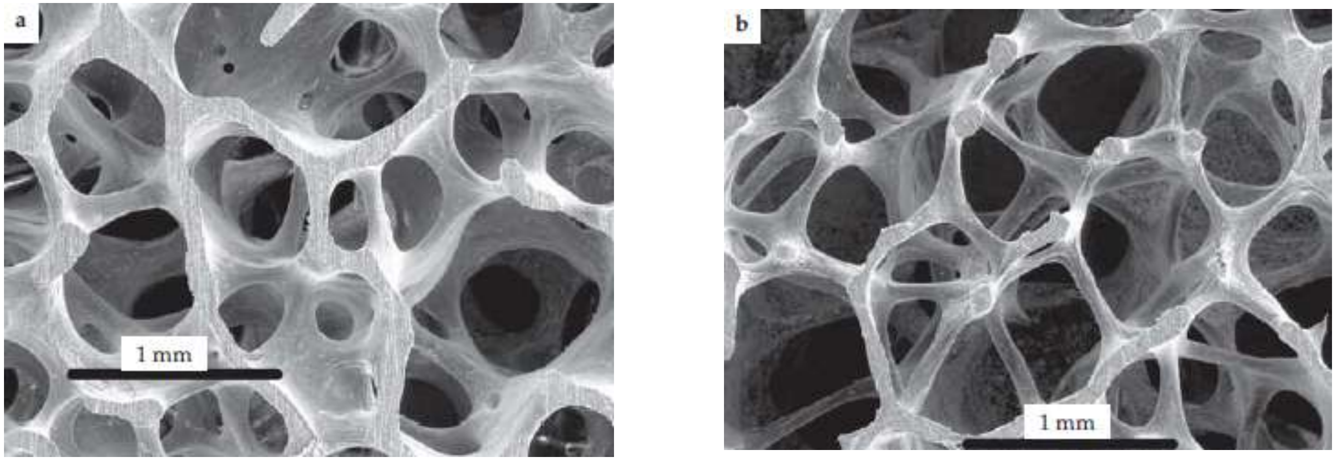
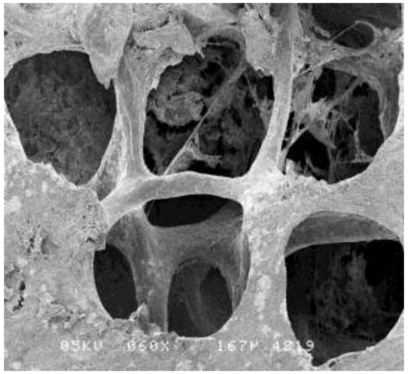
Biomechanics of Fractures & the hierarchical structure of bone tissue



Andreas Panagopoulos
Assistant Professor in Orthopaedics
University Hospital of Patras

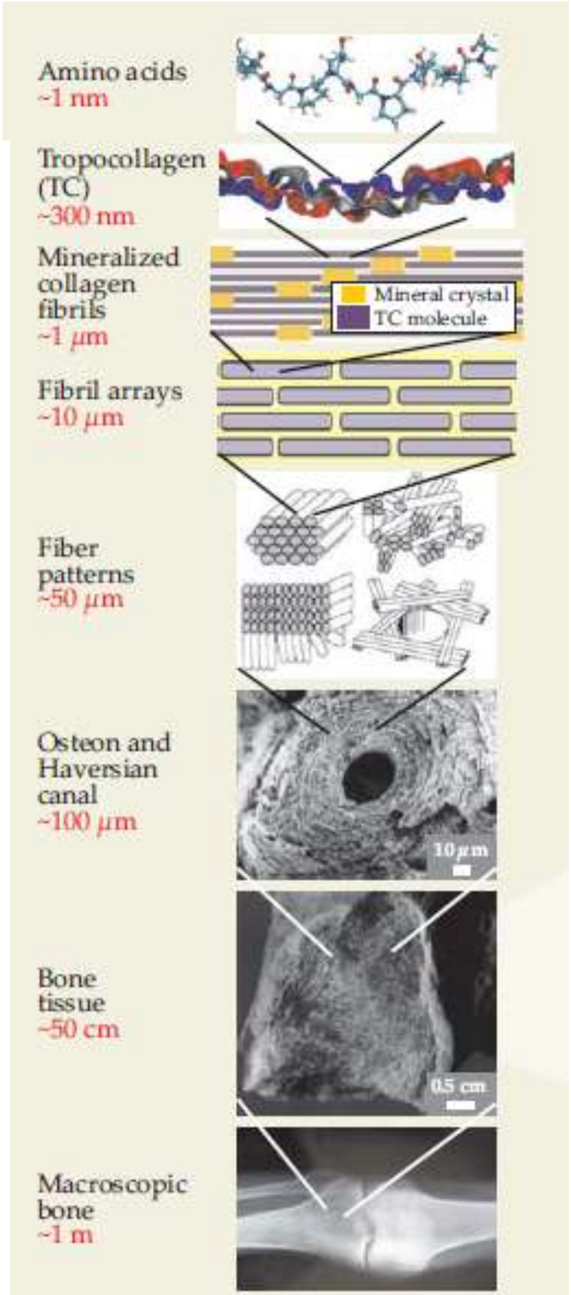
Bone is a Complex System

- Bone is a **hierarchical material** with complex random structures at several different scales
- Bone is a **living tissue** with continuously evolving microstructure (mechanical, biological & chemical factors)
- Bone is a **multi-functional** material:
  - Provides frame
  - Protects organs
  - Manufactures blood cells
  - Stores useful minerals
  - Maintains pH in blood, detoxifies, contributes to movement



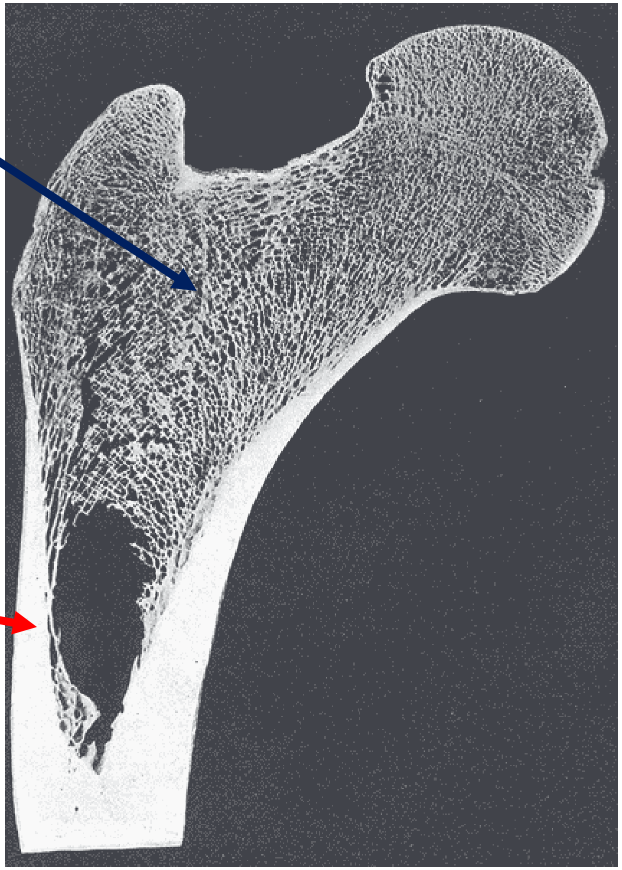
Hierarchical bone structure

- Macrostructure (1-50 cm)
  - Whole bone
- Mesostructure (0.5–10 cm)
  - Trabecular network
- Microstructure (10–500 mm)
  - Single trabecula
- Sub-microstructure (1–10 mm)
  - Single lamella
- Nanostructure (below 1 mm)
  - Collagen fibrils
  - Apatite crystals



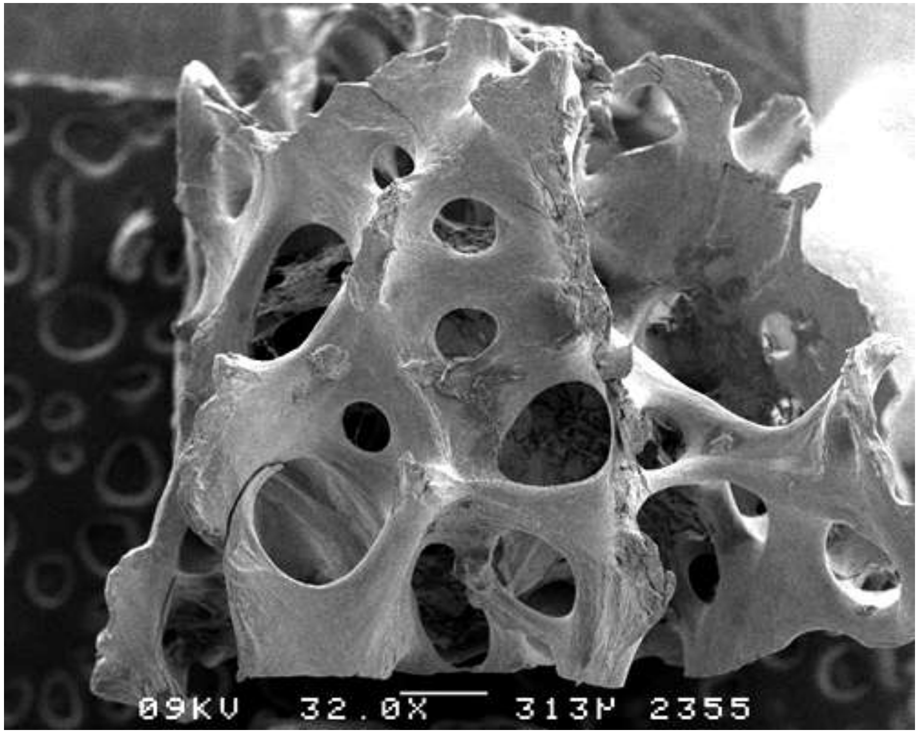
Macrostructure (1 – 50 cm)

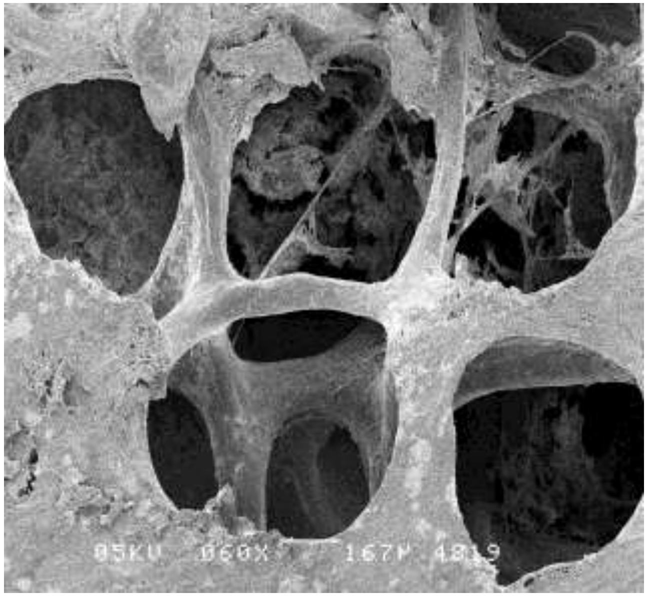
- Trabecular bone
  - High porosity
    - 30 to 90%
  - Skeletal mass
    - 20 to 25%
- Cortical bone
  - Low porosity
    - 5 to 30%
  - Skeletal mass
    - 75 to 80%



Frontal longitudinal midsection of upper femur

Mesostructure (1 – 10 cm)



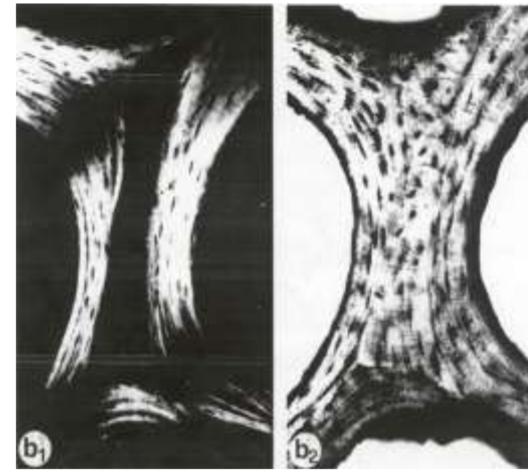


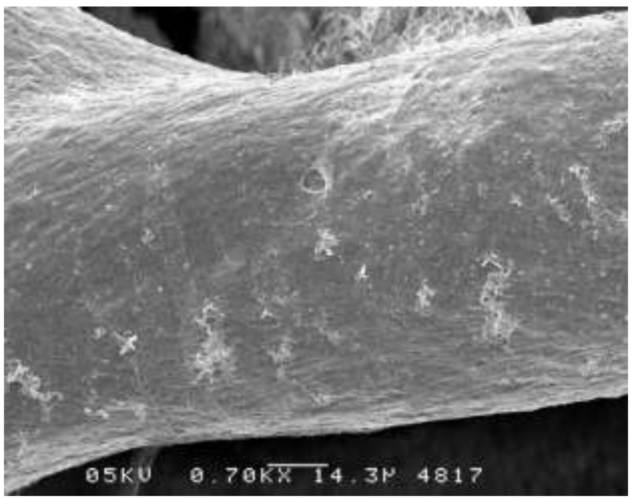
Scanning Electron images

Porous random network of trabeculae

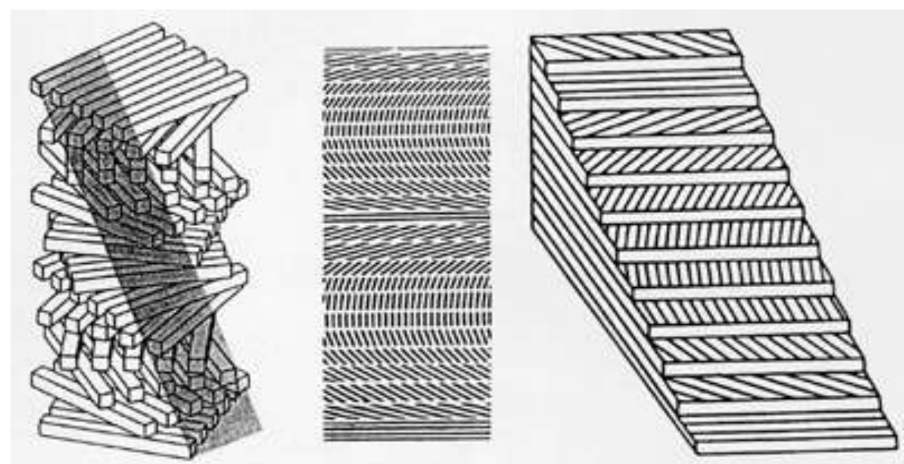
Microstructure (10 - 500 mm)

Trabecular packets
- 50 μm mean wall thickness
- 600 μm radius



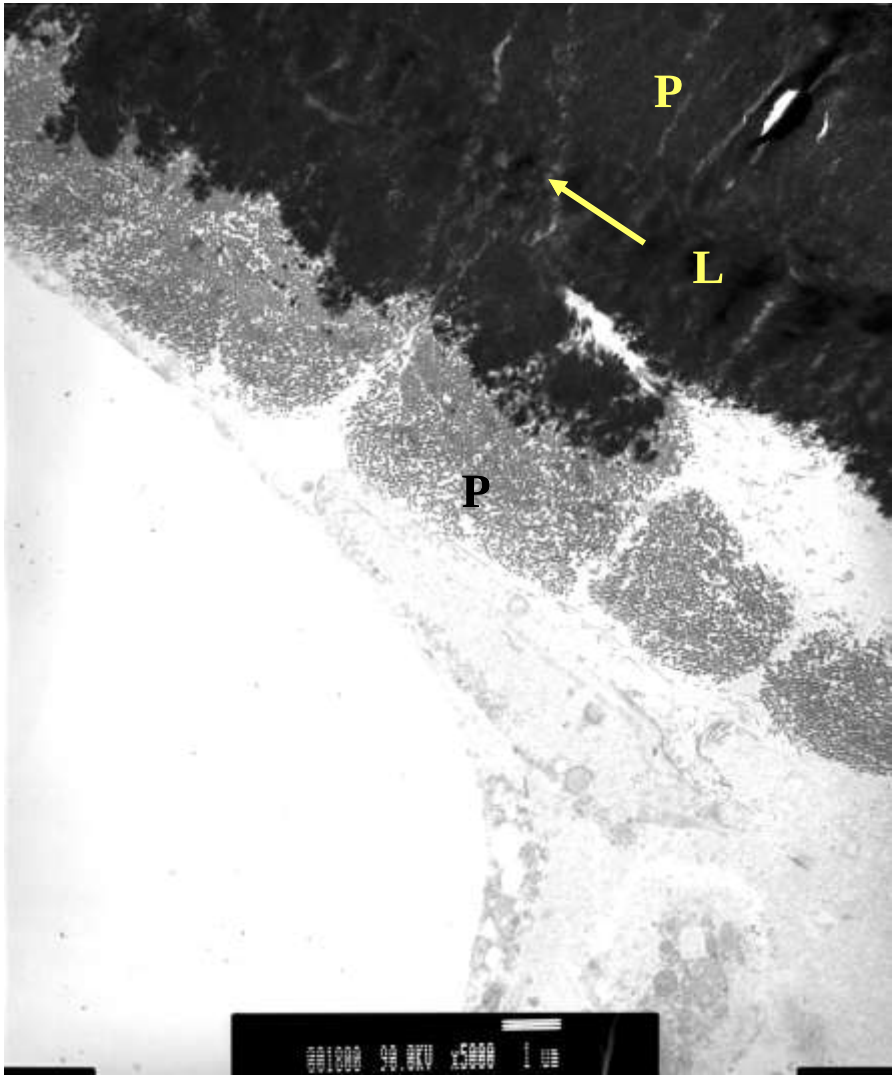


trabecula

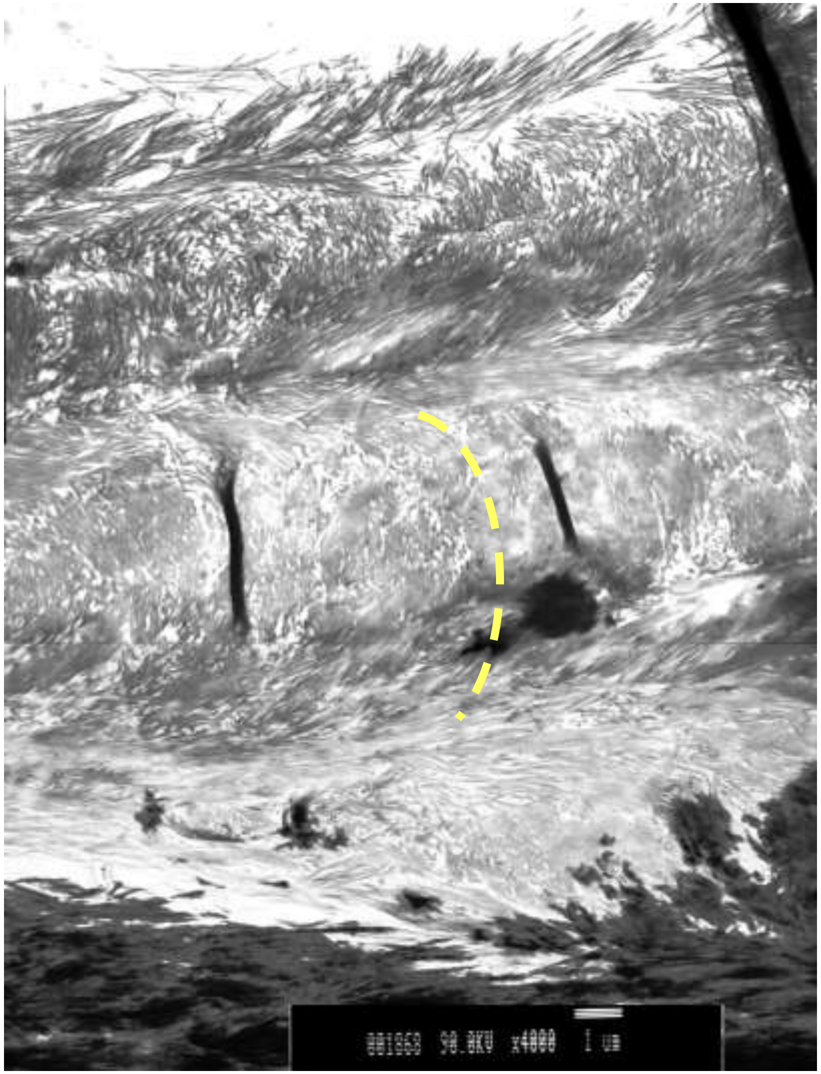


Plywood arrangements

TEM - Plywood Arrangements of lamellae



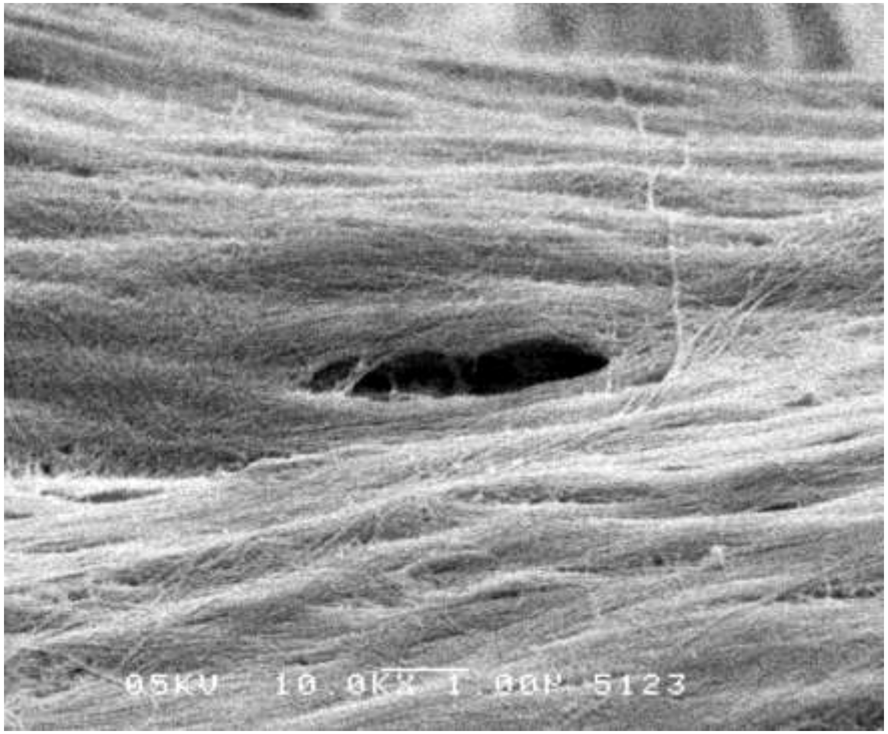
Orthogonal plywood motif (0/90)



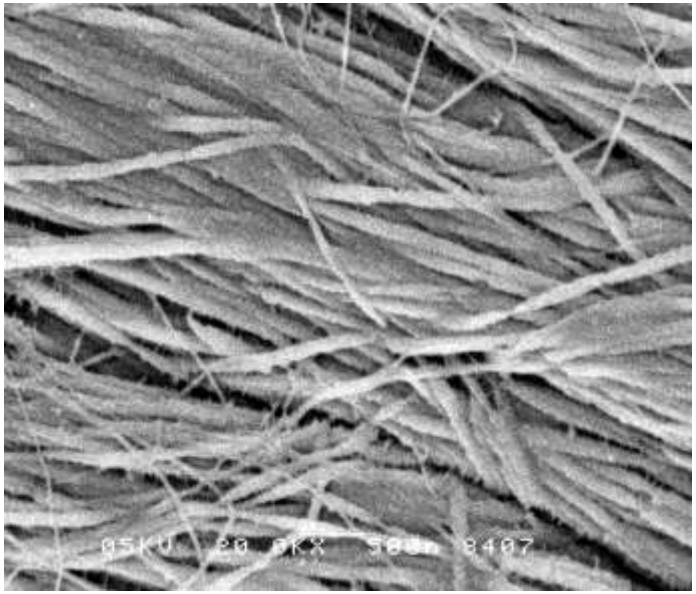
Twisted or rotated plywood motif

Sub-microstructure (1 - 10 mm)

- Single lamella level

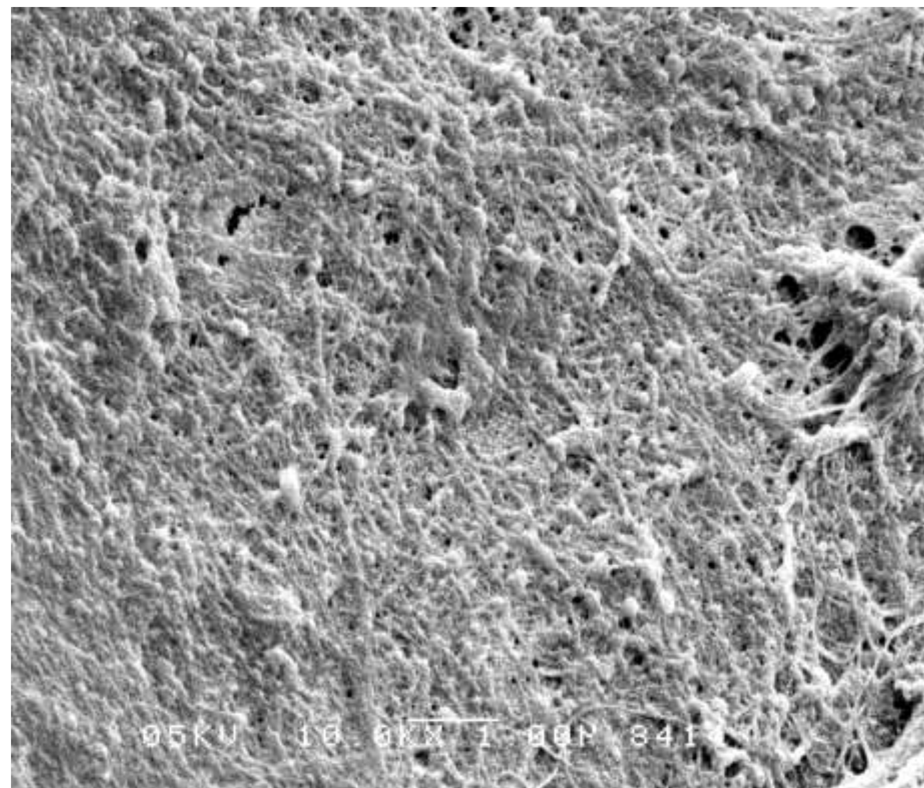


  - single lamella (3 to 7 μm thick)
  - branching bundles (1 - 2 μm diameter)
  - fibrils show splaying, less than 10º
  - ellipsoidal cavities - lacunae (1-2 μm diameter, 20 μm long)



Lamellar structure:
collagen fibrils aligned

Woven bone structure:
no preferential fibril arrangement



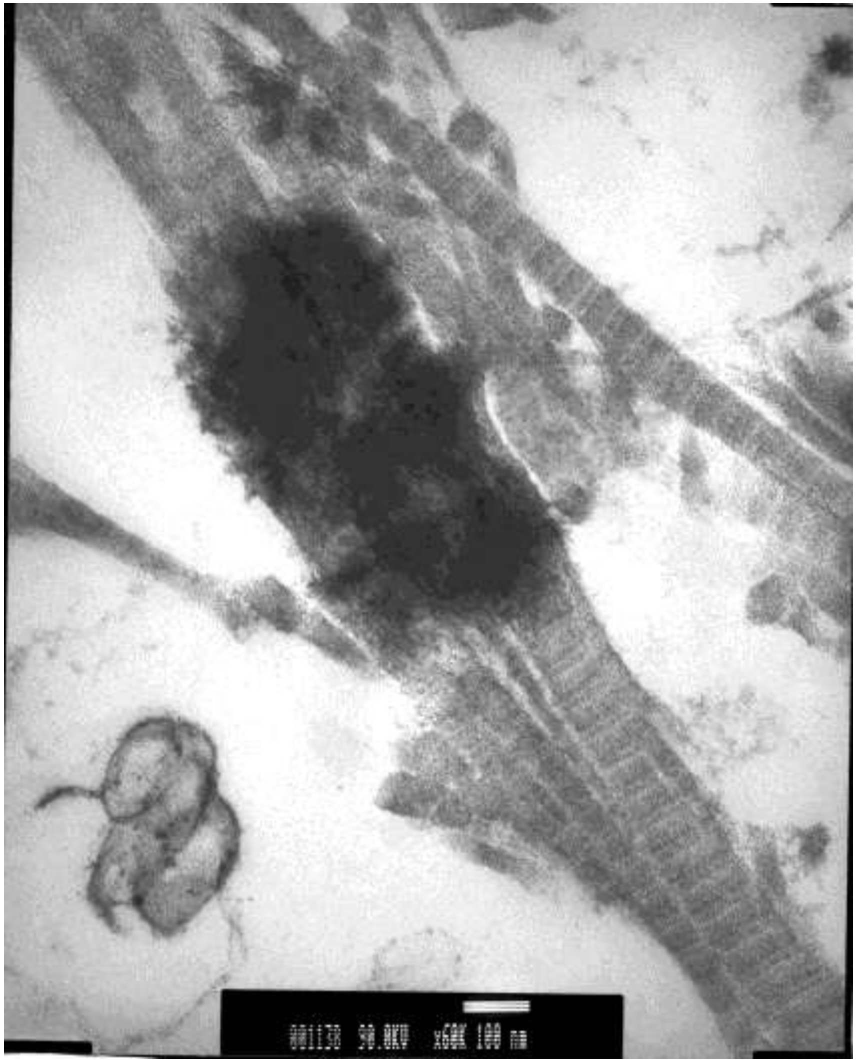
Nanostructure (below 1 mm)

- Collagen (Type I) fibrils
  - 20 – 100 nm diameter
  - 60 – 67 nm periodic pattern

- Apatite crystals (calcium & phosphorus)
  - Shape?
    - Irregular Plates

    [e.g. Robinson, 1952; Weiner *et al*,1986]
      - 50 x 25 x 5 nm

- Other proteins, fluids



TEM – HA crystals in longitudinally-sectioned fibrils

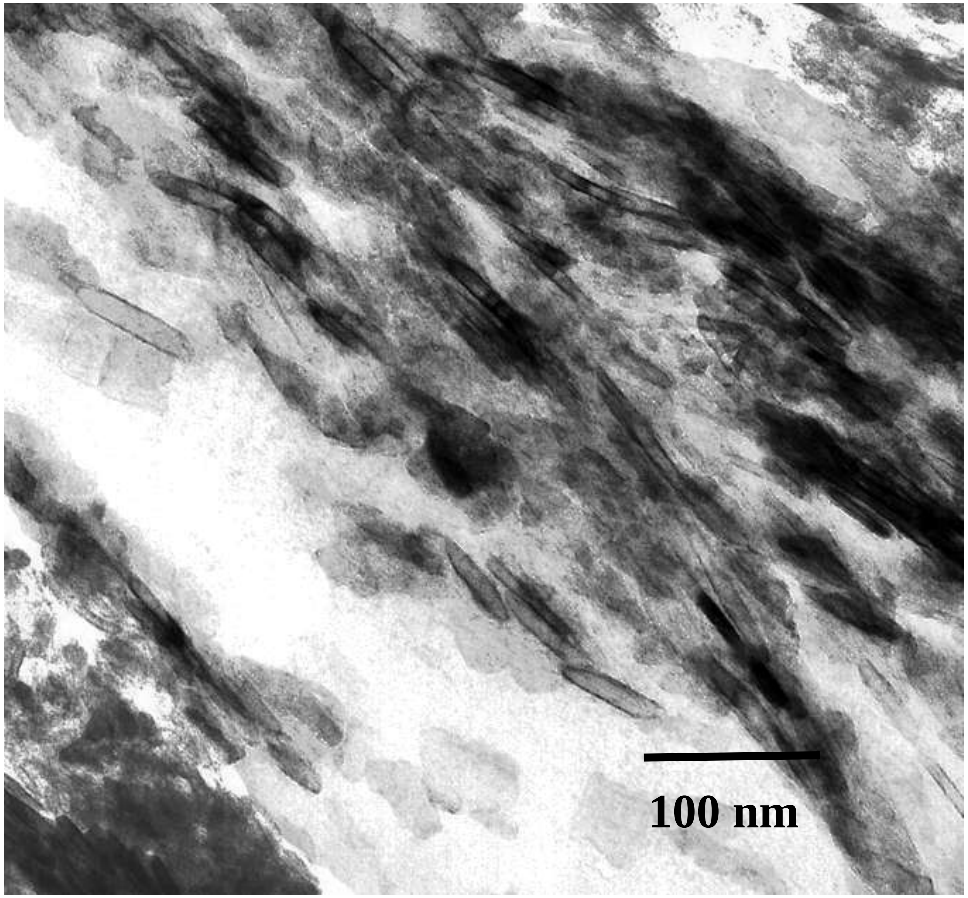
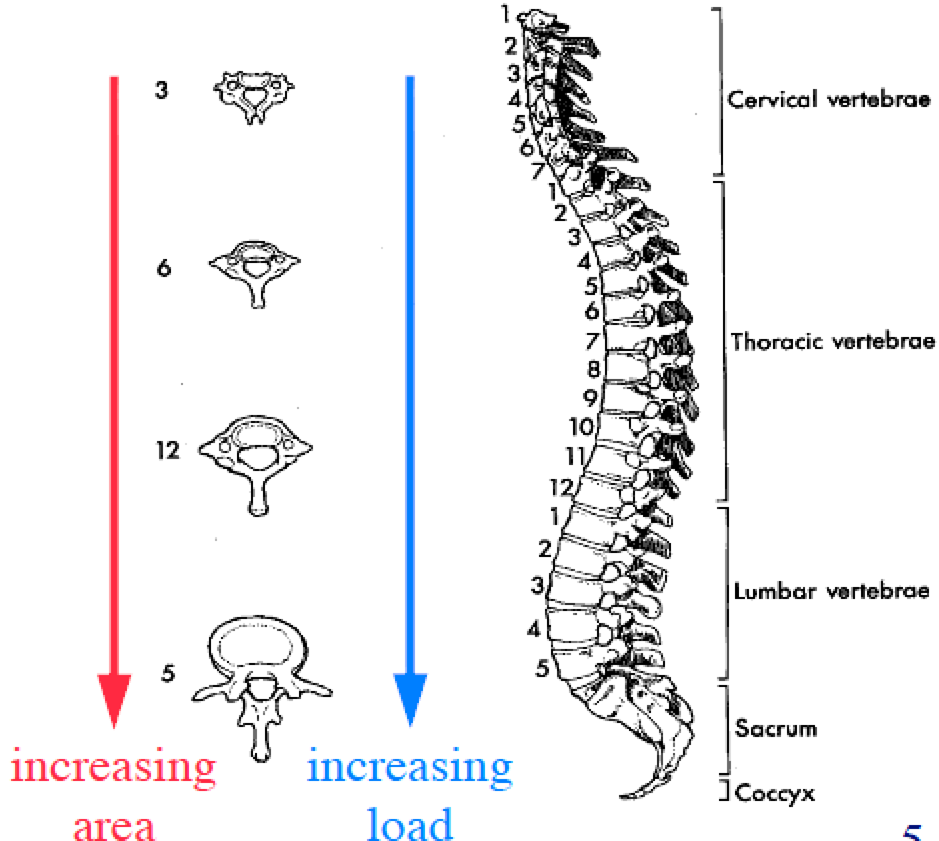


Plate-like shape

Aligned in fibril direction

Skeletal structures are adapted to support musculoskeletal loads

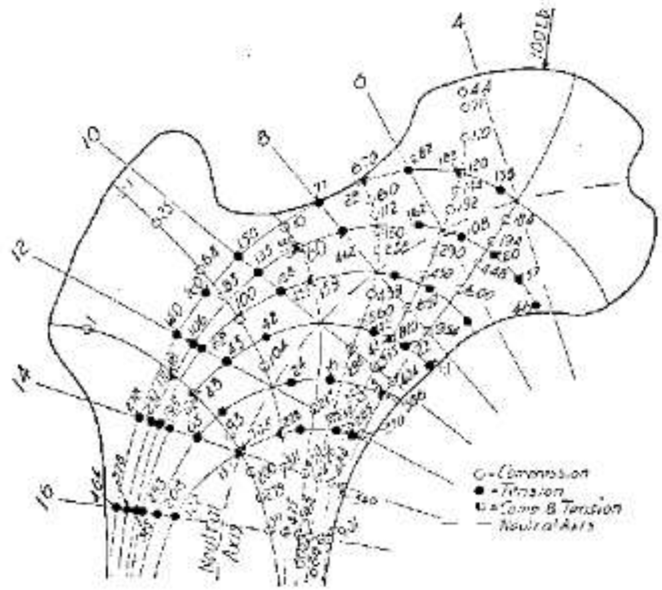
- the structural properties and failure force of skeletal structures is well adapted to functional loads
- does this reflect adaptation of human activity or adaptation of musculoskeletal anatomy?



Optimal design theories: Wolff's Law

Wolff (1869): *Wolff's Law*:
"bone adapts (remodels) in response to the mechanical loads placed on it"

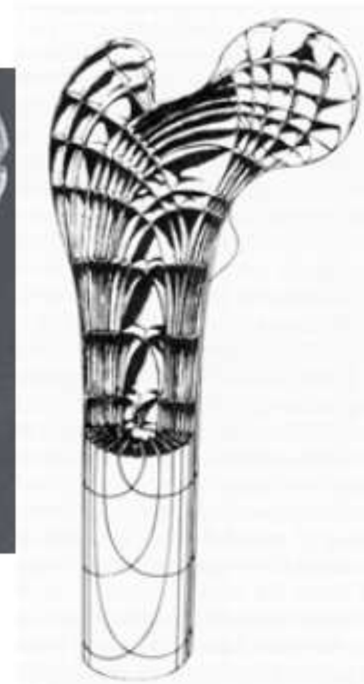
Roux (1895): *Principle of functional adaptation*: bone adapts to its function by "practicing the latter." *Principle of maximum-minimum design*: trajectorial system of trabeculae provides maximum strength with minimum material



Optimal design theories (cont)

(1) *theory of uniform strength*: structure is designed so every bit of material is subjected to the same maximum stress under a specific set of loading conditions

(2) *theory of trajectorial architecture*: structure is designed so material is located only in the paths of transmission of forces, elsewhere there are voids



Basic Biomechanics

Material Properties

- Elastic-Plastic
- Yield point
- Brittle-Ductile
- Toughness

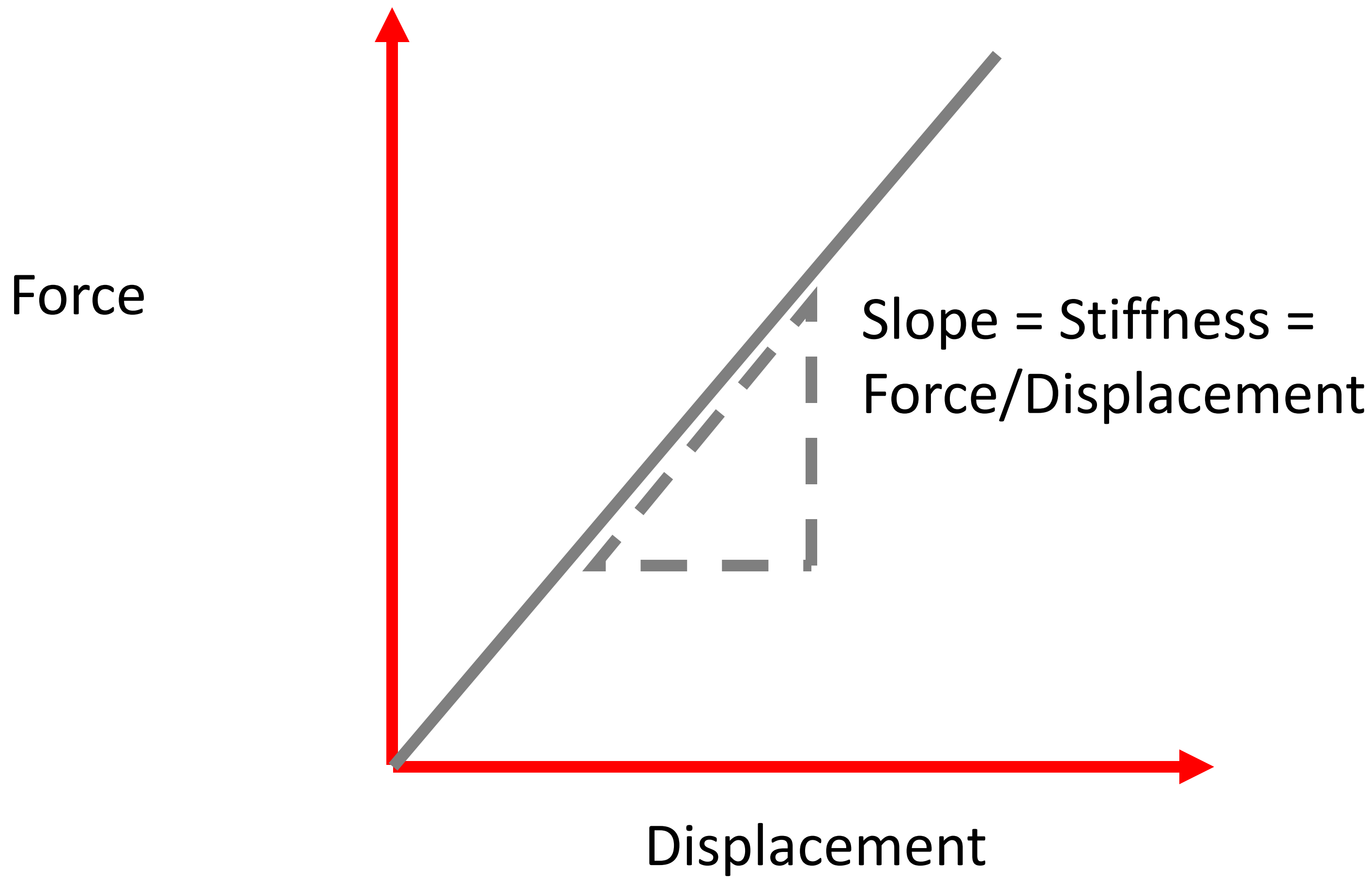
Independent of Shape

Structural Properties

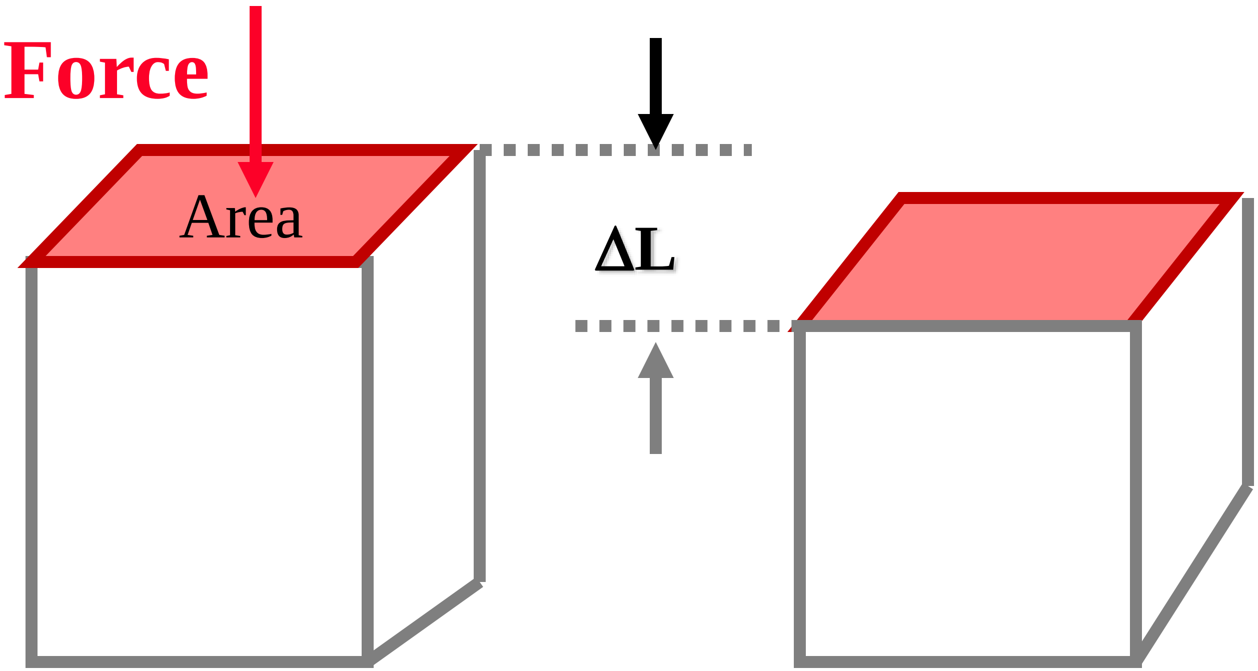
- Bending Stiffness
- Torsional Stiffness
- Axial Stiffness

Depends on Shape and Material

Basic Biomechanics



Basic Biomechanics

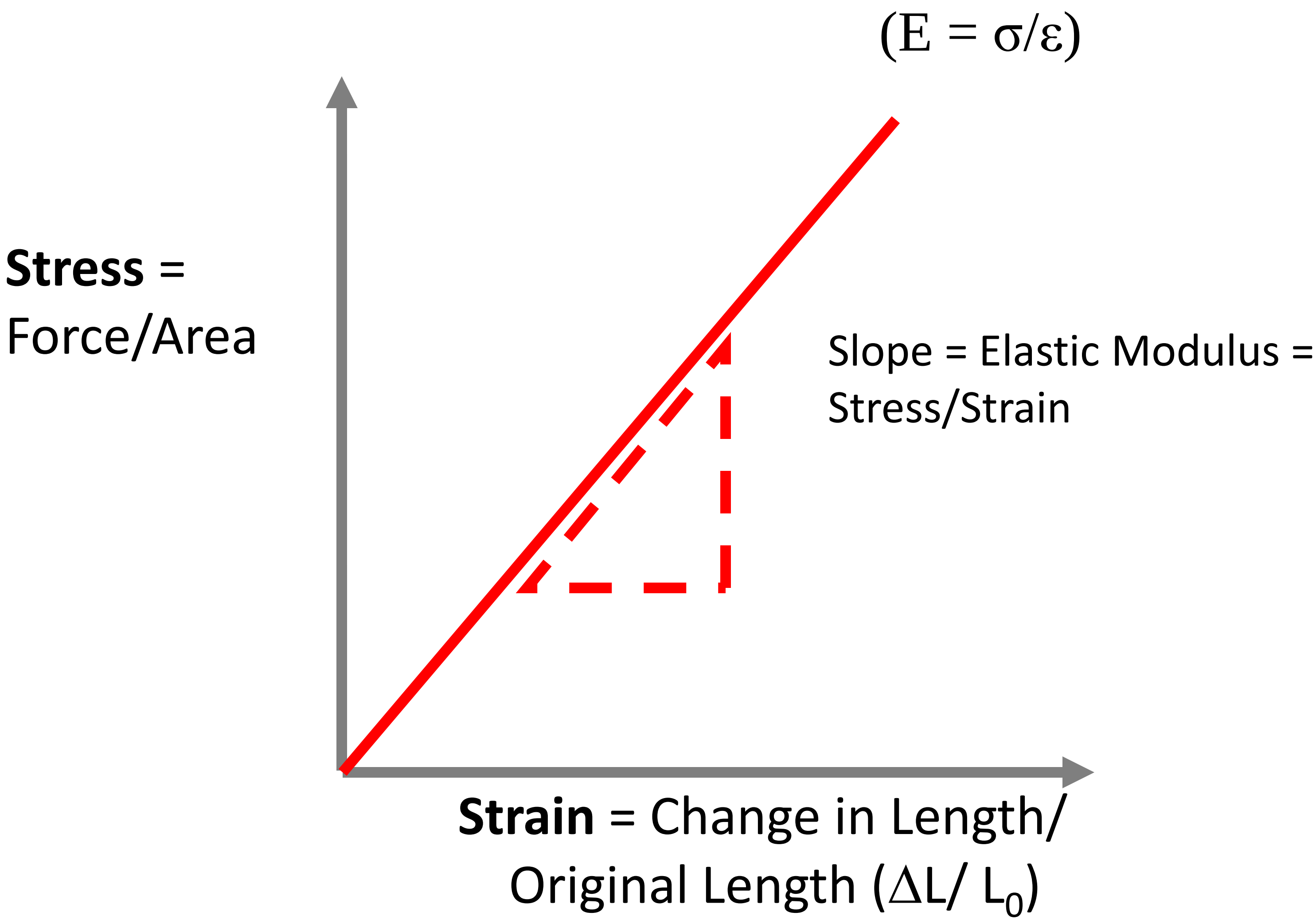


Stress = Force/Area

Strain = $\dfrac{\text{Change Height } (\Delta L)}{\text{Original Height} (L_0)}$

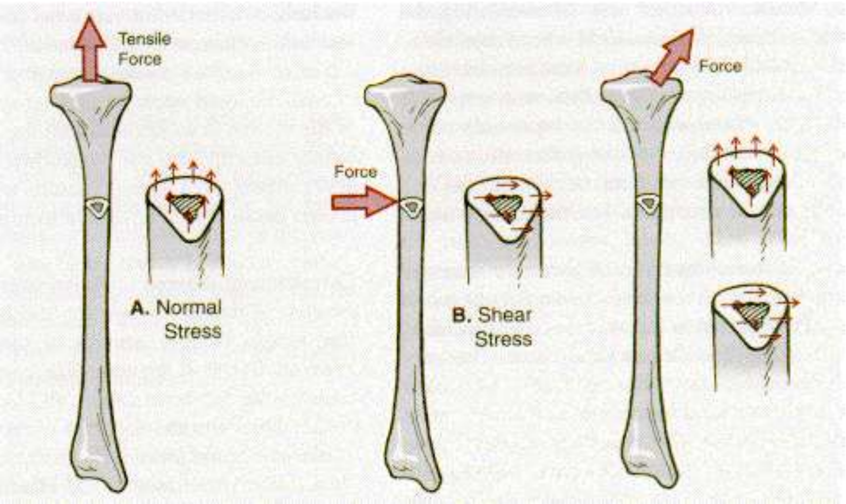
Compressive
Tensile
Sheared

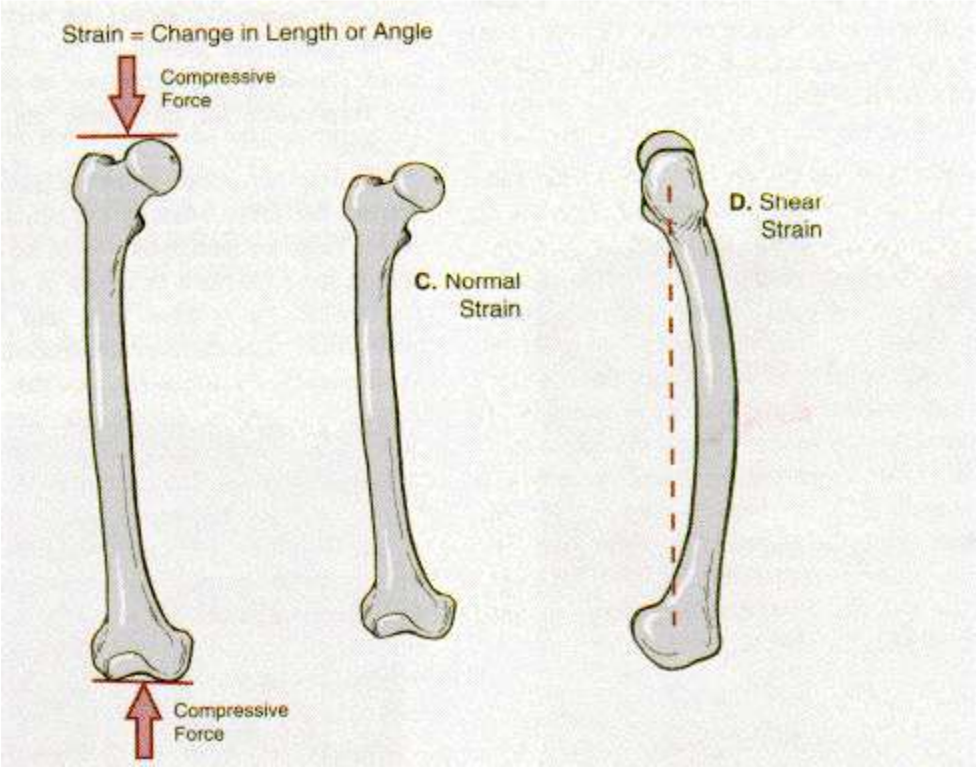
Basic Biomechanics



Stress **= Force/Area**

Strain **= Change in Length/Angle**





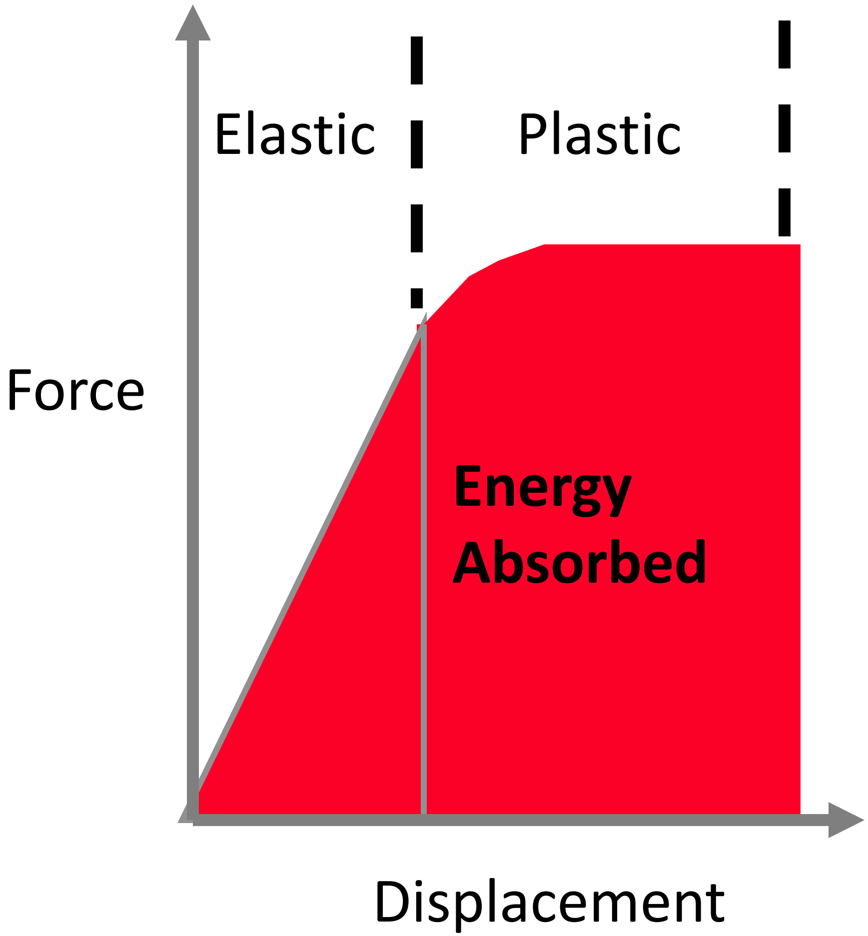
Note: Stress-Strain curve is a normalized Load-Deformation Curve

Basic Biomechanics

Elastic Deformation

Plastic Deformation

Energy



Elastic Biomaterials (Bone)

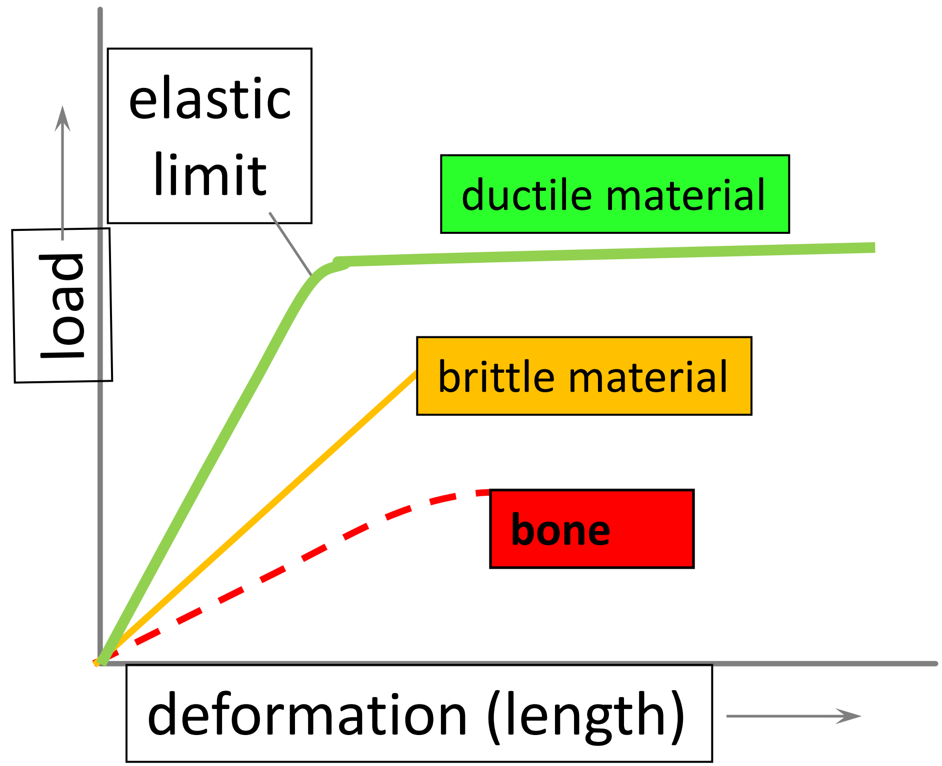
Elastic/Plastic characteristics

Brittle material fails before permanent deformation

Ductile material deforms greatly before failure

Bone exhibits both properties

Load/deformation curves

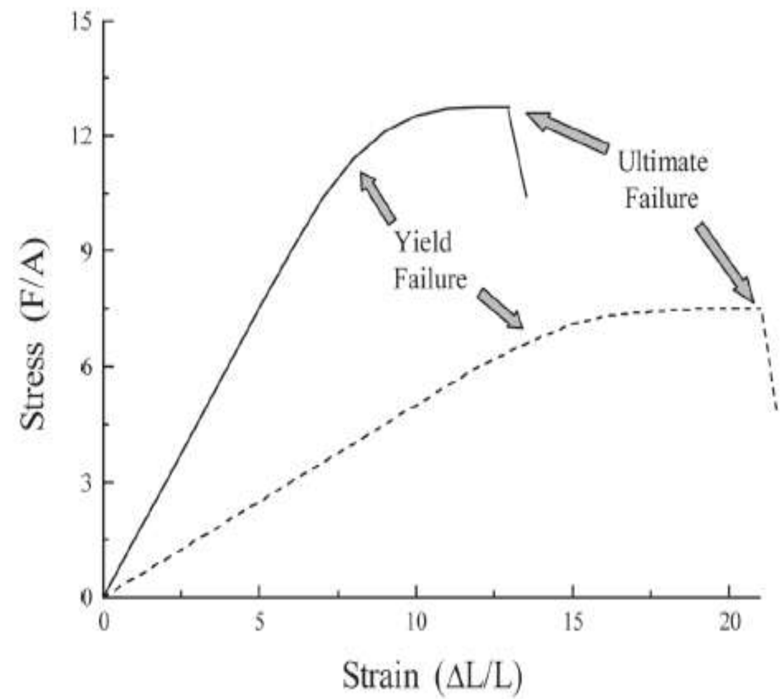


Basic Biomechanics

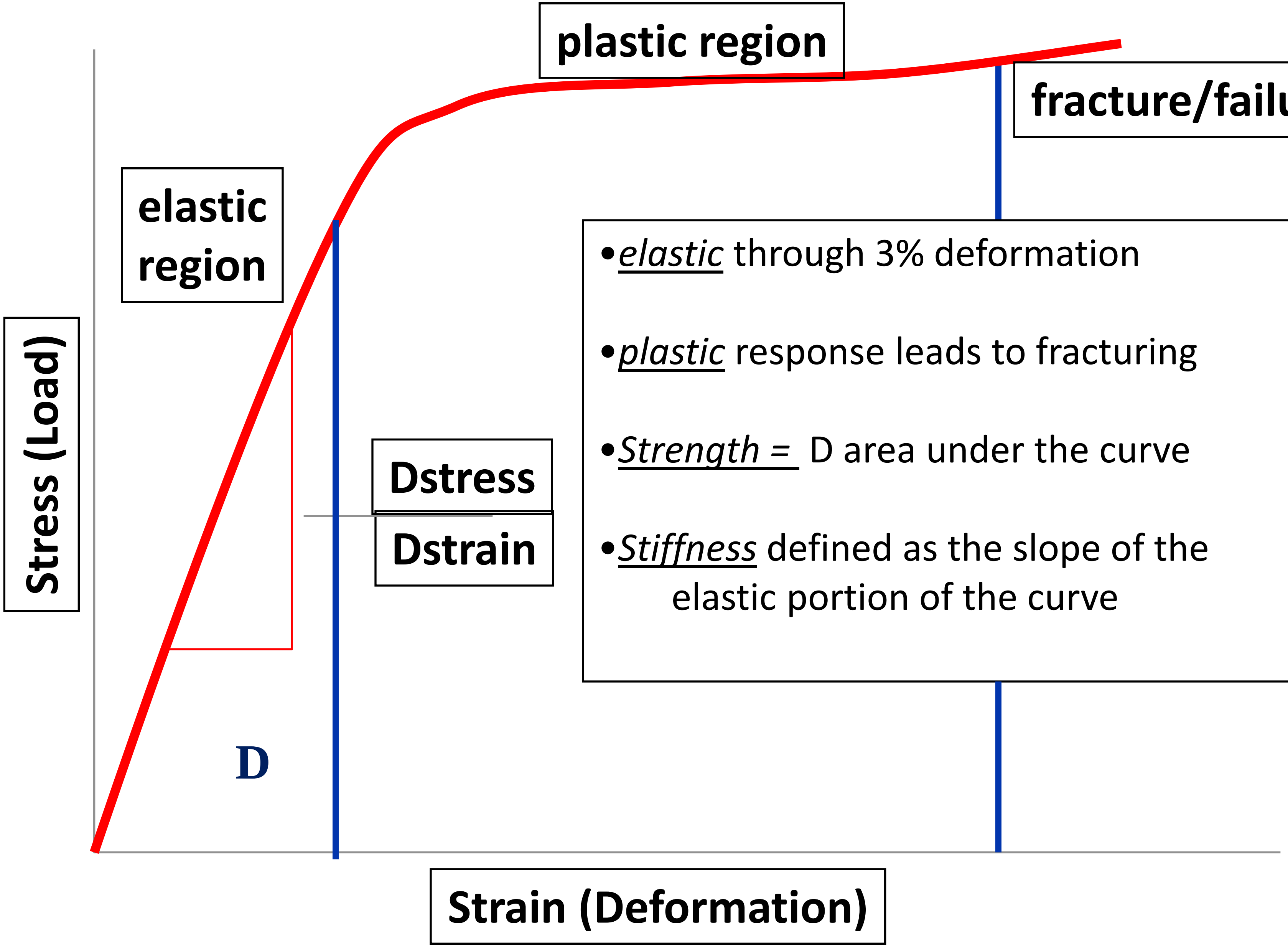
Yield failure first arises through ultrastructural microcracks within the hydroxyapatite and the disruption of the collagen fibrils

As loading continues in the plastic region, the material will eventually experience **ultimate failure** (catastrophic).

The point of breakage is the ultimate stress in:
tension (140 MPa),
shear (65 MPa)
and **compression** (200 MPa)



Elastic & Plastic responses



- *elastic* through 3% deformation
- *plastic* response leads to fracturing
- *Strength =* D area under the curve
- *Stiffness* defined as the slope of the elastic portion of the curve

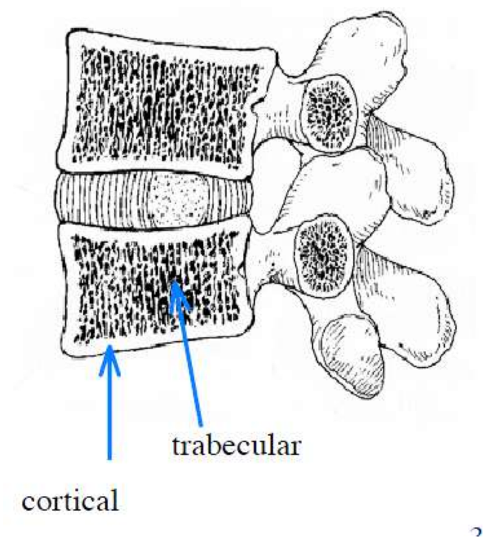
Bone modulus

Cortical bone (5-30% porosity)

18 GPa (longitudinal direction)
12 GPa (transverse direction)
3.3 GPa (shear)

Cancellous bone (30-90% porosity)

(0.1 to 3.5 Gpa) more complex,
- trabecular orientation
- connectivity
- density
- location

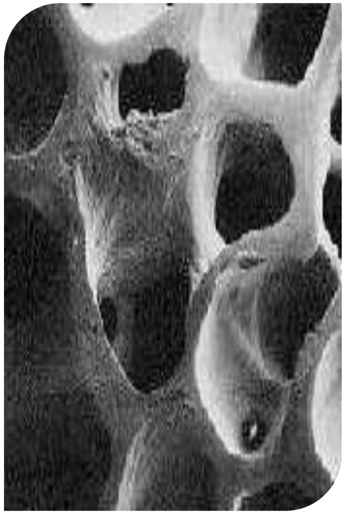


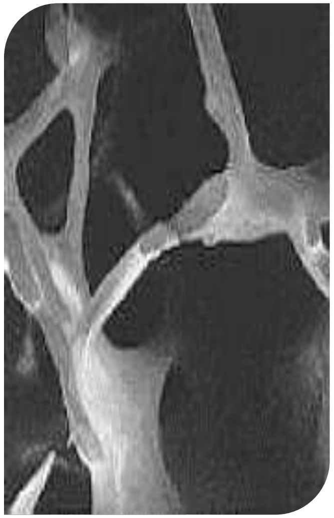
Bone modulus

- Reductions in the degree of mineralization (e.g., immature bone),
- Increases in porosity (e.g., osteoporosis), or
- microstructural damage (accumulation)

will greatly compromise the stiffness of the bone

and thereby lower the elastic modulus.





Basic Biomechanics

Load to Failure

Continuous application of force until the material breaks (failure point at the ultimate load).

Common mode of failure of bone and reported in the implant literature.

Fatigue Failure

Cyclical sub-threshold loading may result in failure due to fatigue.

Common mode of failure of orthopaedic implants and fracture fixation constructs.

Basic Biomechanics

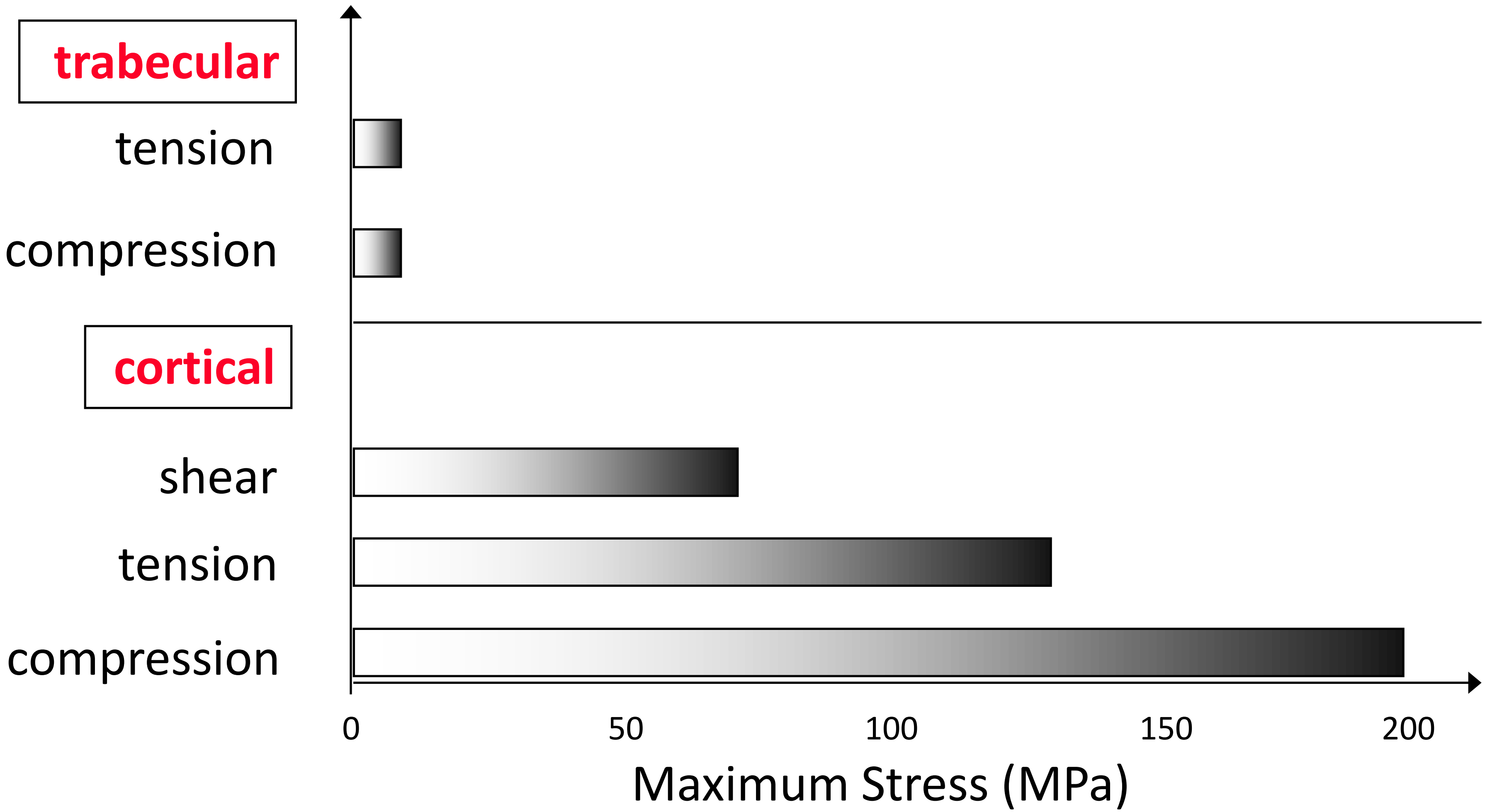
Anisotropic Mechanical properties dependent upon direction of loading

(weakest in shear, then tension, then compression)

Viscoelastic Stress-Strain character dependent upon rate of applied strain (time dependent)

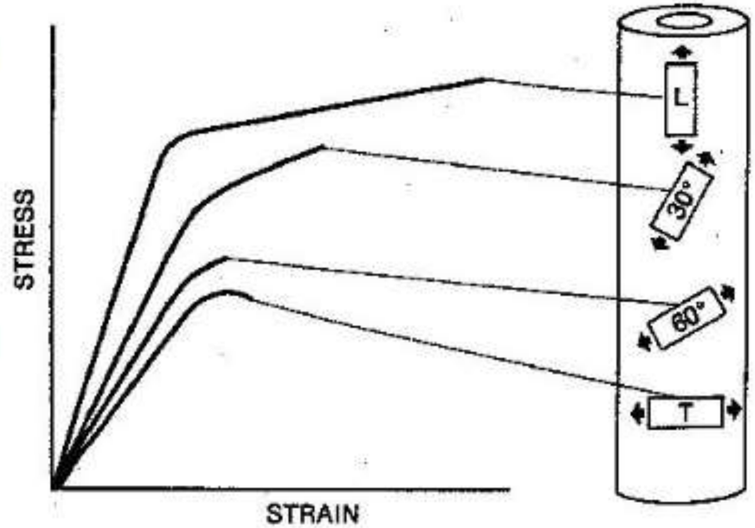
Trabecular bone becomes stiffer in compression the faster it is loaded

Bone Anisotropy



The strength and modulus of bone vary with the direction of loading

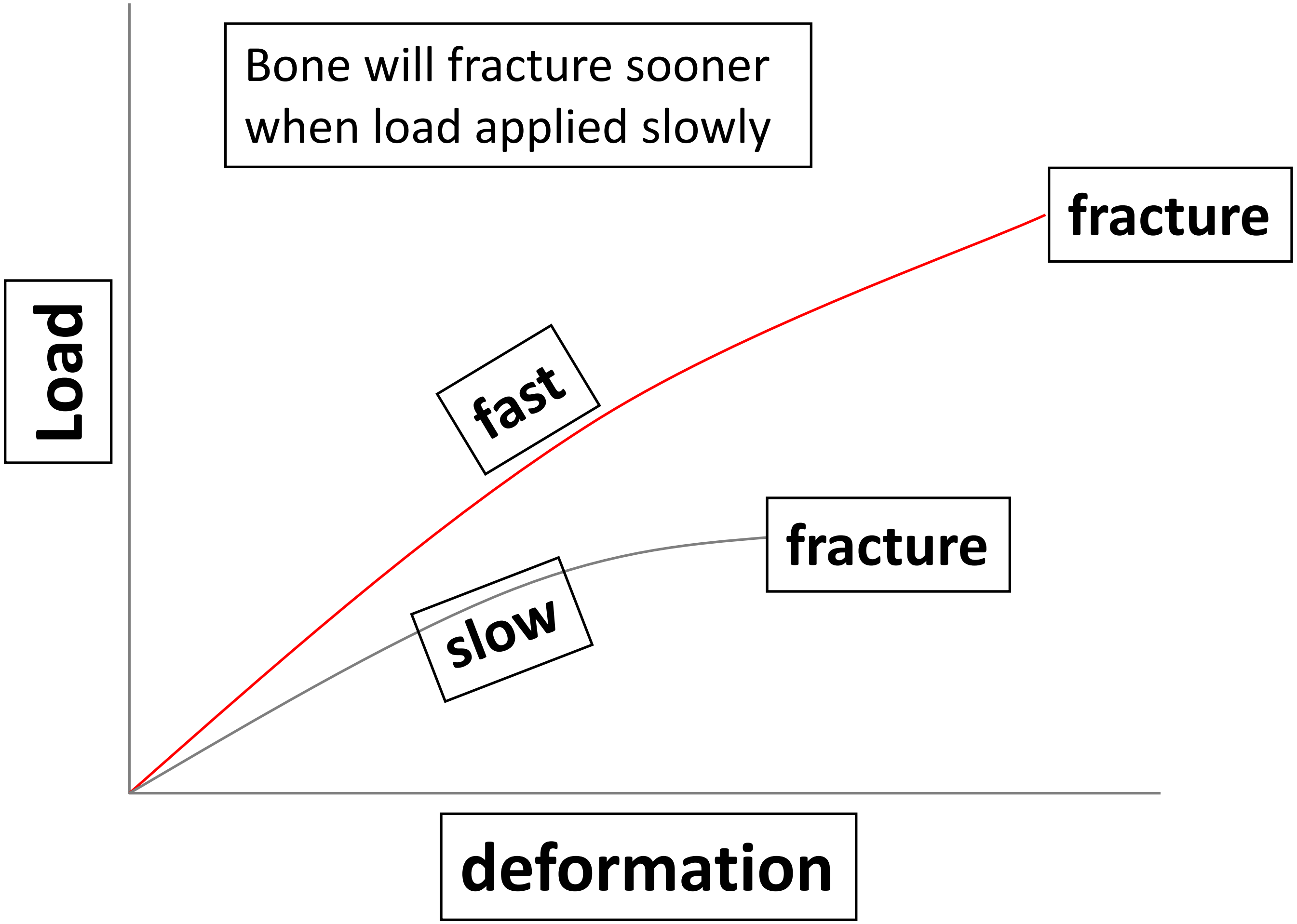
- bone is stronger in compression than tension
- bone is *anisotropic*: its modulus and strength (in tension or compression) depend on the orientation of the tissue with respect to the load



- for cortical bone:

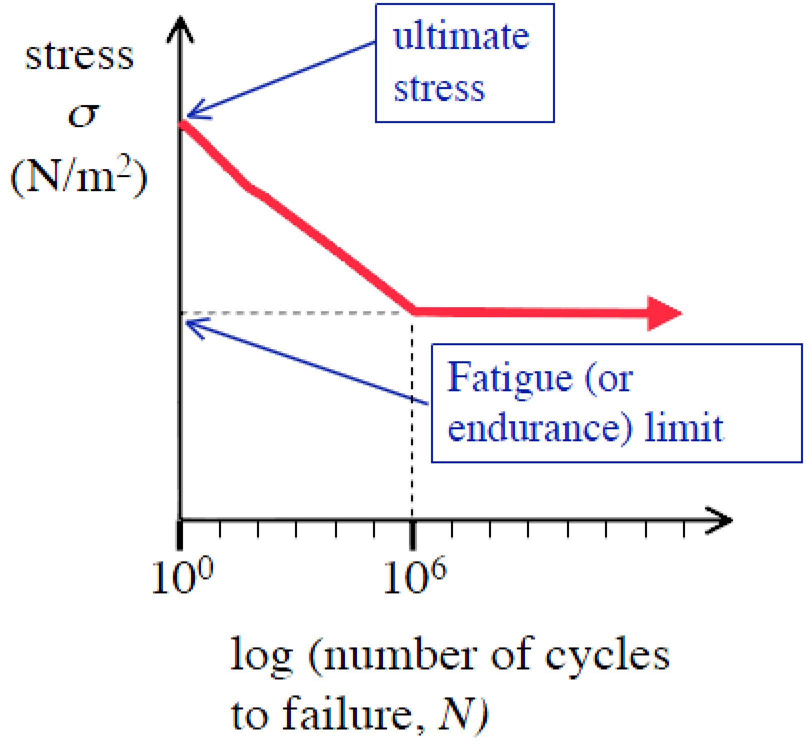
$E_{long} \approx 17$ GPa	$(\sigma_{ult})_{long}^{tension} \approx 135$ MPa	$(\sigma_{ult})_{long}^{comp} \approx -190$ MPa
$E_{trans} \approx 11$ GPa	$(\sigma_{ult})_{trans}^{tension} \approx 50$ MPa	$(\sigma_{ult})_{trans}^{comp} \approx -130$ MPa

Viscoelastic properties



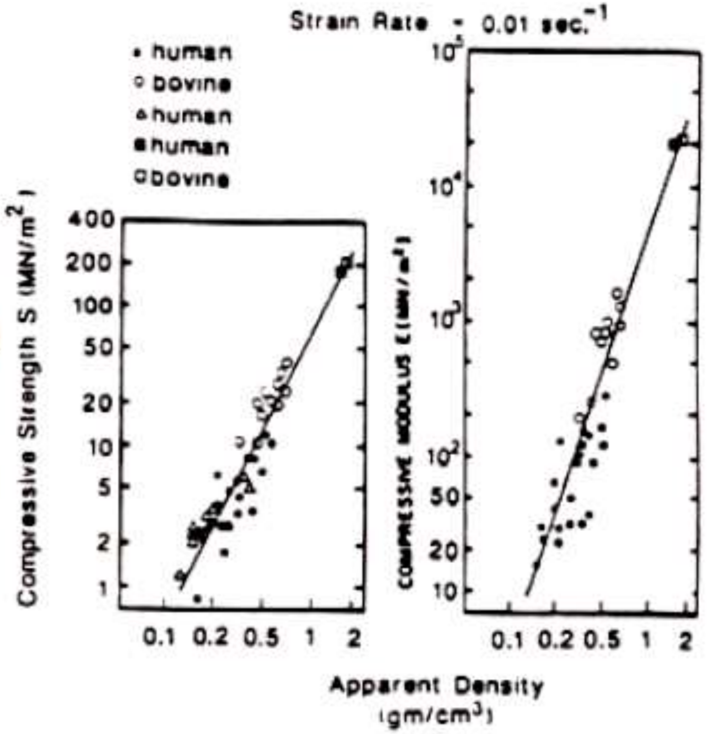
The strength of bone declines with number of cycles during cyclic loading

Like all materials, bone is susceptible to *fatigue* - when subject to repetitive or fluctuating stress, it will fail at a stress level much lower than that required to cause fracture on a single application of load



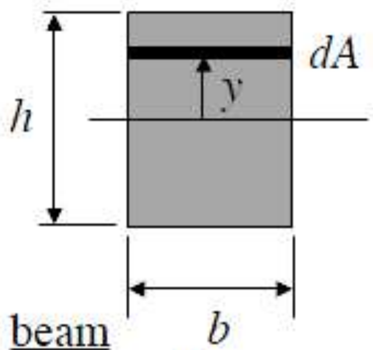
The strength and modulus of bone vary with density (and microstructure)

- the strength and modulus of bone associate with density:
  - strength $\propto$ (density)^2
  - modulus $\propto$ (density)^3
- density however can explain only a small percent of the variability between samples in bone strength (R~0.4); therefore microstructure must also be important



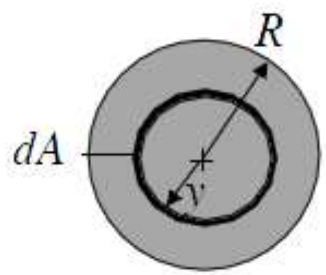
Area moment of inertia affects stresses due to bending moments

Recall that the stress due to an internal bending moment is $\sigma_b = \dfrac{M_b \cdot y}{I_a}$, where M_b is the bending moment, y is the distance from the neutral axis, and I_a is the area moment of inertia of the structure $I_a = \int y^2 dA$.



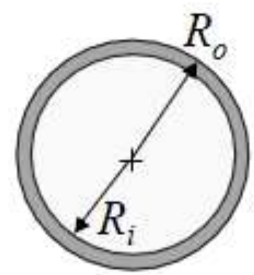
beam

$$I = \frac{1}{12} b h^3$$



solid rod

$$I = \frac{\pi}{4} R^4$$



hollow rod

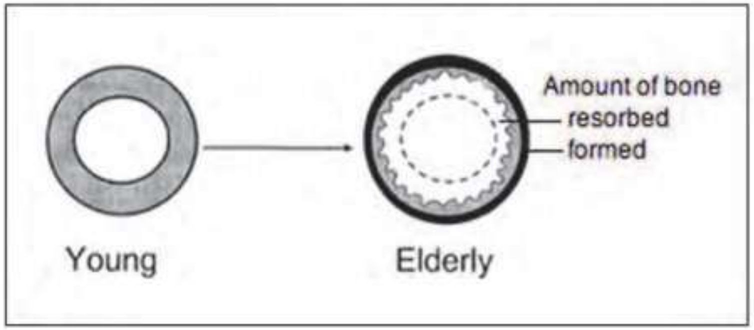
$$I = \frac{\pi}{4}\left(R_o^{\,4} - R_i^{\,4}\right)$$

Influence of bone geometry on biomechanical behavior

Moment of inertia = P (R^4 – r^4) / 4
R: cortical outer diameter; r: cortical inner diameter

In **tension** and **compression**, the load to failure and stiffness are proportional to the cross sectional area of the bone

In **bending** are proportional to the "area moment of inertia" and inversely proportional to the length
(The longer the bone, the bigger the bending moment.)



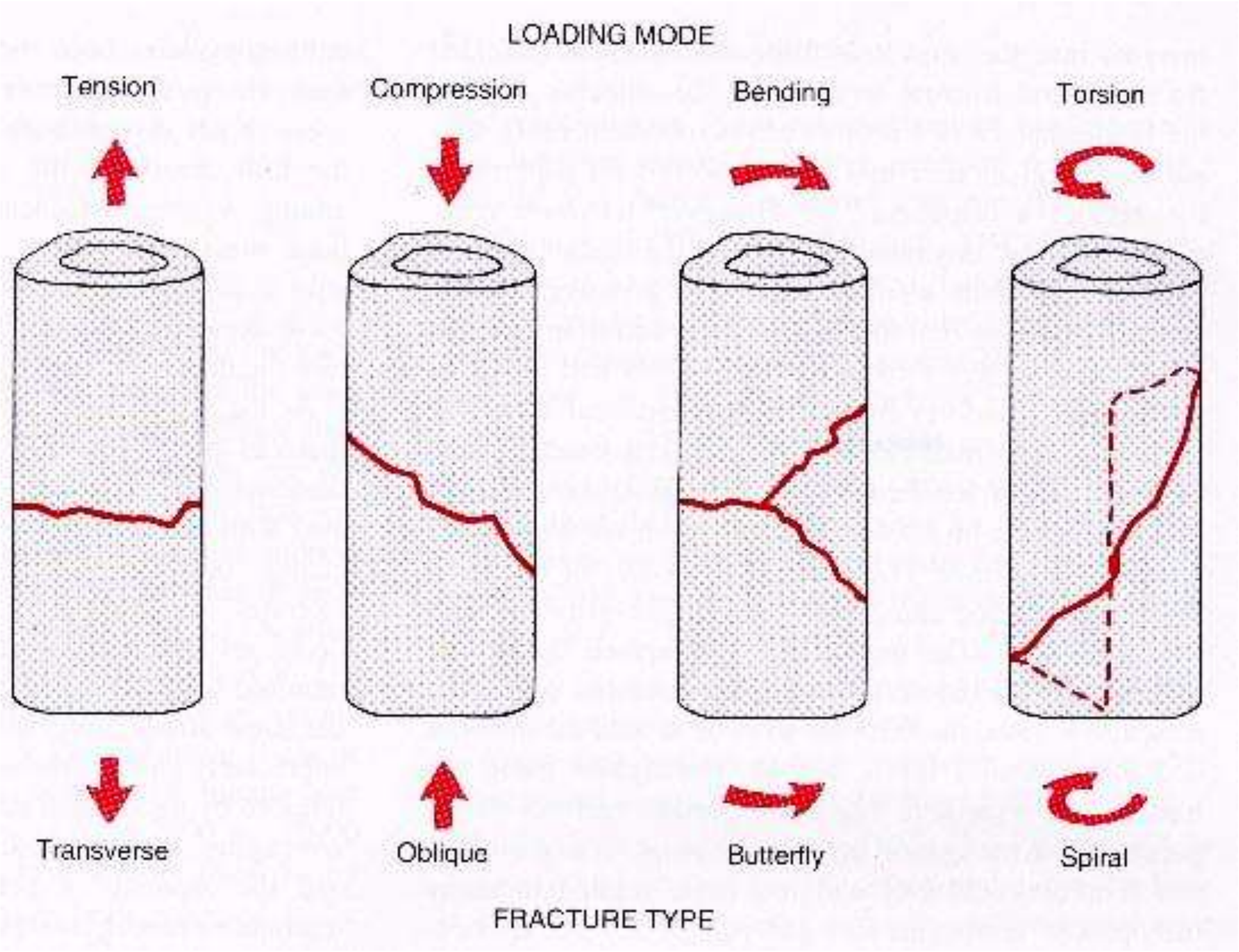
Influence of bone geometry on biomechanical behavior

In Torsion are proportional to the Polar Moment of Inertia (J)

$J = [pi/2]x[Ro^4\text{-}Ri^4] = 2.I$; $T/ø = JG/L$ (T/ø= torsional stiffness, T= torque, ø= angle of twist, G= shear modulus, L= length of shaft)

This takes into account the cross sectional area and the distribution of bone tissue around the neutral axis

Mechanical Loading of Bone

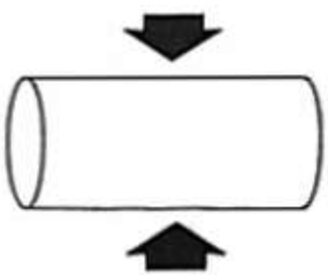


Compressive Loading

Vertebral fractures

Buckle fractures

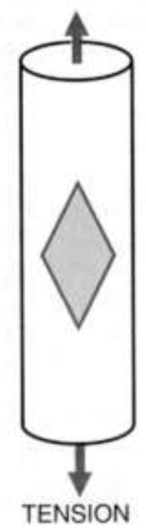
Crush fractures

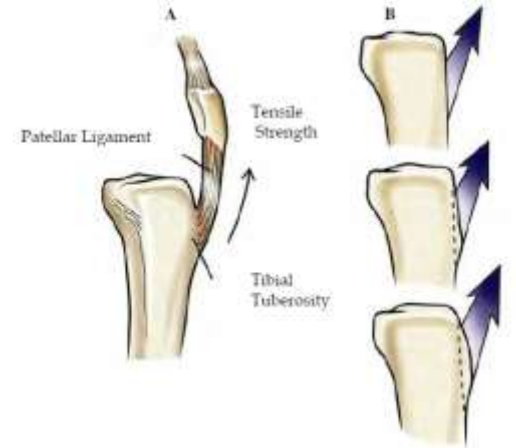


Compression

Tensile Loading

Main source of tensile load is muscle
- tension can stimulate tissue growth
- usually an avulsion
- sprains, strains,
- inflammation, bony deposits



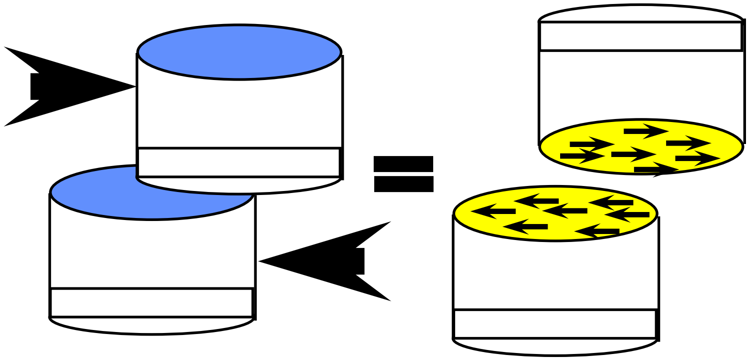


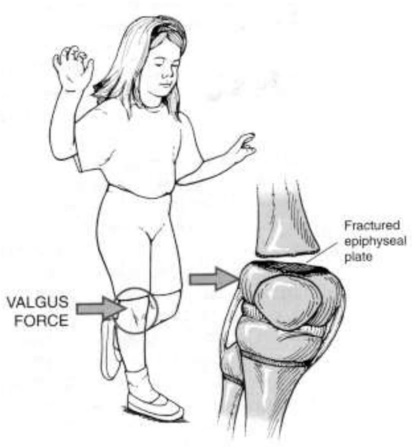
Shear Forces

The bone fails more quickly when exposed to a shear strength rather than a compressive or tensive strength.

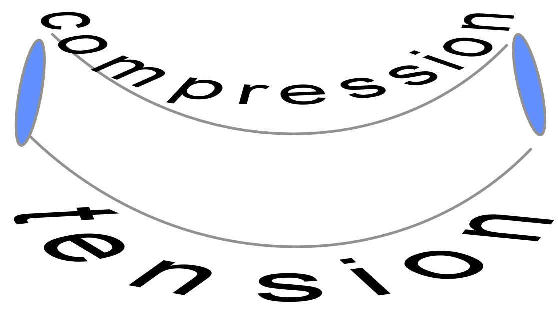
This is because the bone is anisotropic and responds differently when it receives loads of different directions

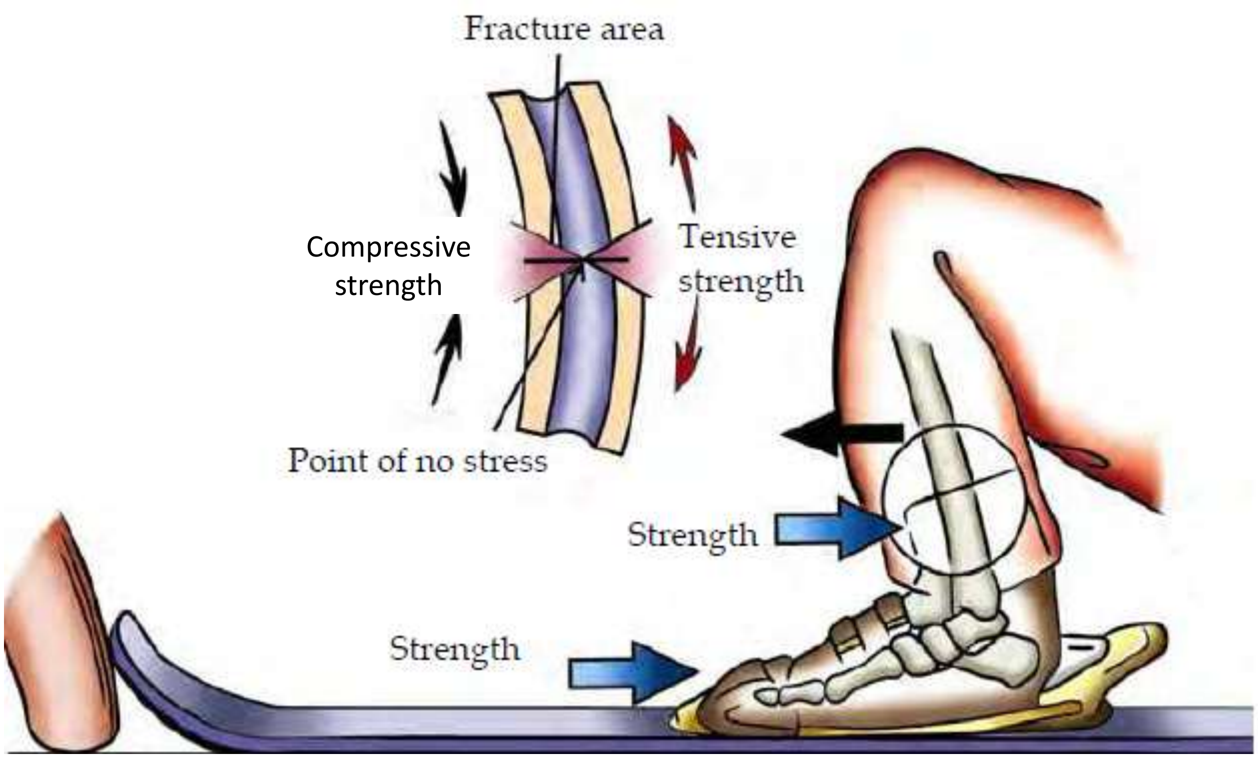
Epiphyseal fractures
Tibial plateau fractures
Femoral condyles fracture





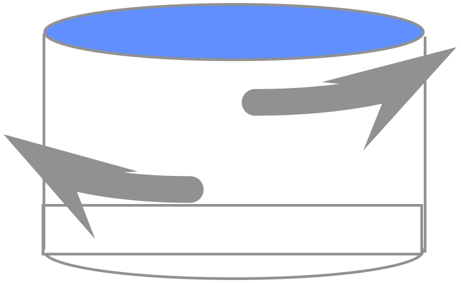
Bending Forces





Usually a 3- or 4-point force application

Torsional Forces

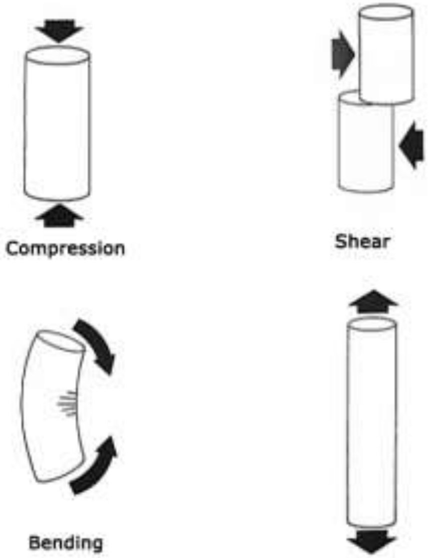


Caused by a *twisting* force produces shear, tensile, and compressive loads

tensile and compressive loads are at an angle

often see a *spiral fracture* develop from this load

Combined Forces



Available online at www.sciencedirect.com
ScienceDirect
Joint Bone Spine 74 (2007) 233–239
JOINT BONE SPINE
http://france.elsevier.com/direct/BONSOI/
ELSEVIER MASSON

Review

Evaluation of macrostructural bone biomechanics

Hélène Beaupied*, Eric Lespessailles, Claude-Laurent Benhamou

IPROS, Inserm U658, CHR Orléans, Hôpital Porte Madeleine, 1 rue Porte Madeleine – BP 2439, 45032 Orléans Cedex 1, France

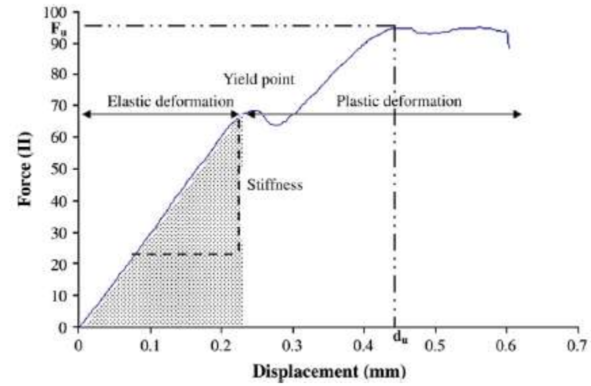
Received 19 June 2006; accepted 27 November 2006
Available online 23 February 2007

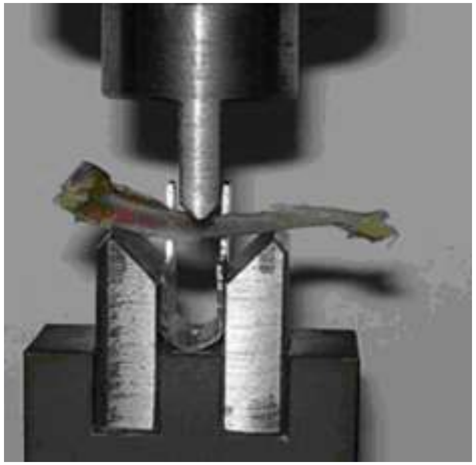
Comprehensive review of available biomechanical tests

- Usually under cycling loading

- torsion, bending, tensile & compressive forces

- extrinsic (force/displacement curve)
- intrinsic (load to failure, deformation at failure, modulus of Young etc)



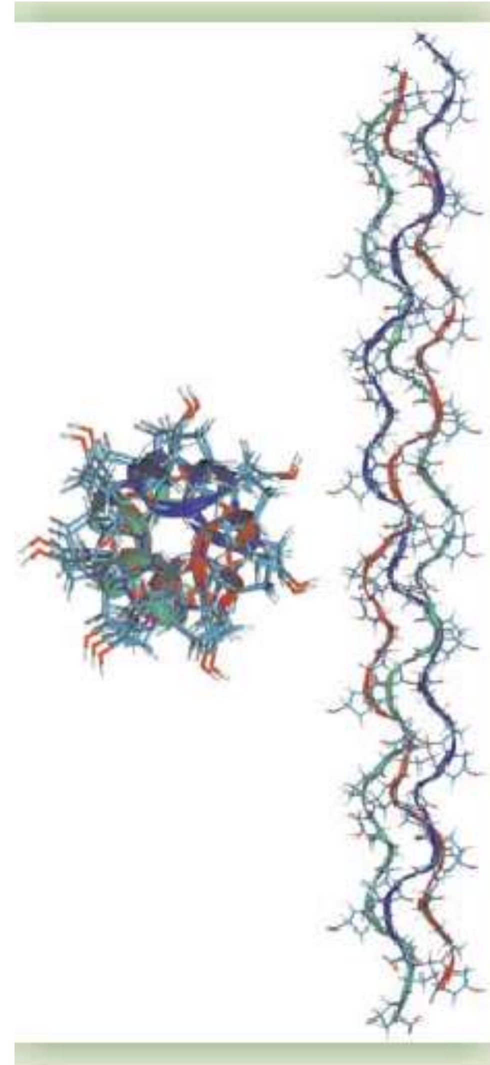


Bone as a Composite Material

New techniques in microscale **bio-mechanology**

- Transimmision electron microscopy
- X-ray microscopy
- Force microscopy
- Raman spectroscopy
- Fourier transform infrared microspectroscopy
- nanoindentation

Results suggest that permanent deformation (plasticity) in bone occurs from multiple, concurrent mechanisms that are active at **all hierarchical** levels



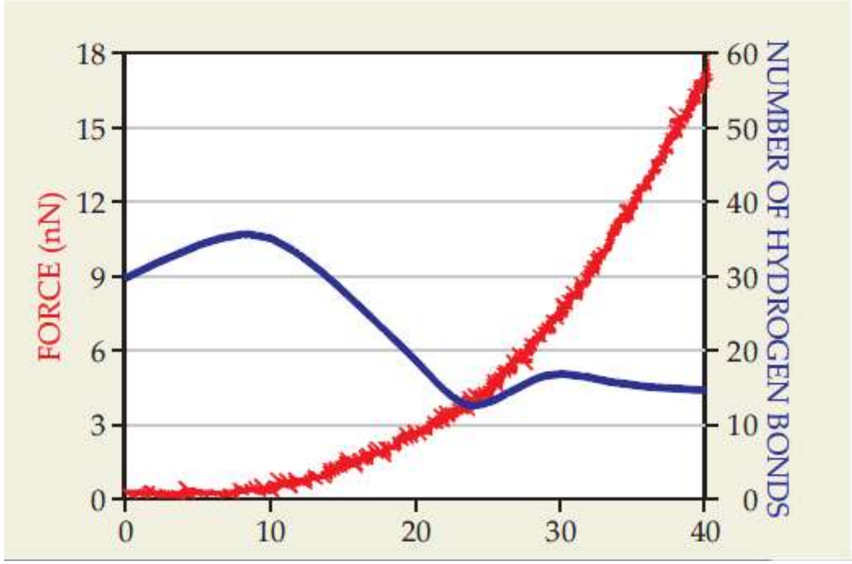
A triple helical tropocollagen molecule

Bone as a Composite Material

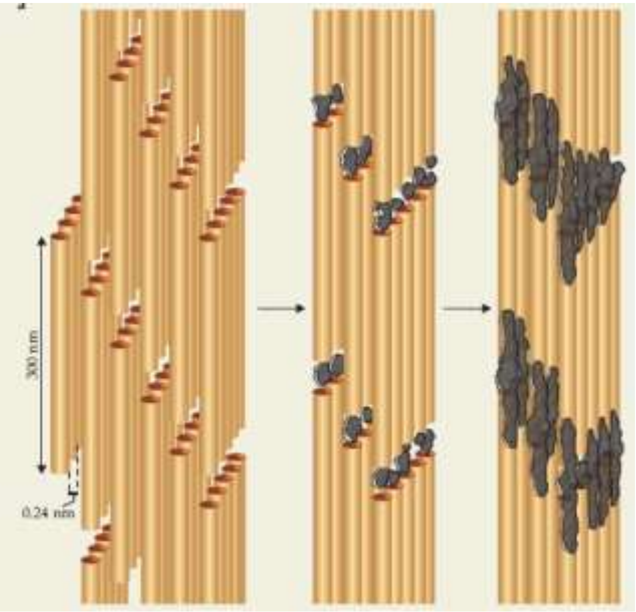
During 1st stage (10%) the number = 30 (~ 1 bond H for each amino-acid triplet)

As tropocollagen straightens H bonds break until 25% strain.

After that remains constant (~15) and appears to continuously break and re-form along the peptide



Bone as a Composite Material

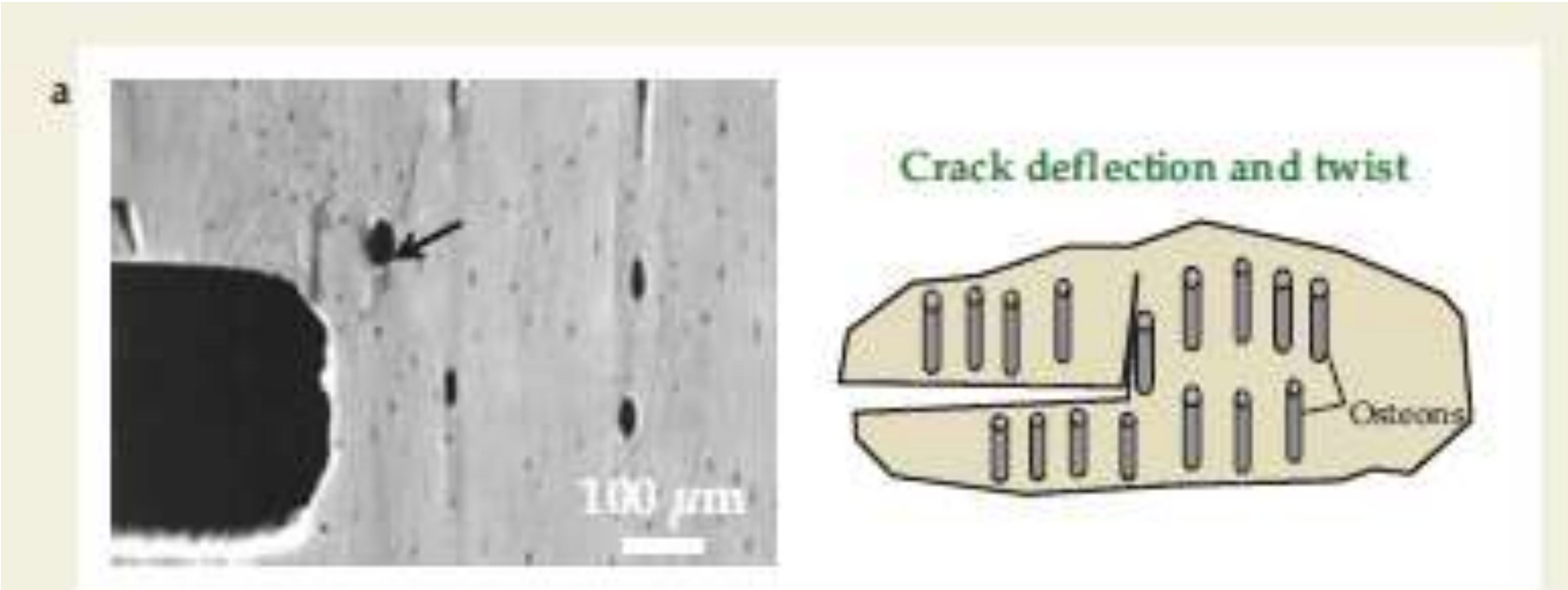


Hydroxyapatite (HA) in bone is critical for stiffness

Continuous glide between HA and tropocollagen molecules dissipates a great deal of energy as soon after yield point.

Fibrils deformation and the glue-like bonding material resists to separation during compressive loads in a vertebra mondel

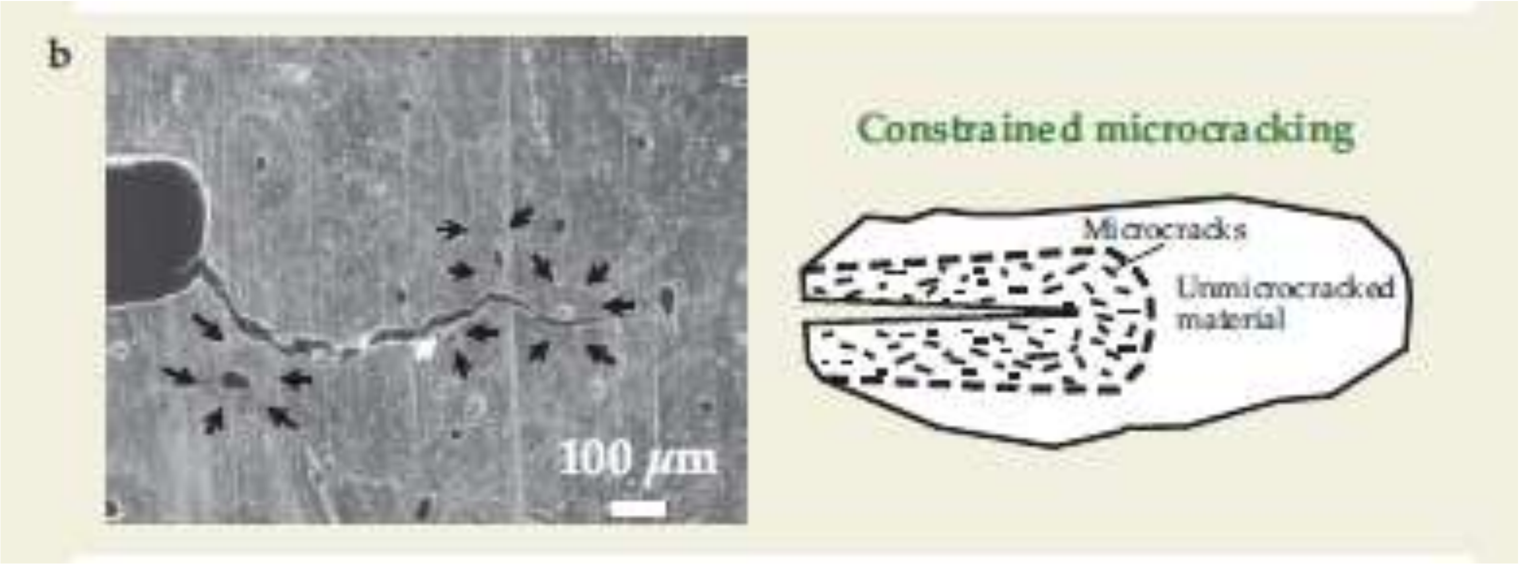
Toughness mecanism in bone



As a crack emerges encounters microcracks at the weak boundaries of the osteons

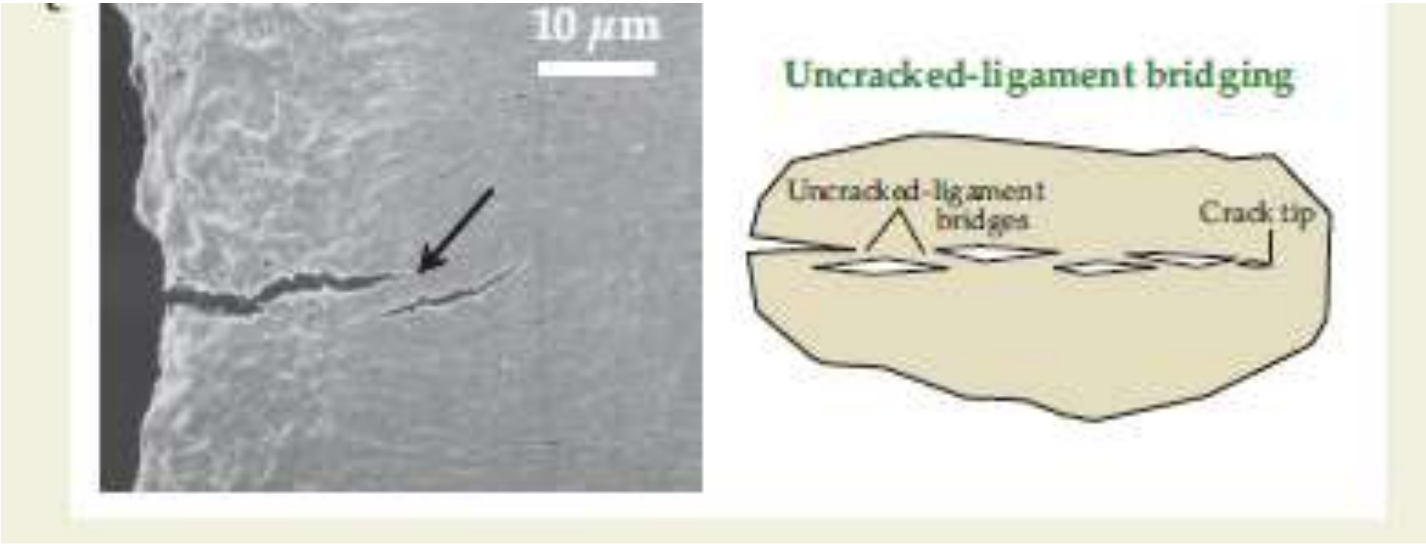
As these cracks consume energy may cause the main crack to deviate from the direction of the max tensile stress thus increasing bone toughness

Toughness mecanism in bone



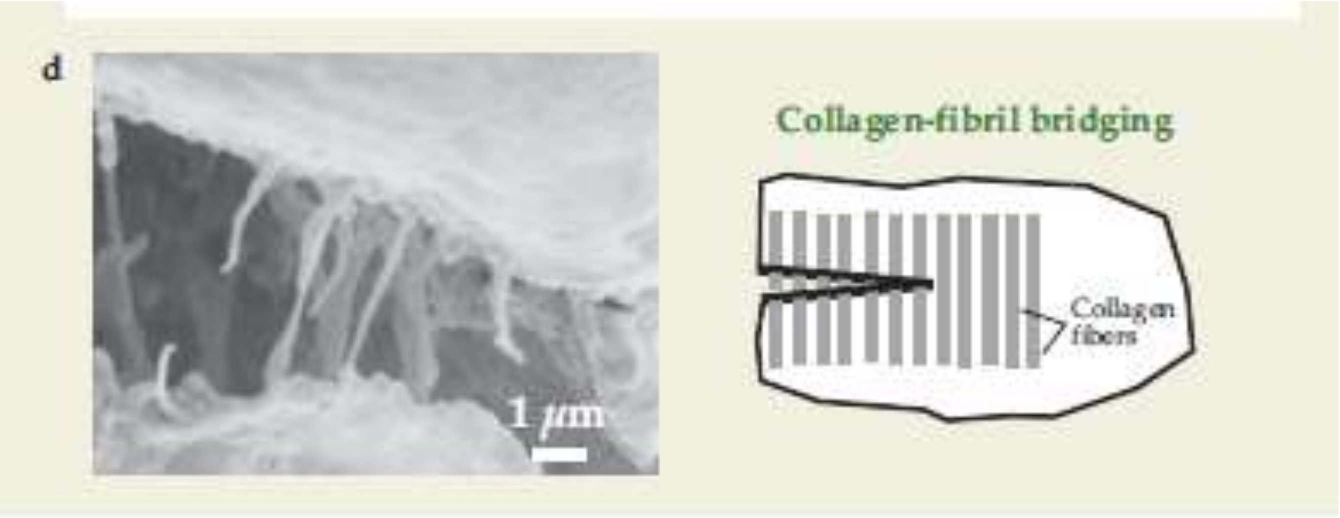
The tendency of bone to develop and the repair microcracks also increase toughness as the microcracks dilate the surrounding to main crack region, which actually compresses the crack.

Toughness mecanism in bone



An unbroken region (arrow) between a primary growing crack and another initiated ahead can carry significant loads to resist to crack spreading

Toughness mecanism in bone



Unbrokken collagen fibrils that bridge the gap formed in a crack resist its propagation by typically 0.1MPa $\sqrt{m}$.

JOURNAL OF THE MECHANICAL BEHAVIOR OF BIOMEDICAL MATERIALS 4 (2011) 1473–1482

available at www.sciencedirect.com

ScienceDirect

journal homepage: www.elsevier.com/locate/jmbbm

ELSEVIER

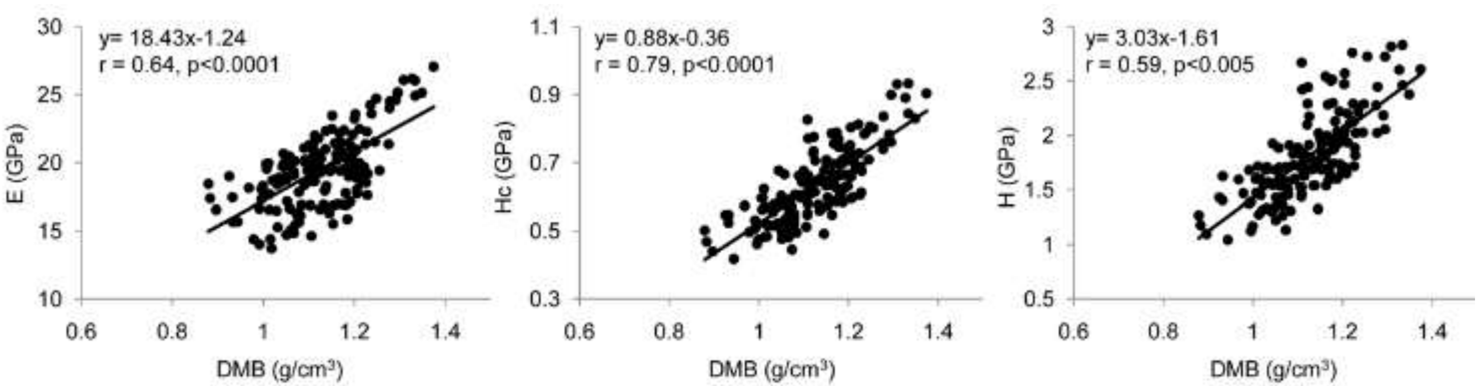
Research paper

Respective roles of organic and mineral components of human cortical bone matrix in micromechanical behavior: An instrumented indentation study

Y. Bala[a,b,c,*,1], *B. Depalle*[a,b,c,1], *T. Douillard*[d], *S. Meille*[d], *P. Clément*[d], *H. Follet*[a,b,c], *J. Chevalier*[d], *G. Boivin*[a]

The aim of this study is to analyze the links between ultrastructural properties and the mechanical behavior of bone tissue at the scale of osteon.

Elastic deformation was only explained by **degree of bone mineralization** whereas plastic deformation was more correlated with collagen maturity



elastic modulus (E), contact hardness (H) and true hardness (H).

JOURNAL OF THE MECHANICAL BEHAVIOR OF BIOMEDICAL MATERIALS 4 (2011) 1787–1796

Available online at www.sciencedirect.com
SciVerse ScienceDirect
journal homepage: www.elsevier.com/locate/jmbbm

Research paper

Similar damage initiation but different failure behavior in trabecular and cortical bone tissue

M.E. Szabó[a], J. Zekonyte[b], O.L. Katsamenis[a], M. Taylor[a], P.J. Thurner[a,*]

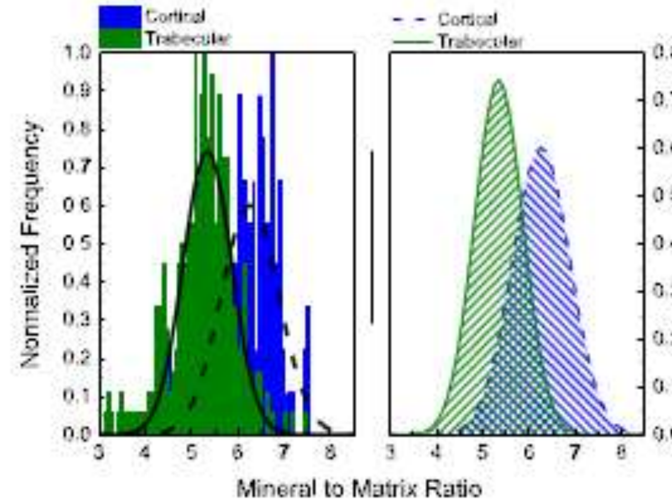
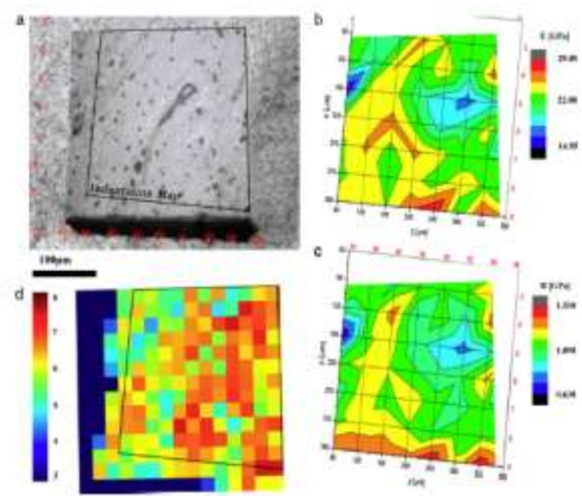


Fig. 5 – Comparison between the mineral to matrix ratios of cortical and trabecular bone samples. (left) Mineral to matrix ratios distribution and (right) Gaussian fit for these data sets.

While cortical bone was found to be overall stiffer and more highly mineralized, no apparent differences were noted in the distribution of modulus values or mineral density

Therefore, differences in the mechanical behavior of trabecular and cortical bone tissue are likely to be in large part due to microstructural and collagen related compositional differences.



Acta Biomaterialia 8 (2012) 1093–1100

Contents lists available at SciVerse ScienceDirect

Acta Biomaterialia

journal homepage: www.elsevier.com/locate/actabiomat

Sub-lamellar microcracking and roles of canaliculi in human cortical bone

Vincent Ebacher[a], Pierre Guy[b], Thomas R. Oxland[b], Rizhi Wang[a,*]

The present study examines the structure–microcracking relations at the lamellar and sub-lamellar levels of human cortical bone subjected to **compressive** loading.

Laser scanning confocal microscopy revealed a clear influence of the local structure and porosity of the Haversian systems' lamellae on microcrack development.

In particular, crack initiation and growth under transverse compression were associated with stress concentration at canaliculi.