joint
  - Sample location
  - Depth of sample (surface = 10MPa, Middle =4.5MPa)
  - Relative orientation
  - Biochemical comp/ molecular structure
  - State of degeneration (Normal =10MPa, OA=1.4MPa)

Tensile Force

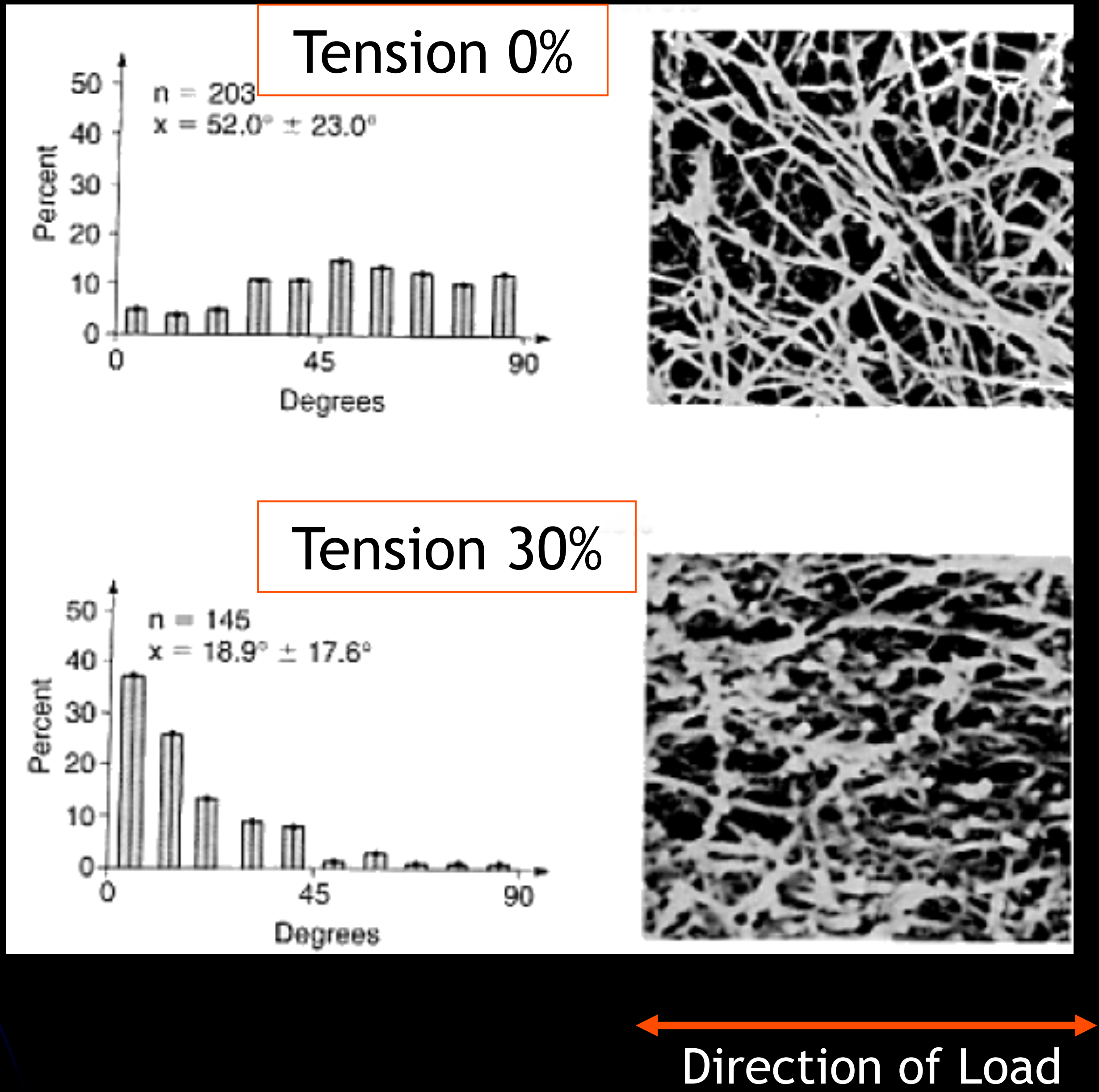




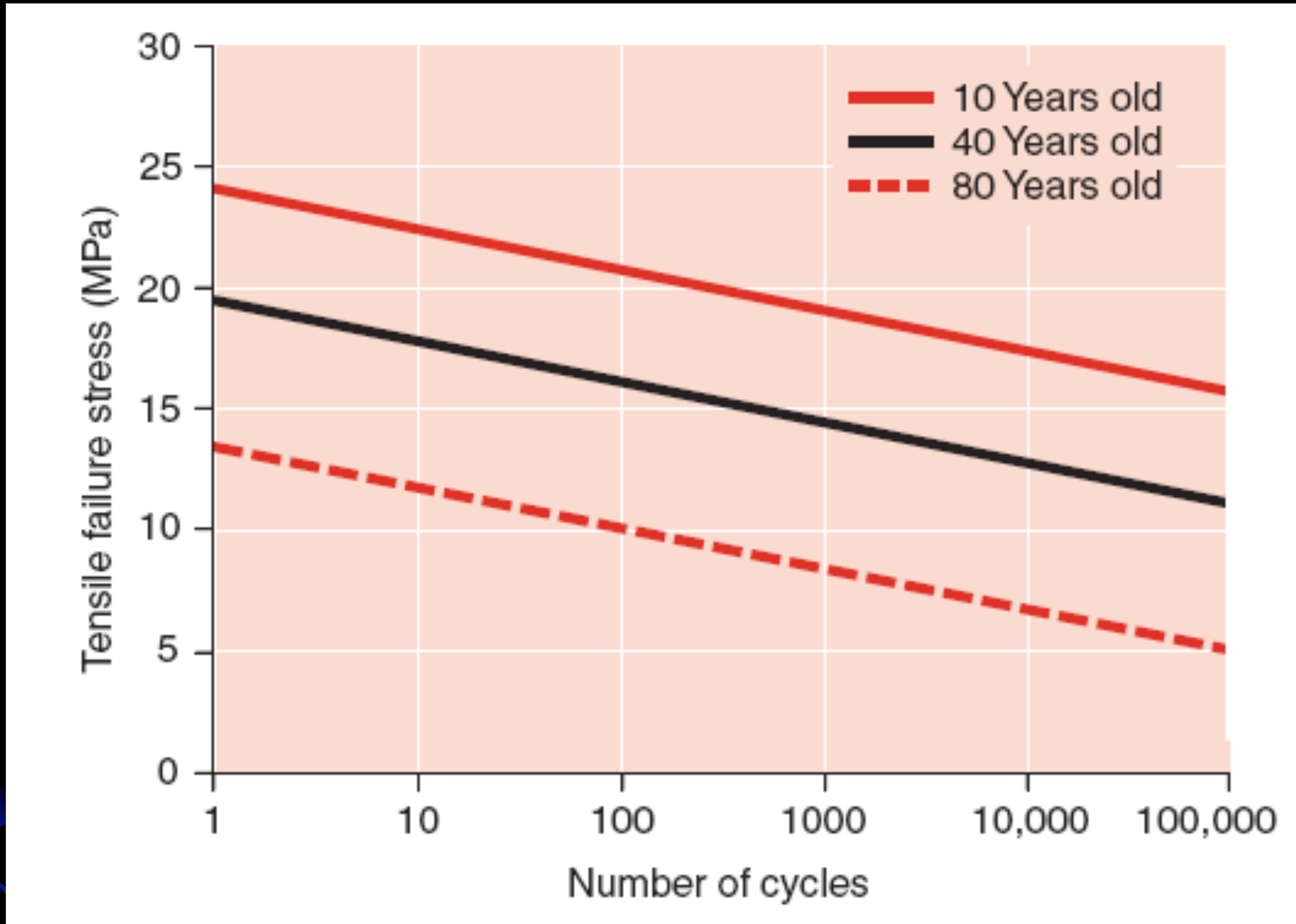
Tensile Force

a) Random alignment of collagen fibrils

b) Histogram of measurements made from micrograph show distribution of fibril alignment



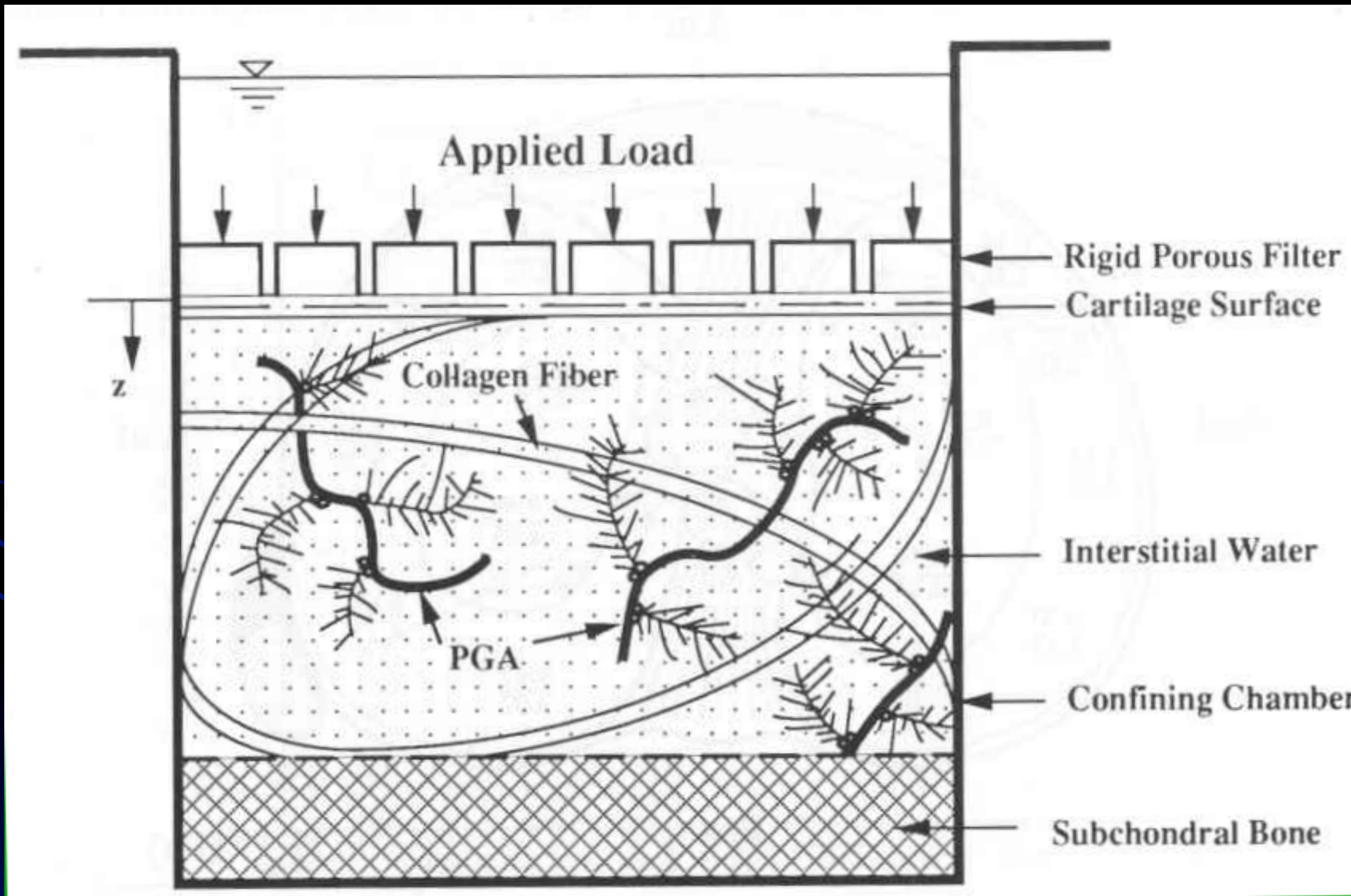
Repeated tensile loading and tensile strength of cartilage



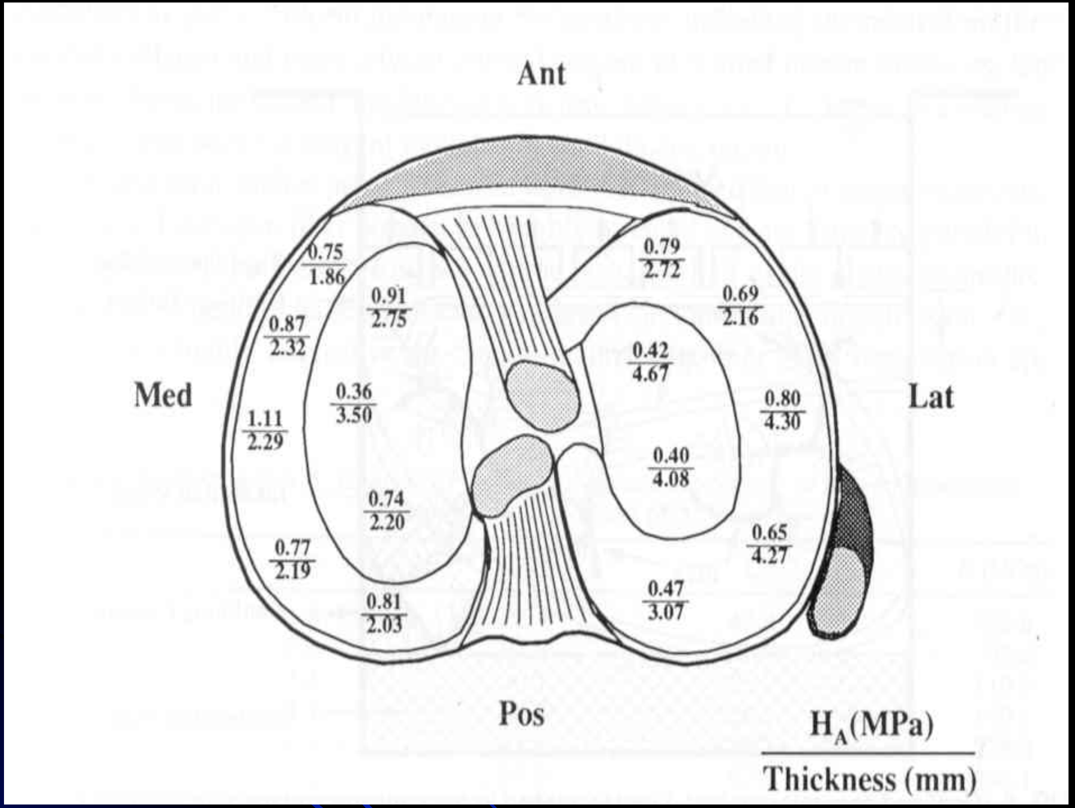
- As the tensile loading stress increases, fewer cycles of loading are needed to cause failure.
- Cartilage from older individuals fails at a lower stress than that from younger people.

Compression

Compressive Aggregate Modulus (H_A)(0.4-1.5 MPa)

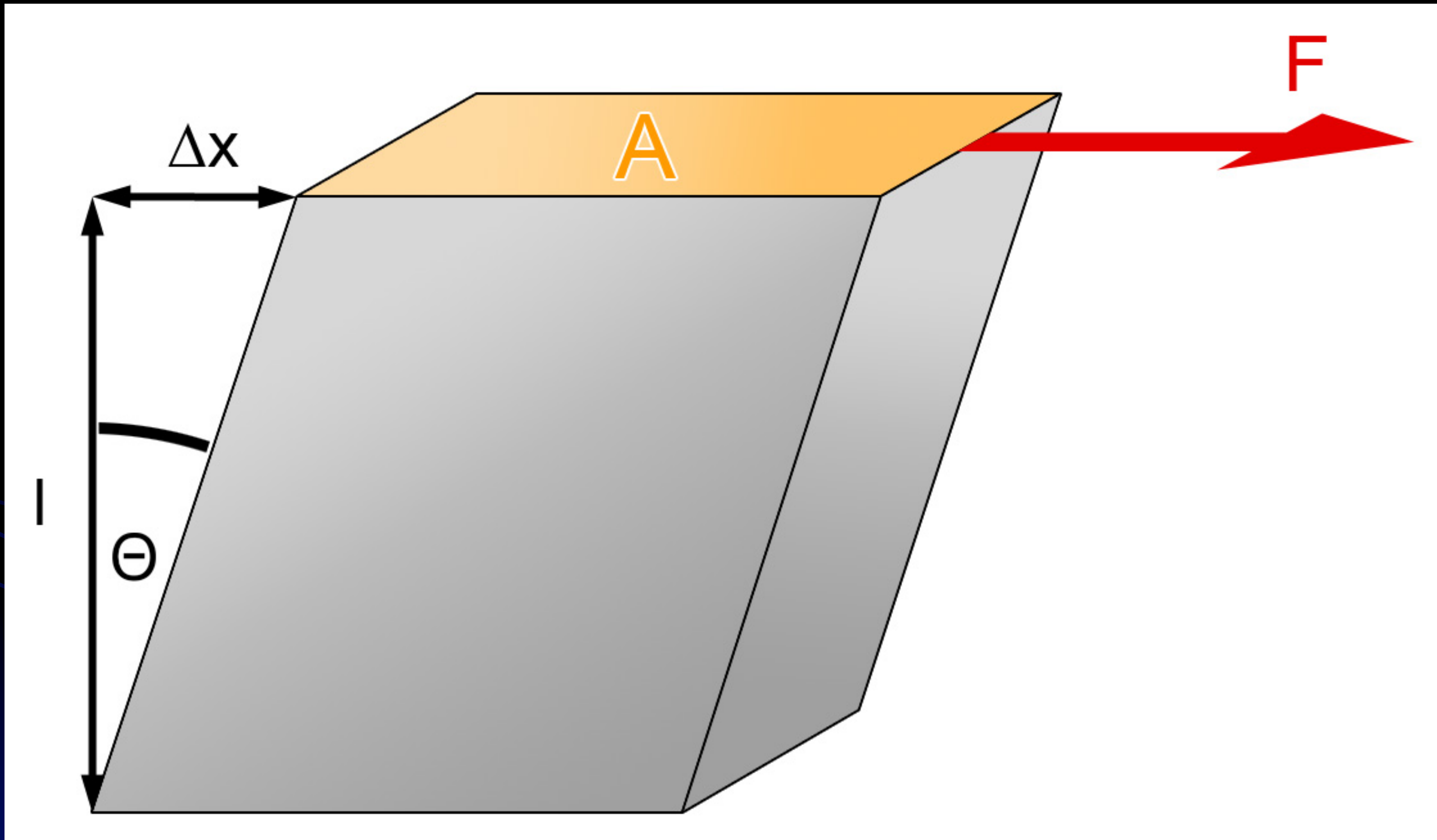


Compression

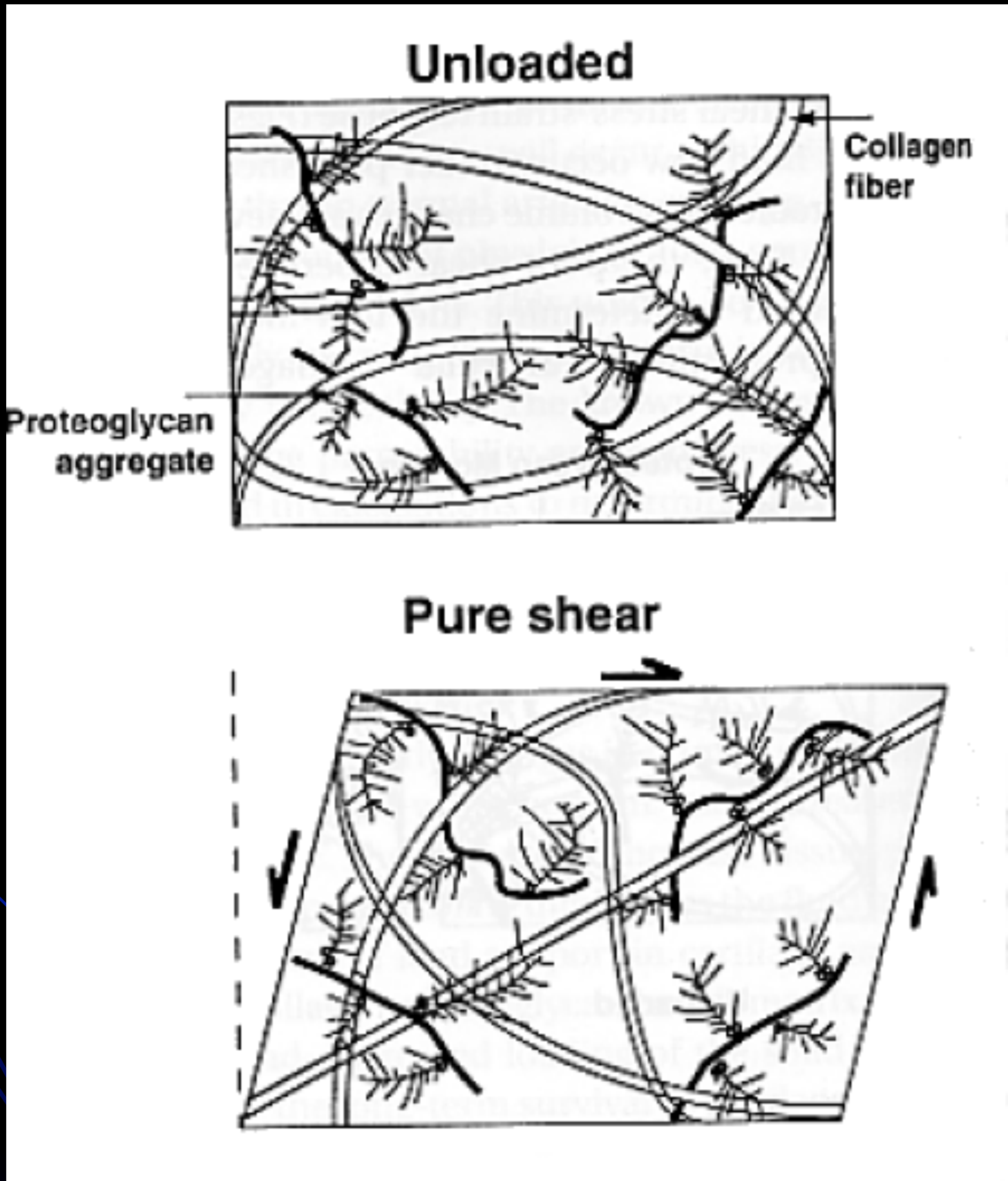


- H_A varies inversely with water content *OA patients have increased water
- H_A varies directly with PG content
- Not dependent on collagen content

Shear

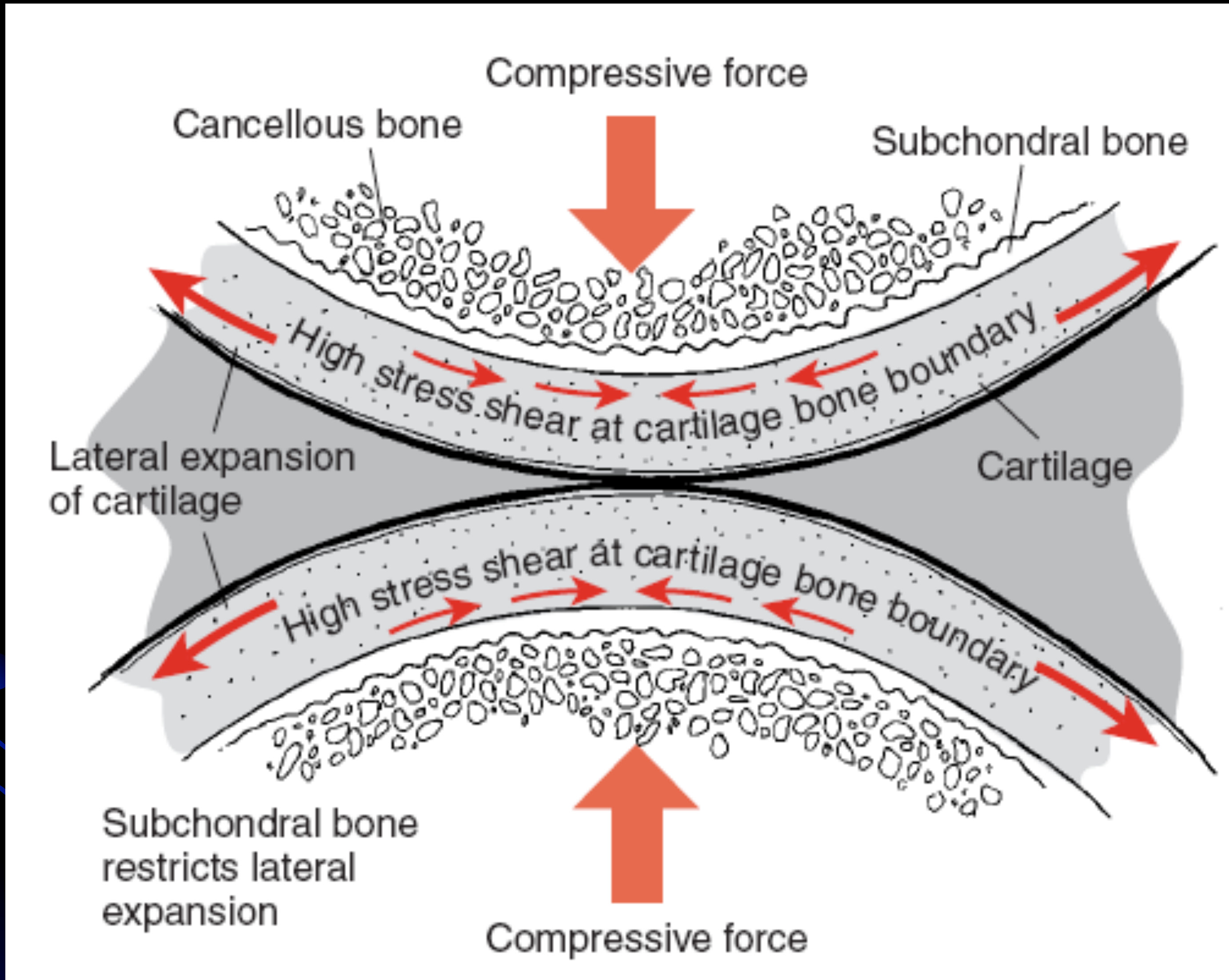


Shear Force



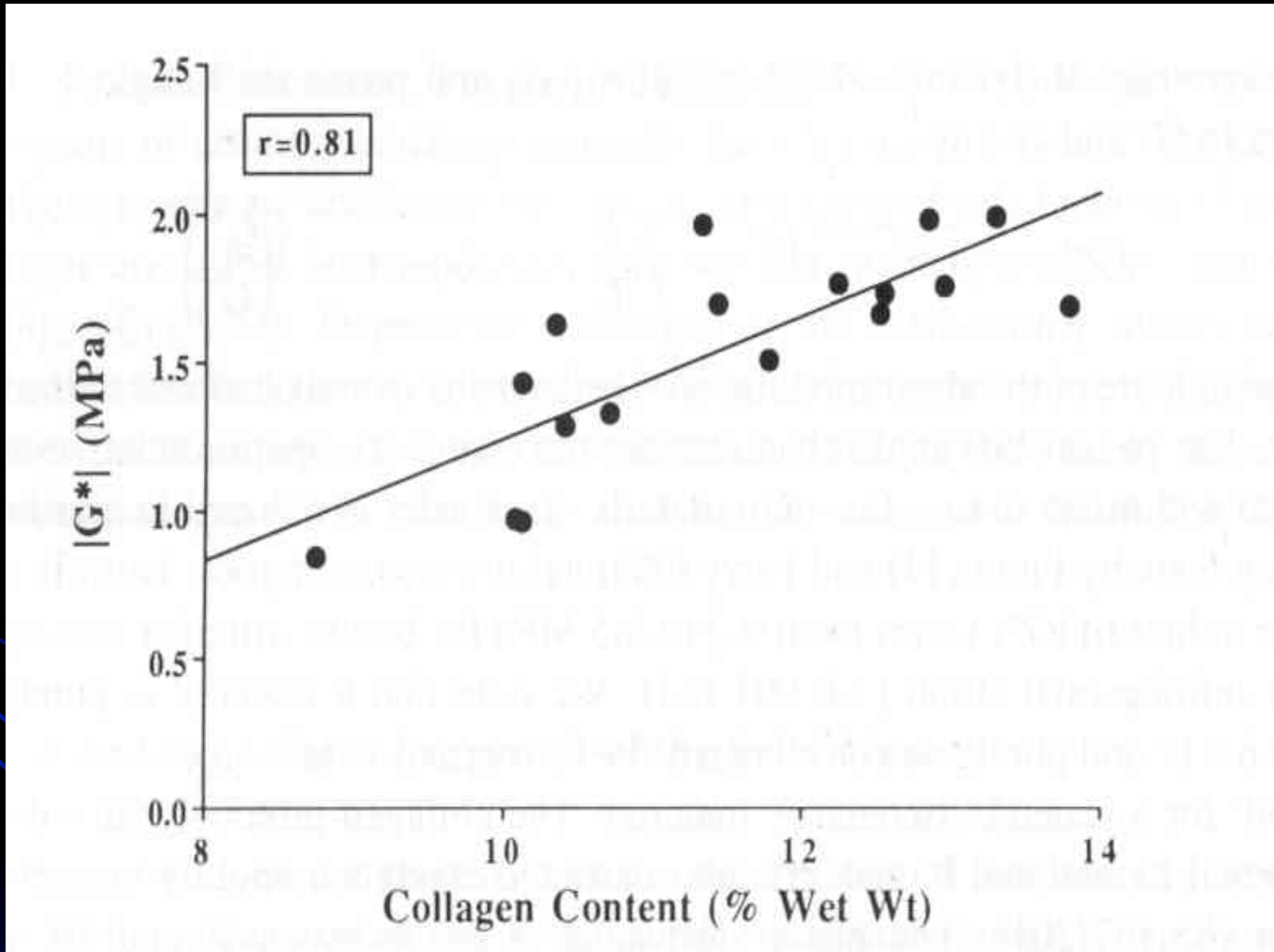
Unloaded
Collagen fiber
Proteoglycan aggregate
Pure shear

Shear



Compressive force
Cancellous bone
Subchondral bone
High stress shear at cartilage bone boundary
Lateral expansion of cartilage
Cartilage
High stress shear at cartilage bone boundary
Subchondral bone restricts lateral expansion
Compressive force

Shear



Viscoelasticity

Viscoelastic Properties

- Biological tissues are viscoelastic materials.

- Their behavior is time and history dependent.

- The stress response of the material depends on both the strain applied and the strain rate at which it was applied.

- A viscoelastic material possesses characteristics of *stress-relaxation, creep, strain rate sensitivity, and hysteresis*.

Examples

Biological materials
Polymer plastics
Metals at high temperatures

Cartilage: Biphasic Theory

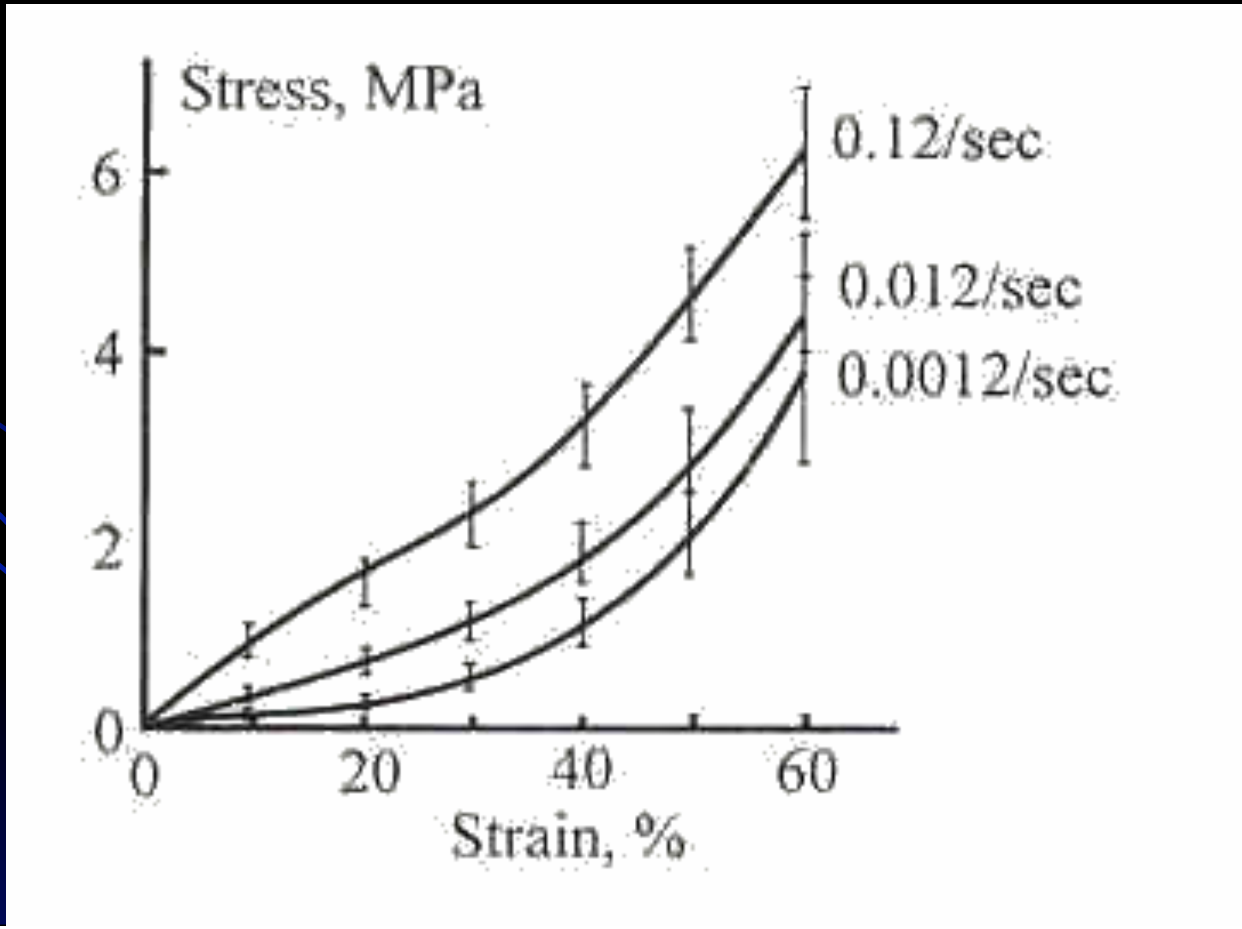
- liquid phase flows under the influence of pressure gradients produced by mechanical deformations

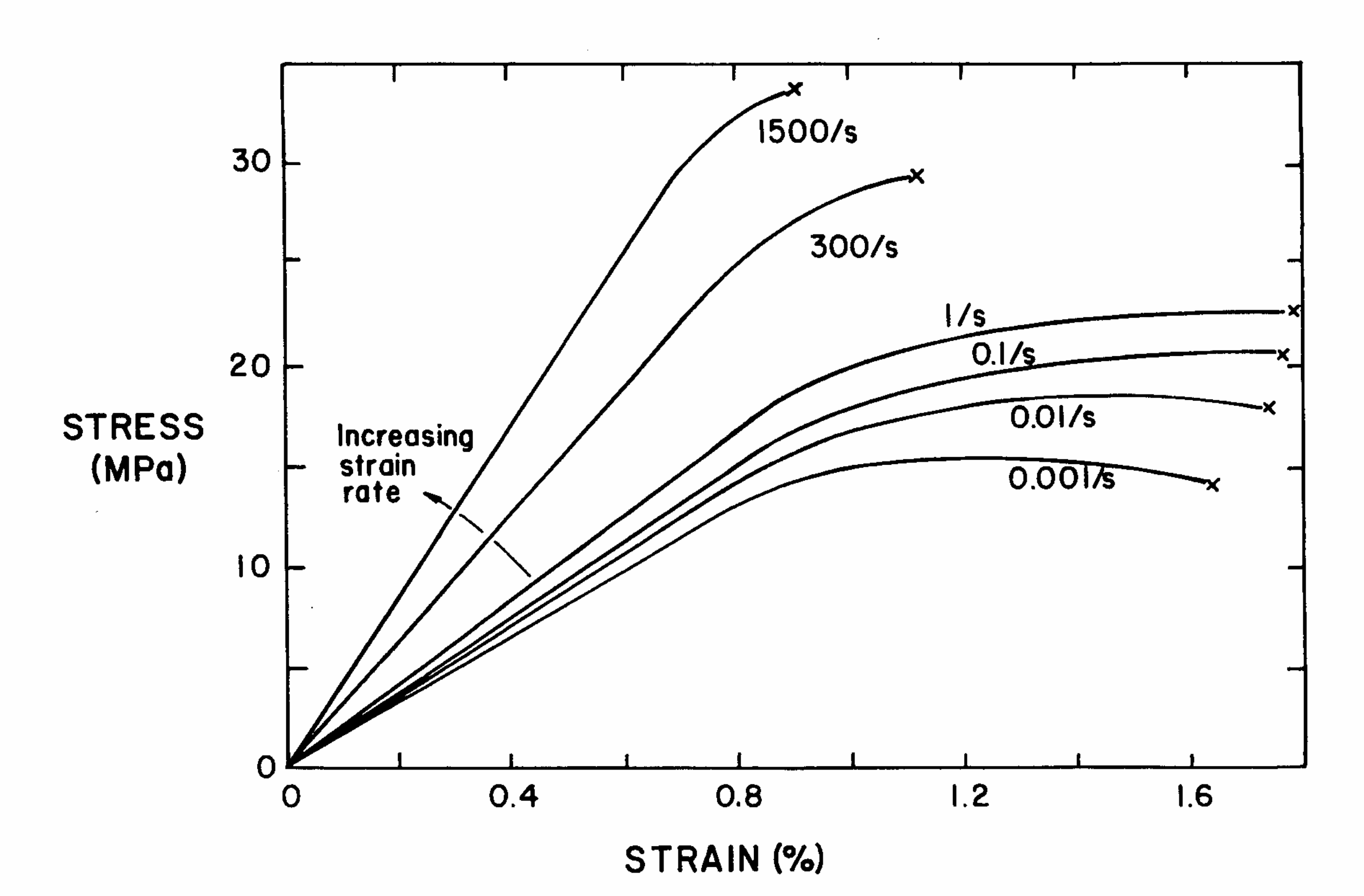
- movement of fluid phase contributes to viscoelasticity of cartilage

Cartilage: Viscoelasticity

Stress-strain curves for cartilage at 3 strain rates.

- modulus at slow rate → 1-2 MPa
- modulus at high rate → 500 MPa!





Stress as a function of strain and strain rate for human compact bone

Viscoelasticity

Flow independent

- Polymeric matrix molecules reorient at their own pace
- No volume changes

Flow dependent (biphasic)

- Water is forces out of the tissue like a sponge
- Very small pores increase time
- Requires volume changes

The viscoelastic response of articular cartilage relies on 2 different physical mechanisms:

(1) the intrinsic viscoelastic properties of the macromolecules that form the organic solid matrix

(2) the frictional drag from the flow of the interstitial fluid through the permeable solid matrix.

Each mechanism promotes the overall viscoelastic response of cartilage under tension, compression, and shear.

Material Properties

- Viscoelastic behavior are dominated by frictional drag of interstitial fluid flow through the porous collagen-proteoglycan solid matrix, thus causing viscous dissipation

Injury and Strain Rate

Under a slow, sustained force, articular cartilage is able to respond accordingly and accommodate this load.

Under a concentrated force, articular cartilage is unable to react to the load, and therefore, the tissue is vulnerable to injury.

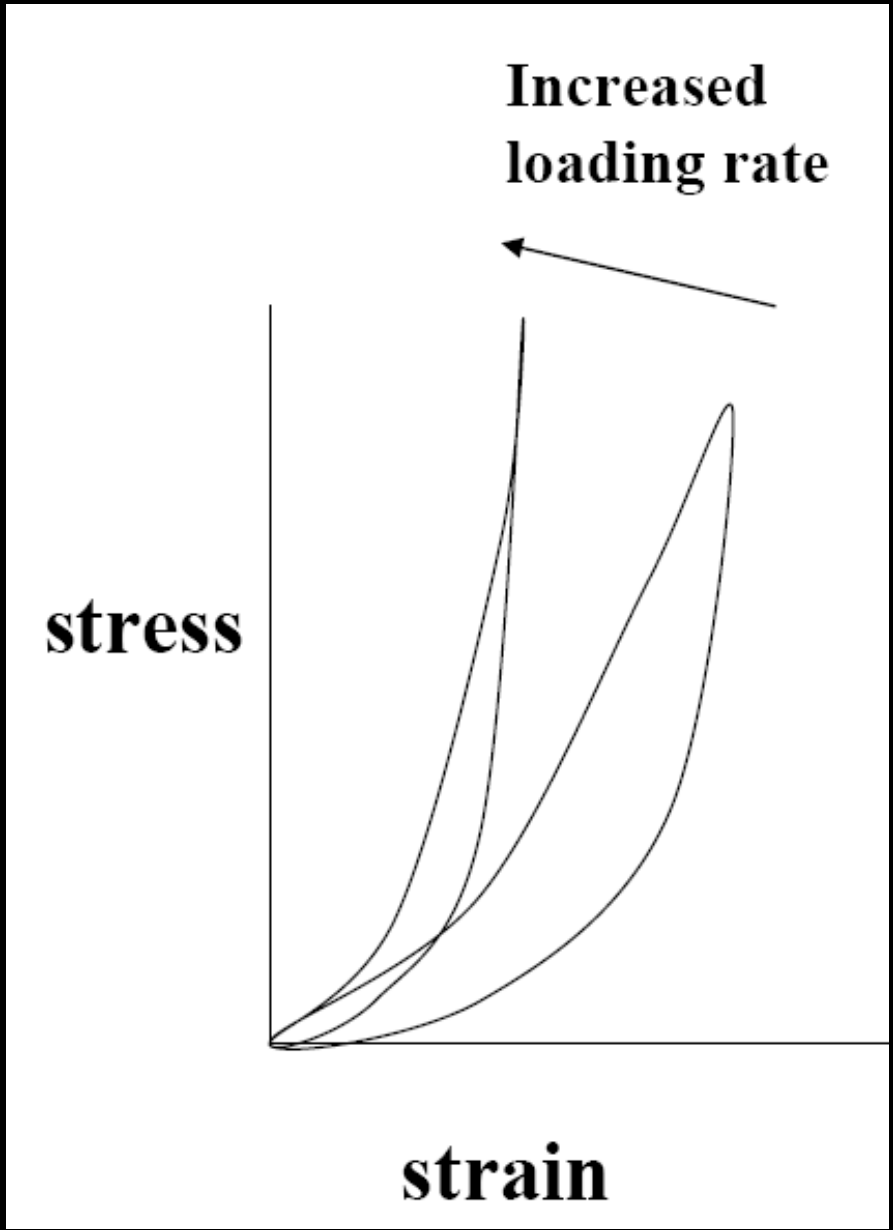
Hysteresis

The loading and unloading curves obtained from a force-deformation test of biological tissues do not follow the same path.

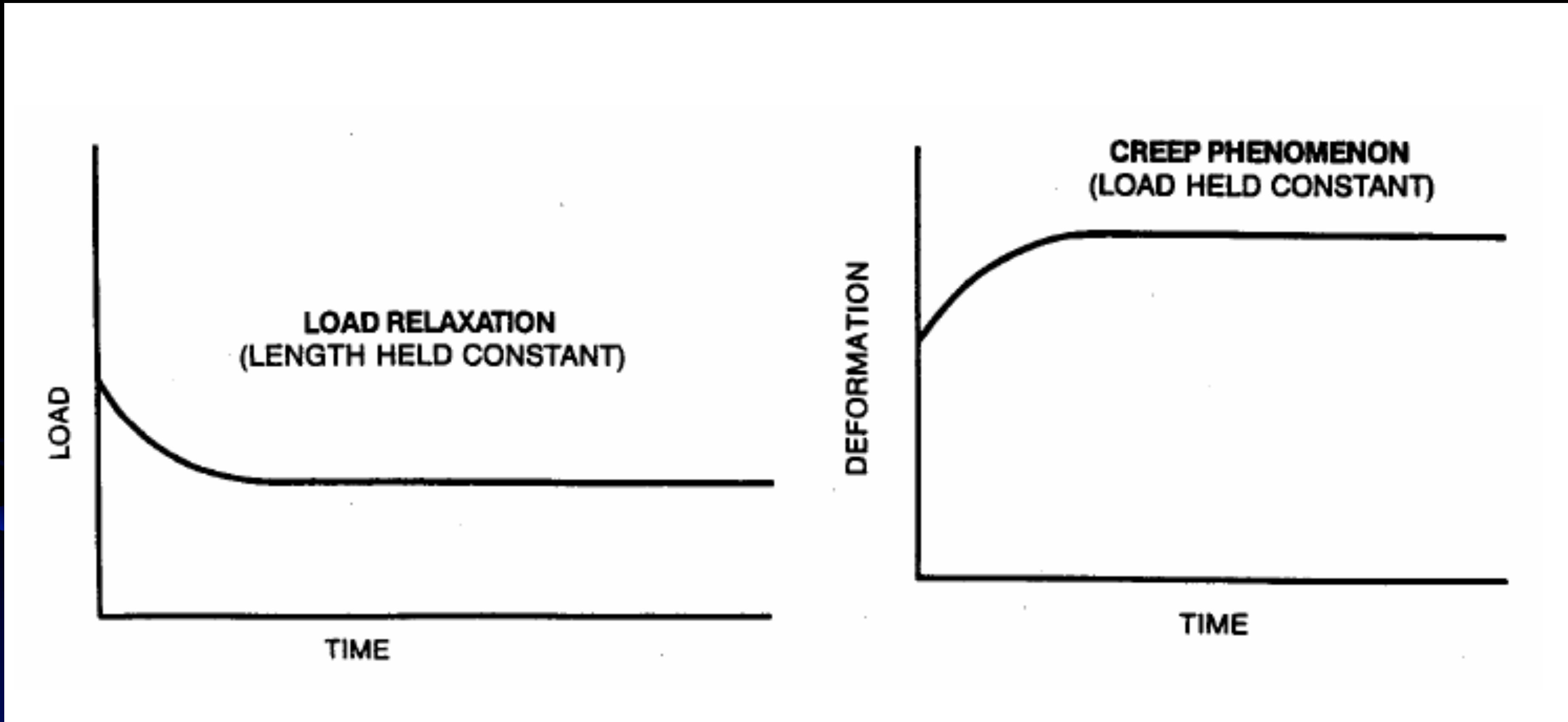
The difference under the loading and unloading curves is termed the area of hysteresis and represents the energy lost due to internal friction in the material.

The amount of energy liberated or absorbed during a tensile test is defined as the integral of the force and the displacement.

Hence the maximum energy absorbed at failure equals the area under the force-displacement curve.



Stress-Relaxation and Creep



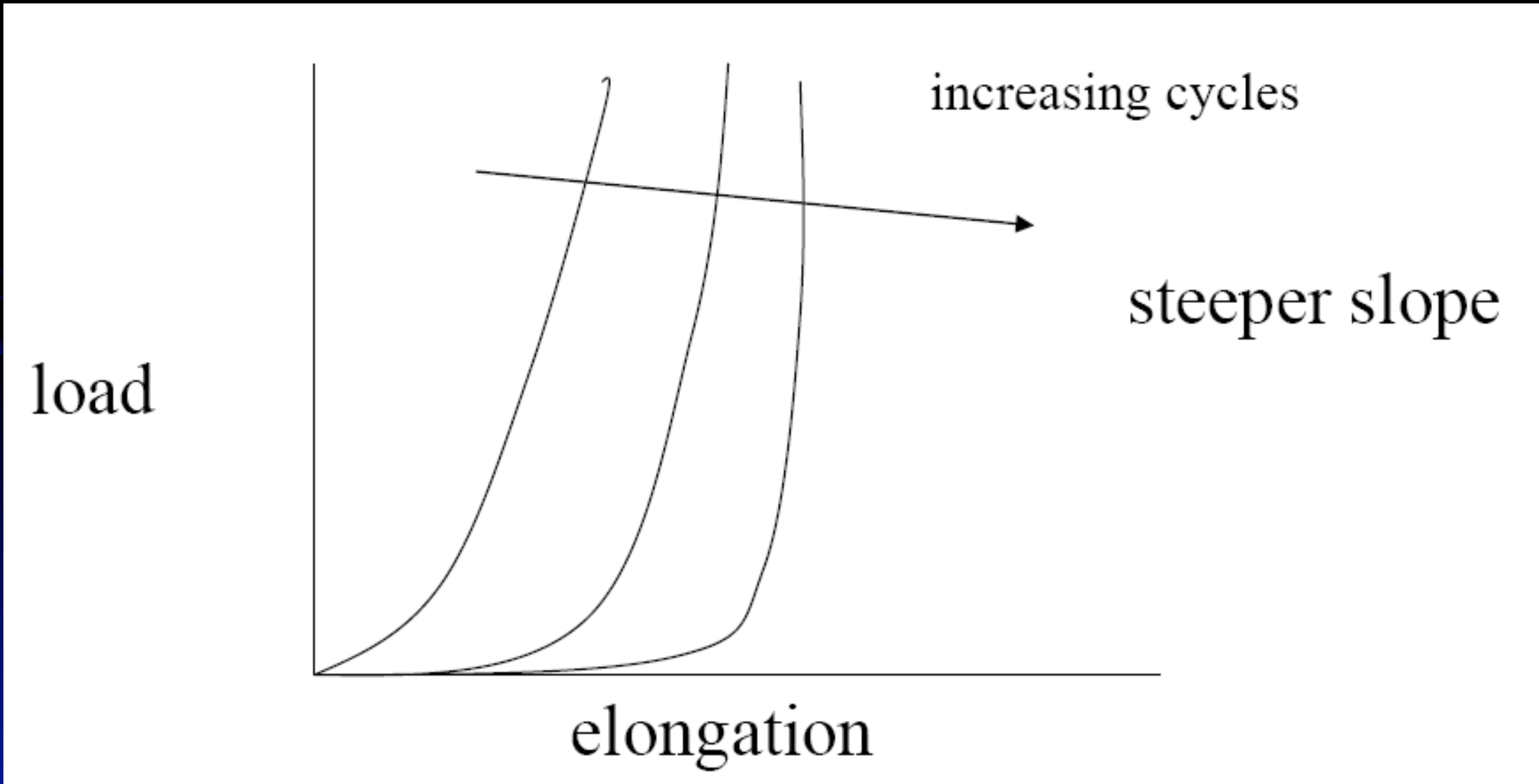
Stress-Relaxation and Creep

- Examples of taking advantage of the long-term viscoelastic response
  - Child with a club foot treated with a plaster cast
  - Brace for a scoliosis spine

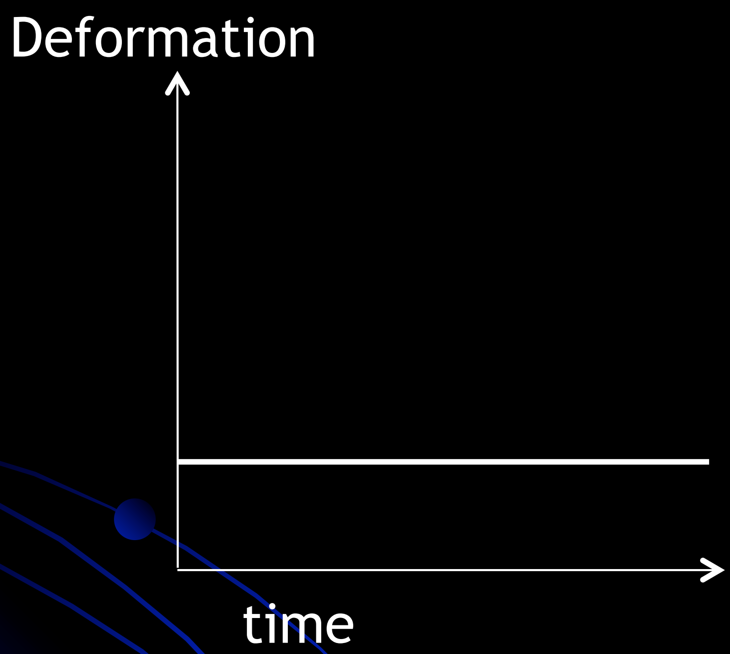
- Gradual then a faster change over time

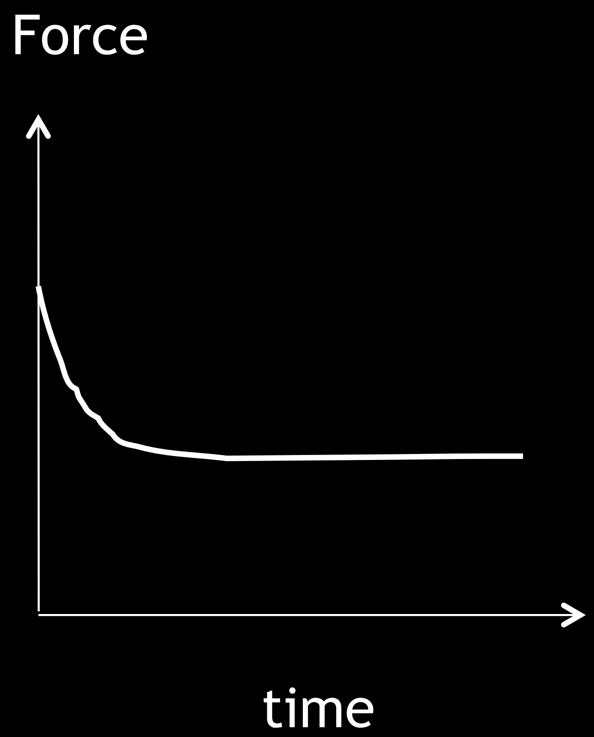
Biomechanical Properties

- With an increased number of loading cycles, the following has been observed *in vitro*
- strong load time dependence



Tensile Stress Relaxation Test





Stress -Relaxation

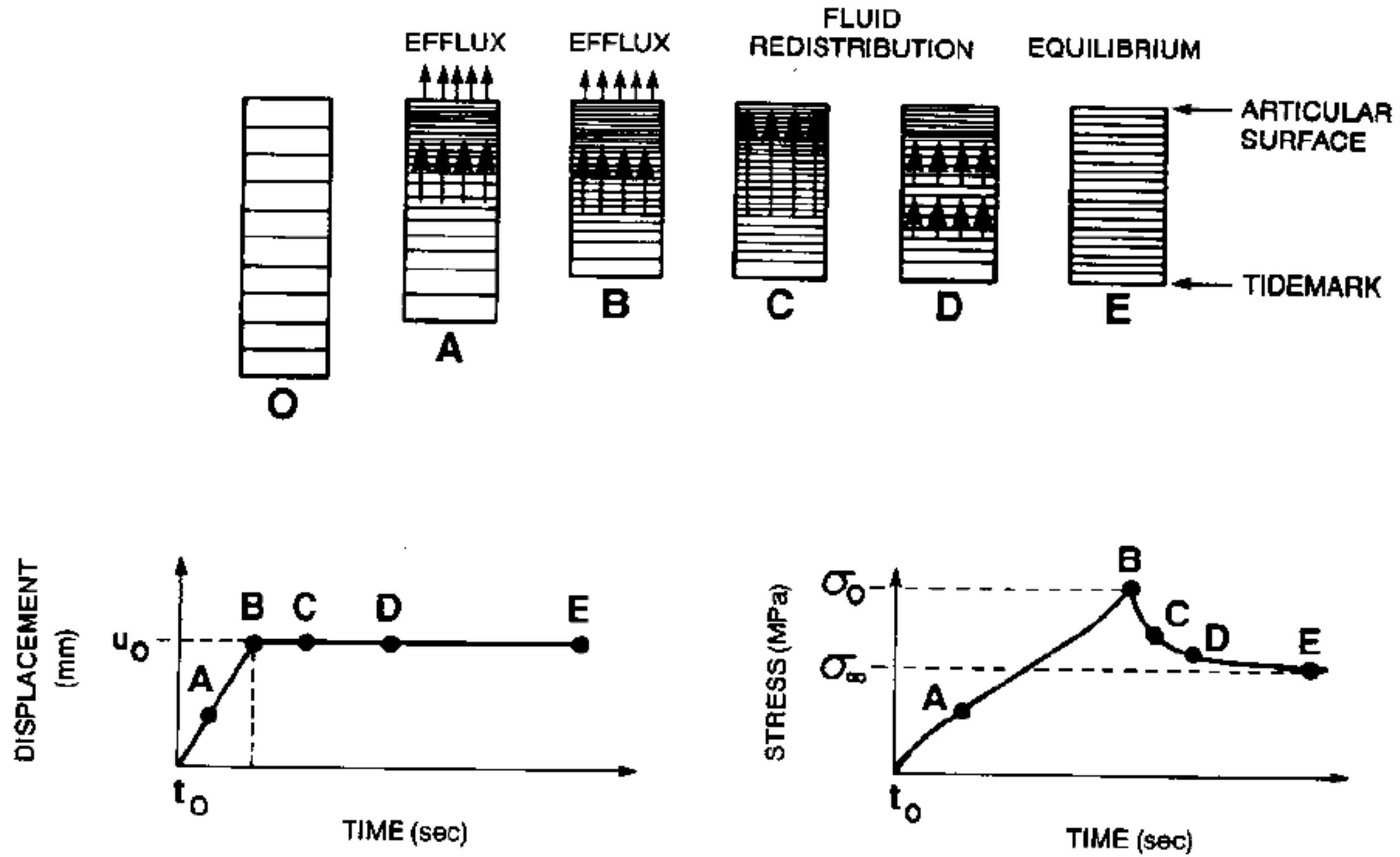
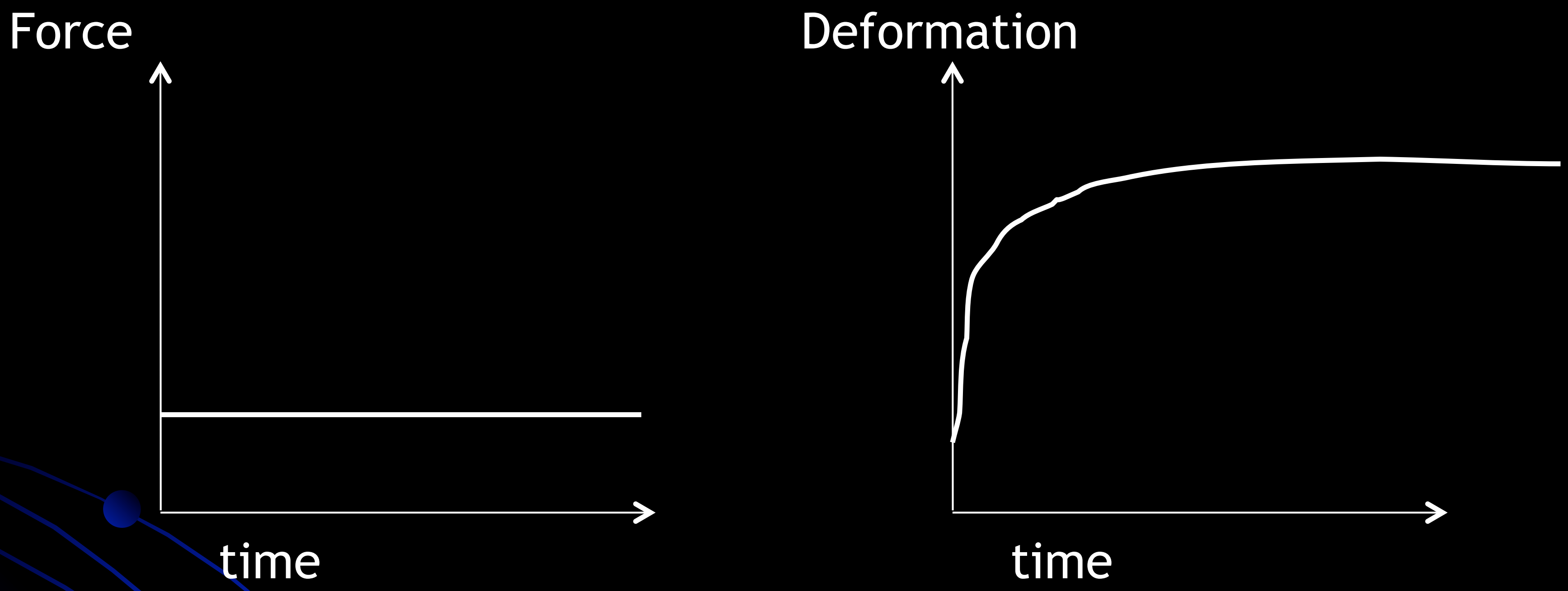


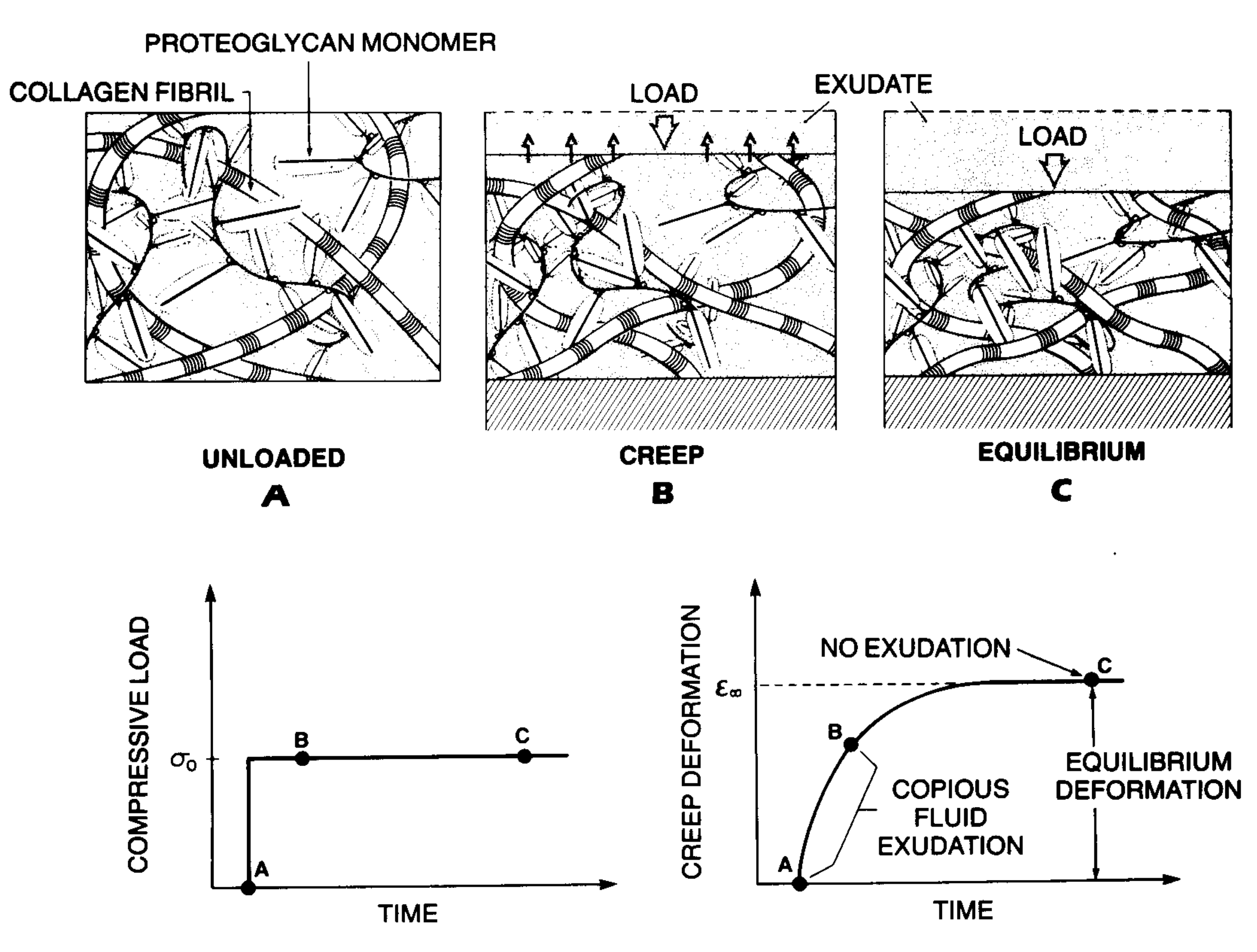
FIG. 2–10

Confined Compression Creep Test



- a constant force is applied across the tissue.
- the displacement of the tissue is measured as a function of time.
- The general shape of the displacement-time curve depends on the past loading history (e.g. peak force, loading rate).

Biphasic Creep Response of Articular Cartilage in Compression



Biphasic Creep Response of Articular Cartilage in Compression

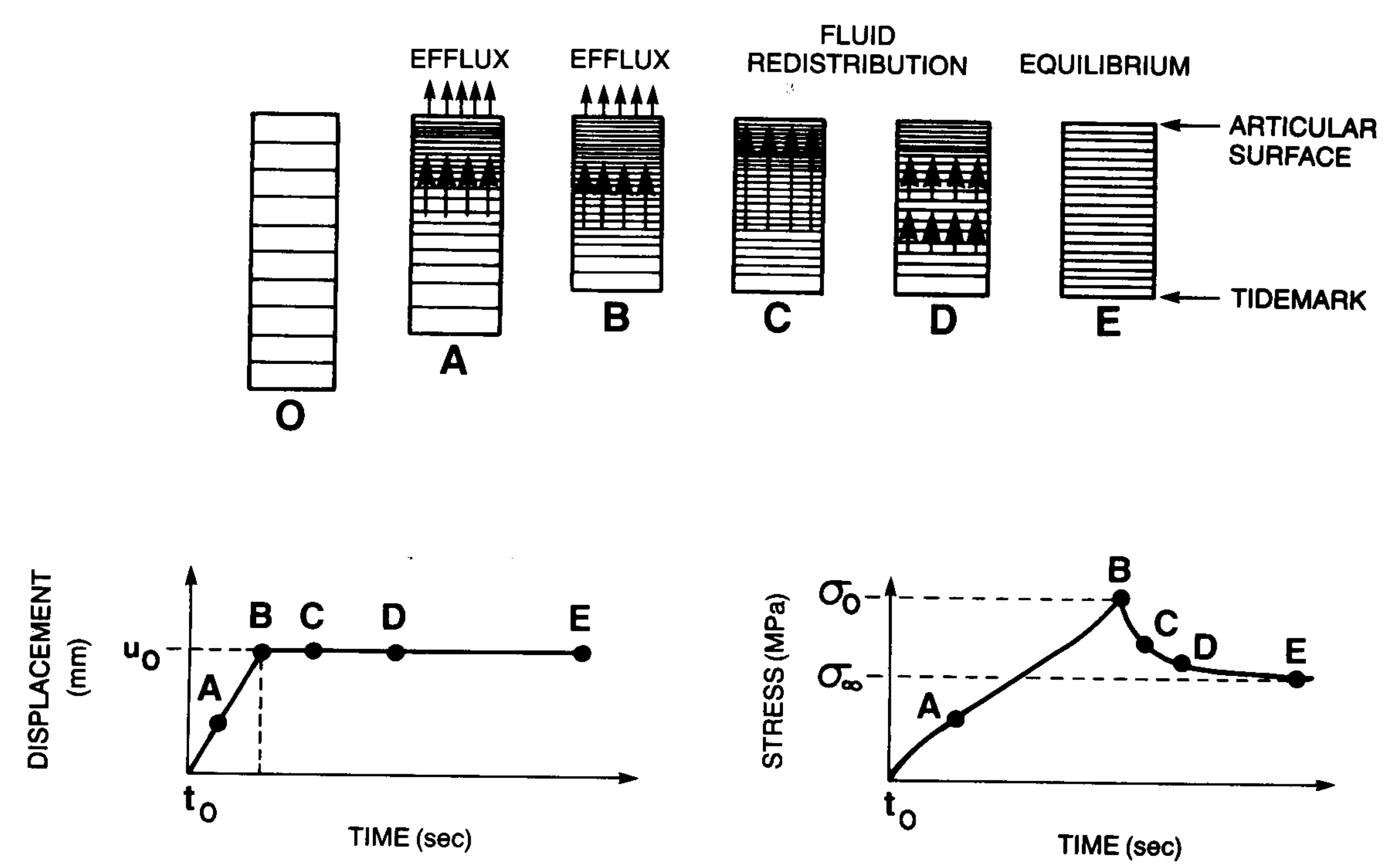
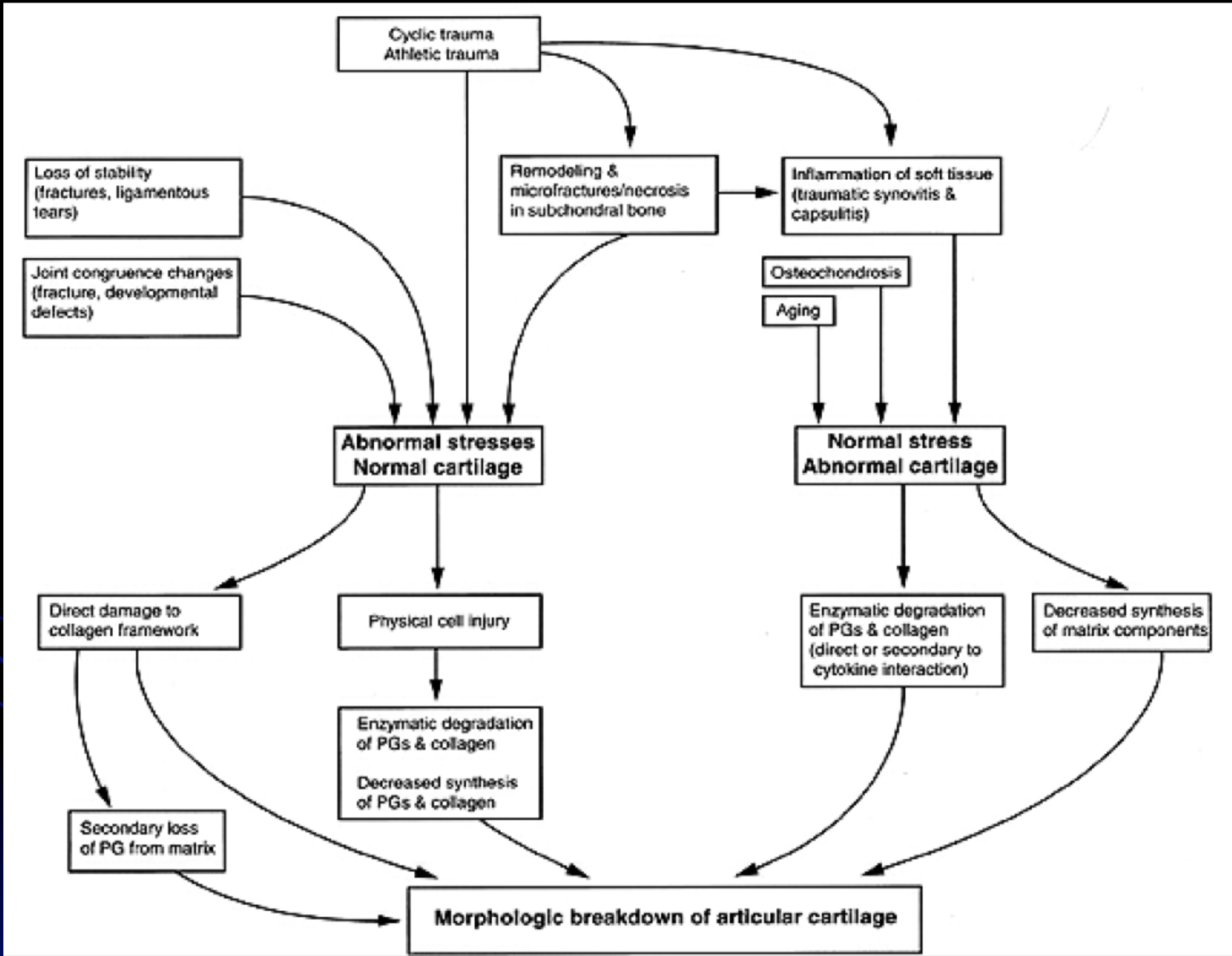


FIG. 2–10

Pathology of Articular Cartilage

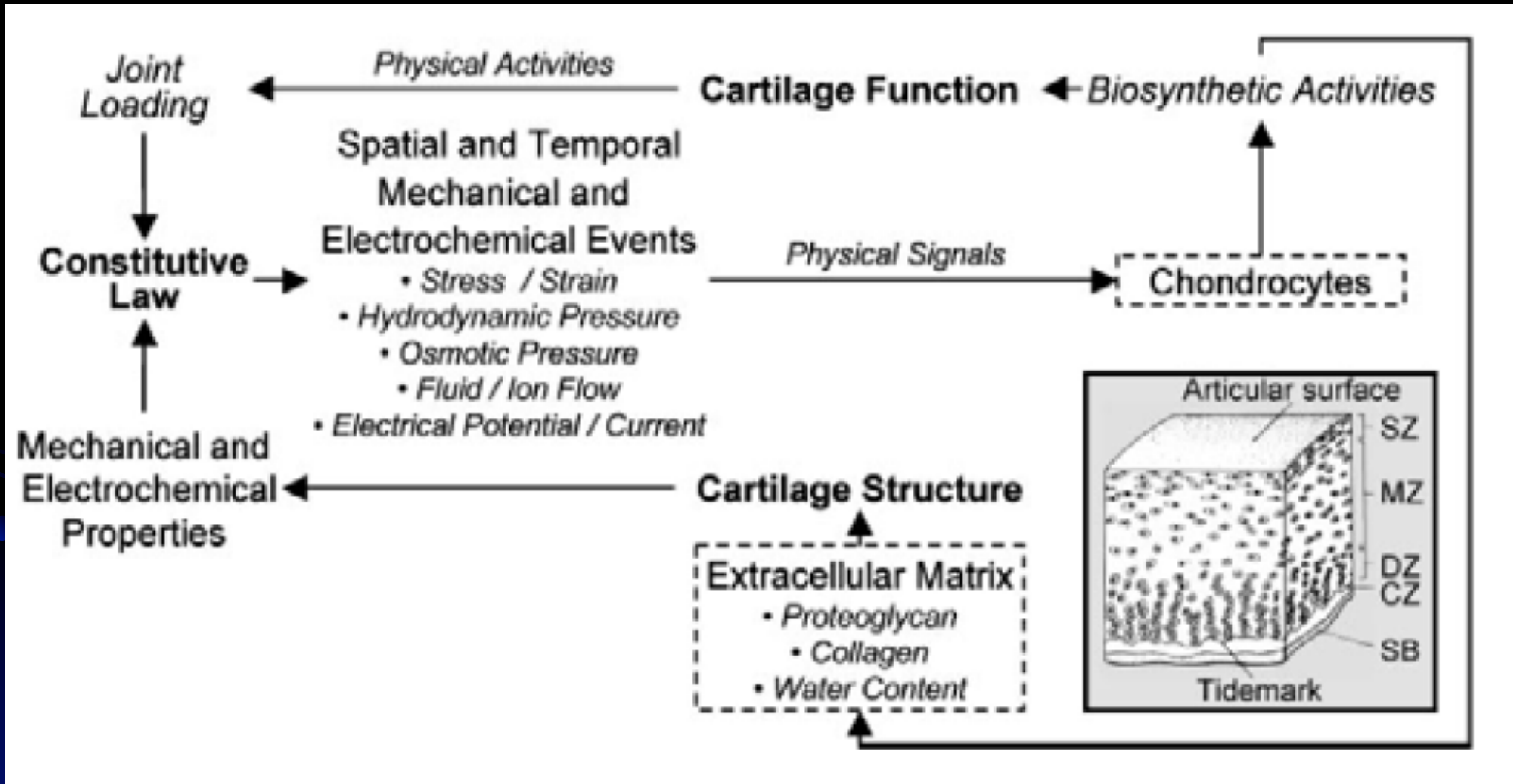

Cyclic trauma
Athletic trauma
Loss of stability (fractures, ligamentous tears)
Remodeling & microfractures/necrosis in subchondral bone
Inflammation of soft tissue (traumatic synovitis & capsulitis)
Joint congruence changes (fracture, developmental defects)
Osteochondrosis
Aging
Abnormal stresses
Normal cartilage
Normal stress
Abnormal cartilage
Direct damage to collagen framework
Physical cell injury
Enzymatic degradation of PGs & collagen (direct or secondary to cytokine interaction)
Decreased synthesis of matrix components
Enzymatic degradation of PGs & collagen
Decreased synthesis of PGs & collagen
Secondary loss of PG from matrix
Morphologic breakdown of articular cartilage

Effect of training

- Strenuous training
- softening and thinning of articular cartilage properties

- Moderate running training
- positive effects on the articular cartilage properties: thicker, stiffer and higher PG concentration

Immobilization and Cartilage



Effects of joint immobilization (animal studies)

biochemical changes

. proteoglycan concentration decreases 20 to 23%

. collagen concentration stays constant

. collagen crosslinks reduced with 13% .

biomechanical changes

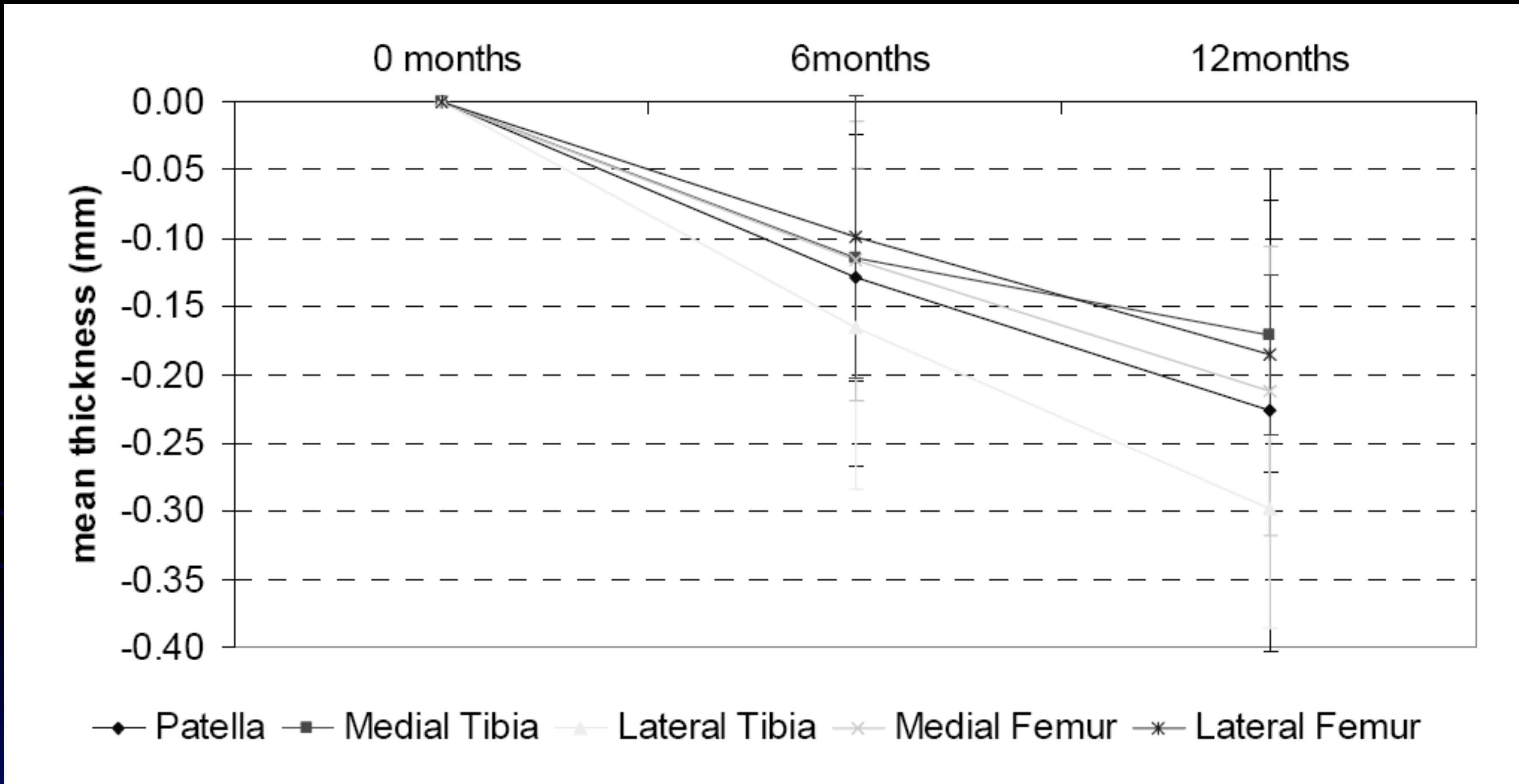
. equilibrium shear modulus (ESM) increased

. instant shear modulus didn‘t change . thickness alterations

. overall decrease of 9% (femur, tibia and patella)

. at the summit of medial condyle of the femur: -20%

Longitudinal Analysis of Cartilage Atrophy in the Knees of SCI Patients



Articular Cartilage: Aging

It is difficult to differentiate between the changes of senescence and the changes of degenerative processes

- Some degeneration of the knee joint is observable in every subject beyond the age of 15
- However, changes seen in thirty-year olds may not be seen in eighty-year olds so it is difficult to attribute changes to aging
- The majority of changes are observed in weight-bearing portions of articular cartilage that are not covered by the menisci

Articular Cartilage: Aging

- In some cases, knee cartilage thins rapidly or remains morphologically normal
- There is no evidence that thickness of articular cartilage changes with age
- Most common changes seen are: non-progressive fibrillation, change in color (yellow-brown tinge), thickness or biomechanical properties
- In most cases, articular cartilage appears visually intact but demonstrates significant biomechanical changes

Cartilage: Biomechanics of Aging

- The number of collagen fibers stays the same but may become thinner and finer resulting in altered mechanics
- There is a decrease in:
- Static tensile fracture strength
- Static stiffness
- Tensile fatigue strength

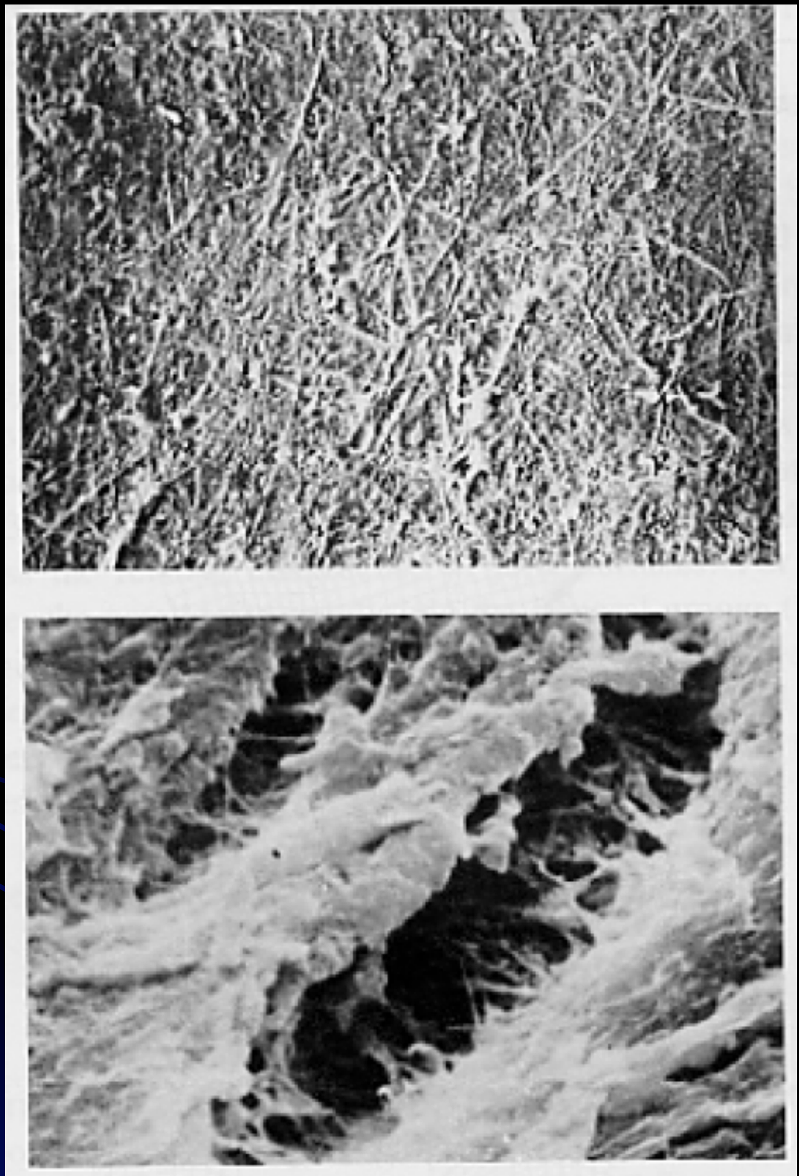
B. Tensile Fatigue Failure Properties

B1. Variation With Stress (Age: 50 Years)

Stress (MPa)	Number of Cycles (Millions)	Failure Life (Years)
2.0	554.0	227.0
4.0	45.0	22.5
6.0	3.6	1.8

B2. Variation With Age (Stress: 2 MPa)

Age (Years)	Number of Cycles (Millions)	Failure Life (Years)
30	6,856.0	3,428.0
60	157.0	78.5
90	3.6	1.8



Normal cartilage from a 21-year old adult (x3000)

The surface changes alter the distribution of biomechanical forces further triggering active changes by the tissue.

Biomechanics of Osteoarthritis

- Proteoglycan content and equilibrium stiffness decrease and the rate of deformation and water content increases.

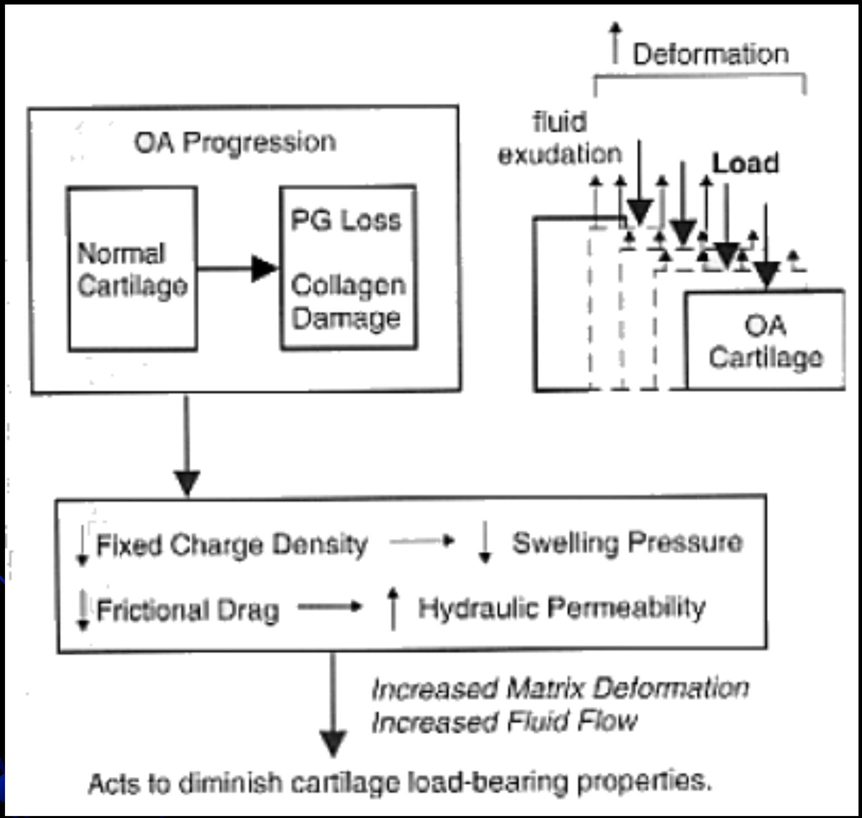
- Decreasing proteoglycan content allows more space in the tissue for fluid.

- An increase in water content correlates with an increase in permeability.

- Increasing permeability allows fluid to flow out of the tissue more easily, resulting in a more rapid rate of deformation.

Proteoglycan loss results in an inability to hold on to water content:

-Decreased resistance to compression –especially with repeated stress



The Future