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Kobry El-Kobbah,
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Bones adaptive loading mechanism

By

M. E. Abo-Elnor* M. S. Abo_Elkhair * A. E. Radi **

Abstract:

The skeleton in human being represents the infrastructure of the whole body. God; the biggest designer ever; has considered all the miss use of human being to their body. This skeleton consists of sets of functional bone arrangements; linked and working together with the ligaments and muscles. These linkages represent sometime a cantilever; others simulate closed and open linkage systems. But in general, we confidently say that all structures are all stable, represent an equilibrium state when free or loaded. Even though; the day movements and hard duties carried out by man adds more and more over stresses to this skeleton. Many references have handled the case of stress distribution, stress waves propagation and attenuation in these mechanisms [7, 8], but none of them has reached the concrete evidence to assure the complete knowledge revealing the clue; how this sophisticated structure withstands overloading that some times exceeds what the similar solid metal structures resist.

* Egyptian Armed Forces
** Professor of orthopedics, Faculty of medicine Ain-Shams Univ. Cairo, Egypt

2- Mechanical properties of bone

Bone, an organic material, can be considered in the same way as man-made engineering materials in many circumstances. However, it is likely to show much more variation in measured properties than typical engineering materials, due to the nature of its synthesis. The mechanical properties will vary as a function of:

- Age of the person
- Sex of the person
- Location in the body
- Temperature
- Mineral content
- Amount of water present
- Disease – e.g. osteoporosis

These variables can be to some extent 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 that any 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 the mechanical properties of bone must be considered in two orthogonal directions:

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

2.1 Modulus

Because bone can be considered to consist primarily of collagen fibers and an inorganic matrix, it can, on a simple level, be analyzed as a fiber composite. Composites are materials that are composed of two or more different components. They are commonly used in engineering and industry where the combination of the two materials creates a composite with properties that are superior to those of the individual 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:

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

Where:

E_f Young's Modulus of fibers

E_m Young's modulus of matrix
E_{ax}, E_{trans} Young's Modulus in axial and transverse directions
f volume fraction of fibers

These formulae indicate that the composite will be stiffer in the axial direction than the transverse. If this is applied to cortical bone it is seen that it will be stiffer in the direction parallel to the osteons, and therefore parallel to the long axis of the bone. However, this is only an approximation as it must be remembered that the collagen fibers are not aligned parallel to the axis of the osteons (see image of the microstructure of bone).

However, this composite model of the microstructure of bone is a highly simplified one. In addition, the bone mineral does not form a continuous matrix, but is in the form of discrete crystals. The actual structure of bone is much more complicated, and the use of the composite model in this way is an over-simplification. 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 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:

Table 1 Mechanical properties of bone constituents

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

2.2- Tensile and Compressive Strength

As mentioned on the structure of bones, 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 lot of variation in measured values of both the tensile and compressive strength of bone. Different bones in the body support different forces, so there is a large variation in strength between them. Plus, age is an important factor, with strength often decreasing as a person gets older.

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

2.3-Elasticity

Hydroxyapatite is a ceramic material and exhibits normal Hookean elastic behaviour– 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.

2.4-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. The cement lines 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. Similarly, if a crack is propagating perpendicular to an osteon, when it reaches a cement line it will change direction, the crack is blunted. This is illustrated in the figure below.

Therefore, although bone is classified as a brittle material, with the major component being mineral, its toughness is excellent. Bone's fracture energy, Gc, is approximately 1.5 kJ m-2, comparable to steel at low temperatures and wood when measured parallel to the grain. This is much tougher than man-made ceramics, due to the presence of the collagen fibers in bone which are viscoelastic – the stress-strain curves for loading and unloading are different meaning the elasticity is time-dependent, a common feature of fibrous proteins.

3- Bones structure

Bone is composed of a cellular component and an extracellular matrix. The cellular component is made of osteoblasts, bone-forming cells, osteoclasts, bone-destroying cells, and osteocytes, bone-maintaining cells which are inactive osteoblasts trapped in the extracellular matrix. The matrix, which is responsible for the mechanical strength of the bone tissue, is formed by an organic and a mineral phase. The organic phase is mainly composed of collagen fibers and the mineral phase of hydroxyapatite crystals. A liquid component is also present. The longitudinal section of human femur is shown in Figure 1.

Trabecular bone is quite porous and it is organized in *trabecules* oriented according to the direction of the physiological load, as shown in Figure 1. The configuration of the trabecular structures is highly variable and it depends on the anatomical site. Figures 3 and 4 show the difference between the structures of trabecules in the L1 vertebra (Figure 3) and in the calcaneus (Figure 4 later) in a 24 year old man:

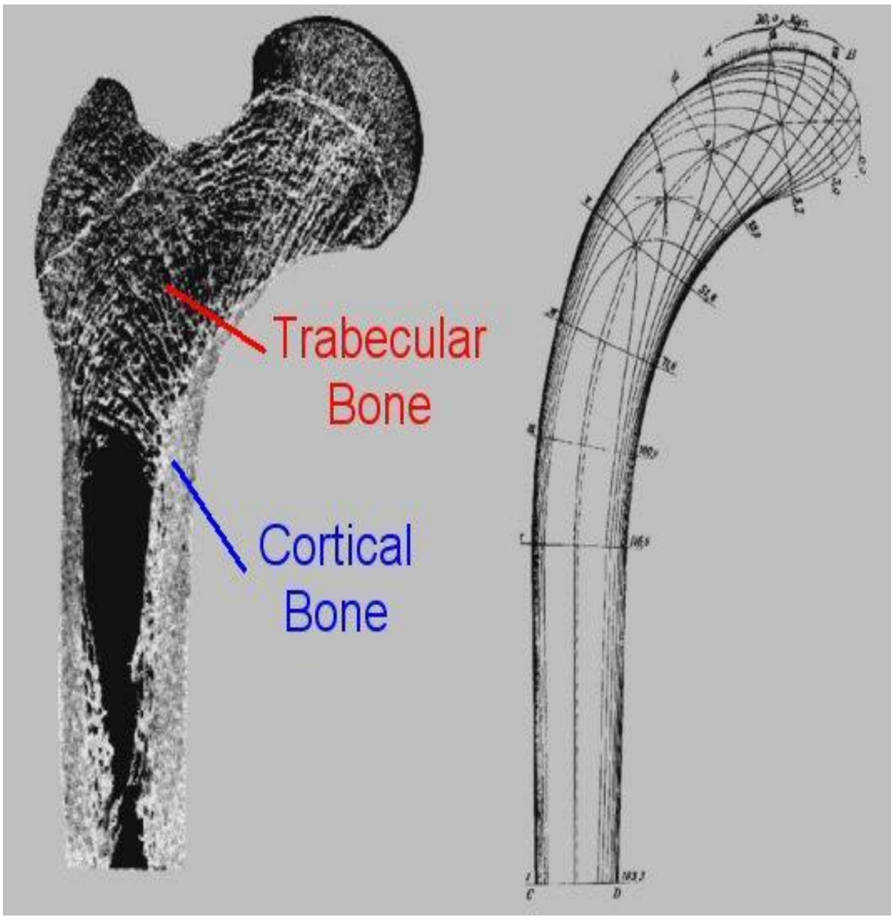

Figure 1: Longitudinal section of human femur.
The direction of principal stresses are shown in the scheme on the right.

The bone is clearly not homogeneous. Two main types of bone can be individuated: the *cortical bone* tissue and the *trabecular bone* tissue

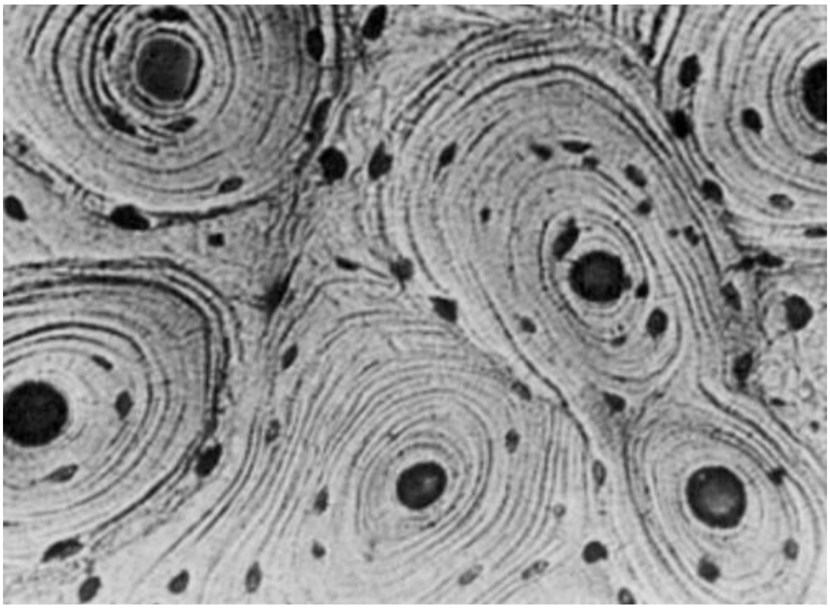

Figure 2: Lamellar structure of osteons in cortical bone

Cortical bone is the more dense tissue usually found on the surface of bones. It is organized in cylindrical shaped elements called *osteons* composed of concentric lamellae:

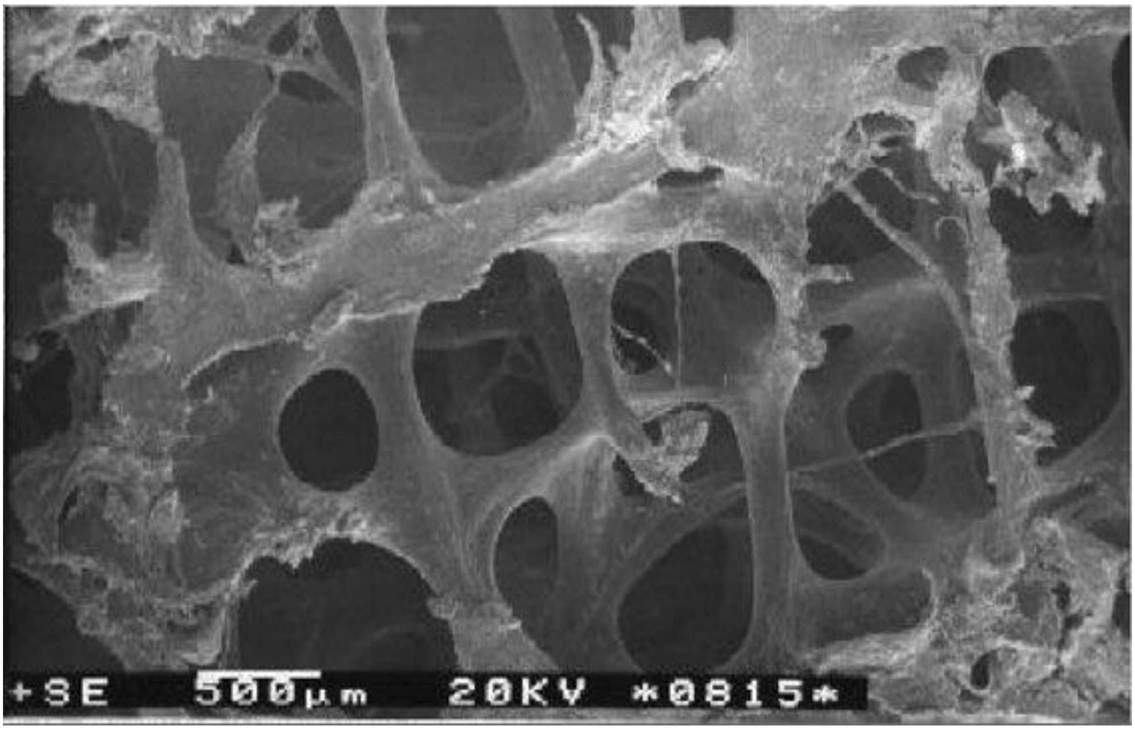

Figure 3: Trabecular structures in the L1 vertebra of a 24 year old

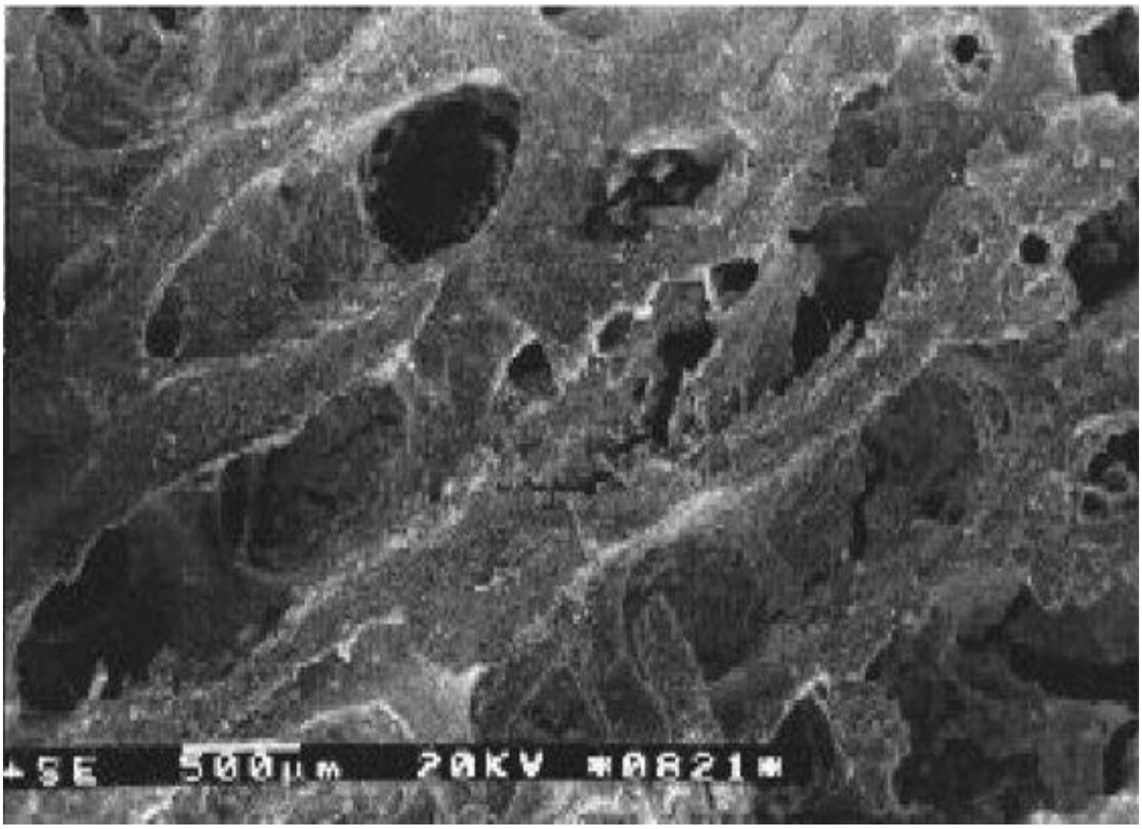

Figure 4: Trabecular structures in the calcaneus of a 24 year old

The effect of aging is show below. Figure 5 shows the trabecular structure of vertebrae in a 36 year old woman and Figure 6 in a 74 year old woman

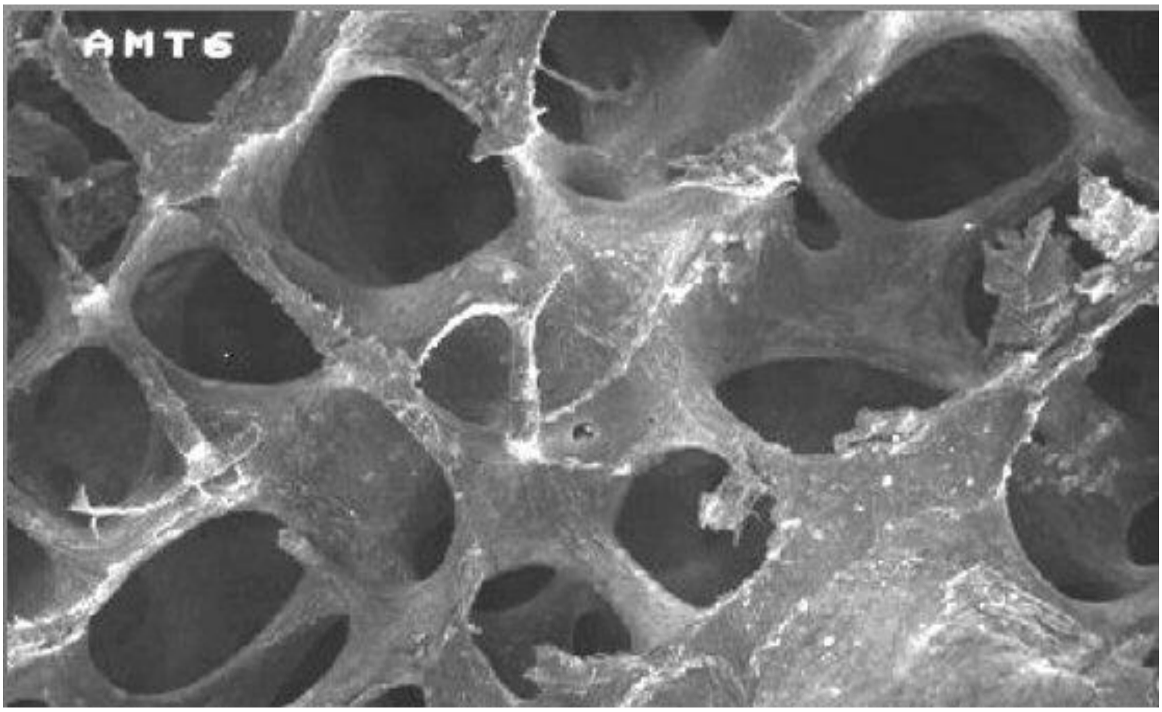

Figure 5: Trabecular structures of vertebrae in a 36 year old woman

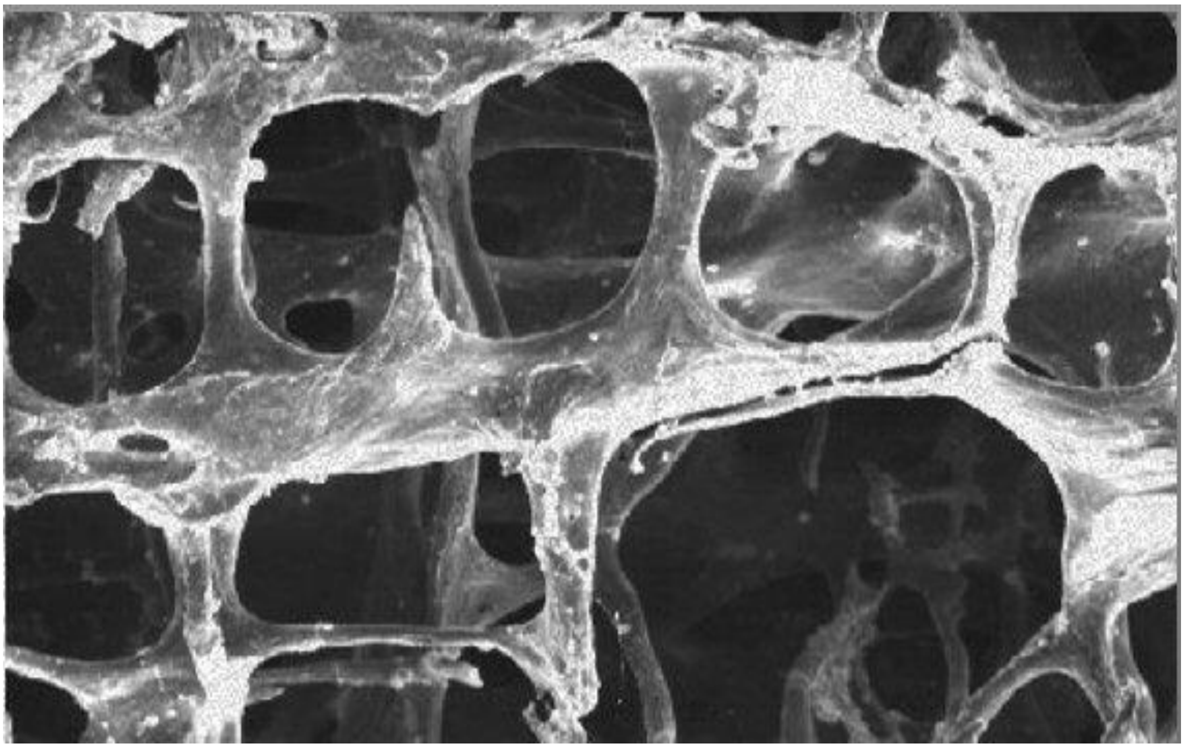

Figure 6: Trabecular structures of vertebrae in a 74 year old woman

4-Effect of Bone structure on mechanical properties

The different structures of cortical bone and trabecular bone result in different mechanical properties. Bone mechanical properties are highly variable according to species, age (Table1), anatomical site, liquid content, etc.

Table 1: Ultimate strength (MPa) and ultimate strain (%) of cortical bone from the human femur as a function of age

Property	Age (Year)						
	10-20	20-30	30-40	40-50	50-60	60-70	70-80
Ultimate Strength σ_u (MPa):							
Tension	114	123	120	112	93	86	86
Compression	--	167	167	161	155	145	-
Bending	151	173	173	162	154	139	139
Torsion	--	57	57	52	52	49	49
ε Ultimat Strain (%):							
Tension	1.5	1.4	1.4	1.3	1.3	1.3	1.3
Compression	--	1.9	1.8	1.8	1.8	1.8	--
Torsion	--	2.5	2.8	2.5	2.5	2.7	2.7

Cortical bone is an *anisotropic* material, meaning that its mechanical properties vary according to the direction of load (Figure 7). Cortical bone is often considered an *orthotropic* material. Orthotropic materials are a class of anisotropic materials characterized by three different Young's moduli E_1, E_2, E_3 according to the direction of load, three shear moduli G_{12}, G_{13}, G_{23} and six Poisson's ratios v_{12}, v_{13}, v_{23}, v_{21}, v_{31}, v_{32}.

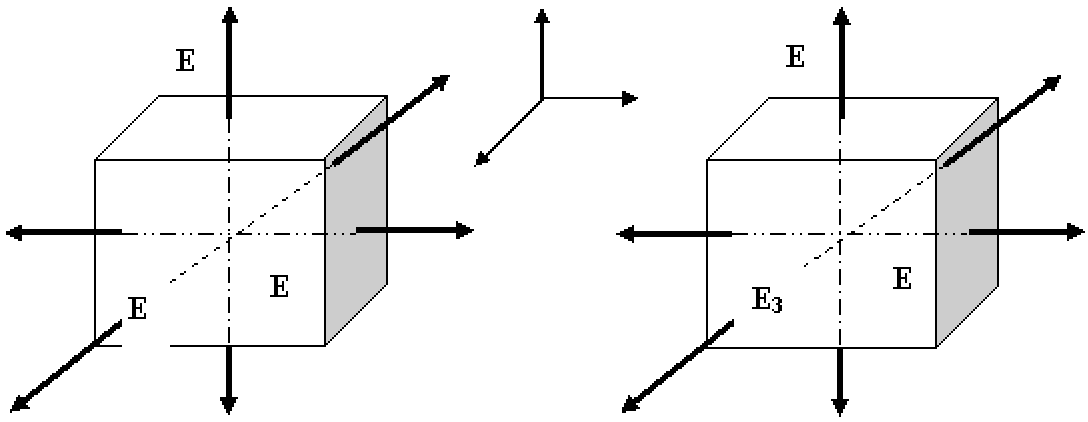

Figure 7: Comparison between the mechanical behaviour of isotropic and anisotropic materials

Table 2: Average elastic constants of mandible bone in corpus and ramus

	corpus	ramus
E_1[GPa]	10.93	11.77
E_2[GPa]	14.78	16.25
E_3[GPa]	18.89	20.42
G_{12}[GPa]	4.24	4.80
G_{13}[GPa]	5.13	5.72
G_{23}[GPa]	6.27	6.67
N_{12}	0.224	0.157
N_{13}	0.295	0.292
N_{23}	0.275	0.273
N_{21}	0.276	0.211
N_{31}	0.501	0.500
N_{32}	0.280	0.033

Table 2: Average elastic constants of mandible bone in corpus and ramus

	inferior	lingual	buccal
E_1[GPa]	10.63	10.85	11.04
E_2[GPa]	12.51	16.39	15.94
E_3[GPa]	19.75	18.52	18.06
G_{12}[GPa]	3.89	4.59	5.02
G_{13}[GPa]	4.85	5.45	4.97
G_{23}[GPa]	5.84	6.49	6.45
v_{12}	0.313	0.138	0.138
v_{13}	0.246	0.338	0.322
v_{23}	0.226	0.332	0.294
v_{21}	0.368	0.178	0.257
v_{31}	0.465	0.572	0.518
v_{32}	0.356	0.357	0.326

Table 4: Average elastic constants of human mandibular bone by tooth location

	Wisdom tooth	Second molar	1st. molar	2nd. Premolar	1st. Premolar	incisors	average
E_1[GPa]	11.59	11.66	11.65	10.14	10.13	10.64	10.95
E_2[GPa]	13.22	15.13	13.61	13.38	12.81	13.49	13.95
E_3[GPa]	17.66	19.82	20.44	16.40	17.42	19.60	18.30
G_{12}[GPa]	4.01	4.19	4.14	3.8	3.6	3.85	3.99
G_{13}[GPa]	4.23	4.98	4.97	4.43	4.72	4.93	4.83
G_{23}[GPa]	5.78	6.39	6.07	5.67	5.73	6.10	6.05
v_{12}	0.357	0.27	0.319	0.229	0.288	0.271	0.264
v_{13}	0.248	0.264	0.229	0.322	0.286	0.268	0.282
v_{23}	0.229	0.235	0.211	0.291	0.256	0.238	0.255
v_{21}	0.407	0.350	0.373	0.302	0.364	0.344	0.336
v_{31}	0.378	0.449	0.402	0.521	0.492	0.486	0.471
v_{32}	0.305	0.308	0.317	0.357	0.348	0.346	0.334

The mechanical characterization of trabecular bone is even more difficult. The mechanical properties of trabecular bone as a whole are due to the mechanical characteristics of single trabecules and to its highly porous structure. Figure 8 shows the dependence of the Young's modulus of trabecular bone from bone density.

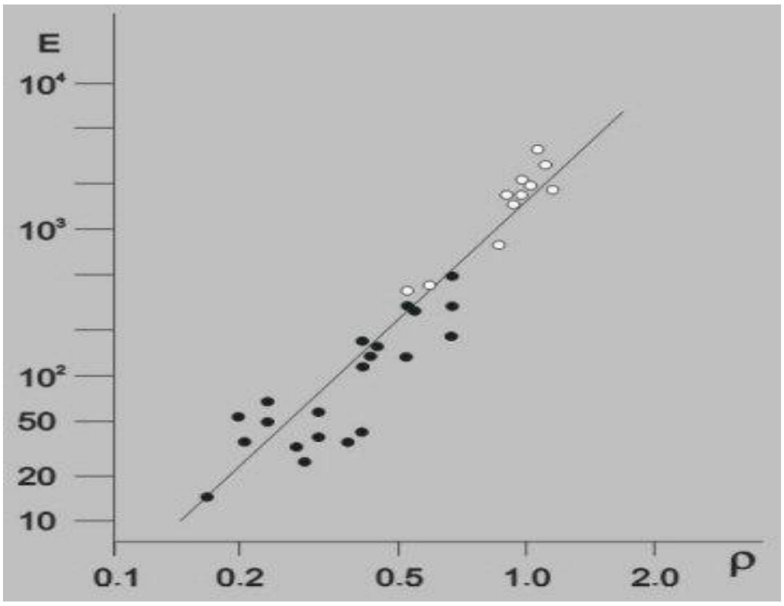

Figure 8: Young's modulus of trabecular bone as a function of density of bone. Bone density ρ is expressed in g/cm^3 and Young's modulus E in MPa.

5-Bone Remodeling

Bone adapts and remodels in response to the stress applied. *Wolff's law* states that bones develop a structure most suited to resist the forces acting upon them, adapting both the internal architecture and the external conformation to the change in external loading conditions. This change follows precise mathematical laws.

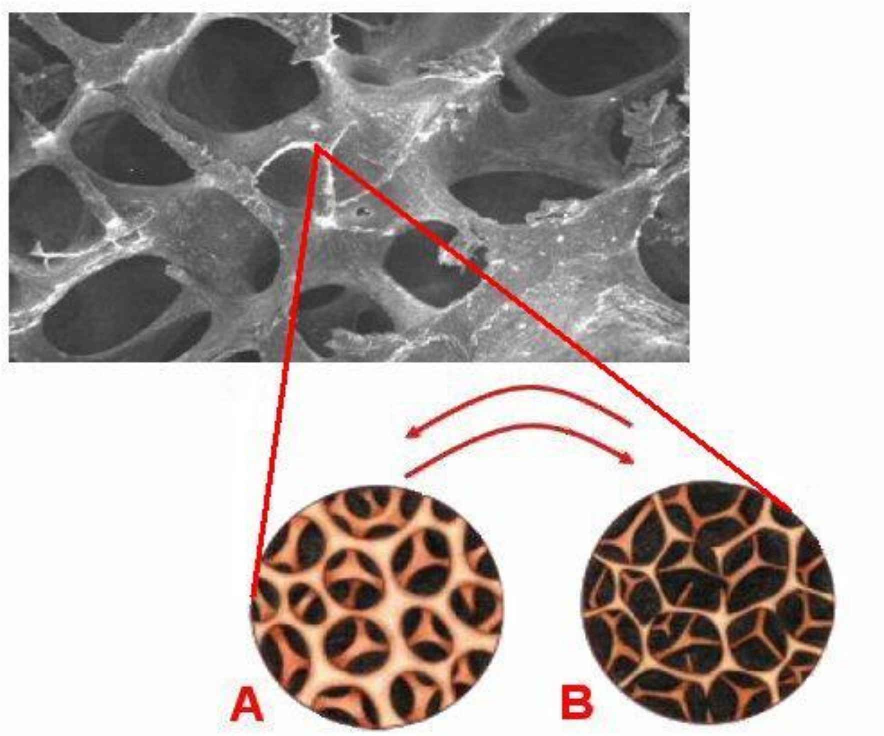

Figure 9: Bone remodeling: effect of reduction (from A to B) and of intensification of strain (from B to A) on bone trabecules

When a change in loading pattern occurs; stress and strain fields in the bone change accordingly. Bone tissue seems to be able to detect the change in strain on local bases and then adapts accordingly. The internal architecture is adapted in terms of change in density and in disposition of trabecules and osteons and the external conformation in terms of shape and dimensions. When strain is intensified new bone is formed. On a microscopic scale bone density is raised and on a macroscopic scale the bone external dimensions are incremented. When strain is lowered bone resorption takes place. On a microscopic scale bone density is lowered and on a macroscopic scale the bone external dimensions are reduced; as shown in figure 9 above.

When the change in strain is due to a change in direction of load on a microscopic scale the structure of trabecules and osteons is rearranged and on a macroscopic scale a change in bone shape may occur.Remodeling is carried out by the cellular component of bone. When resorption takes place osteoclasts reabsorb collagen and mineral phase (Figure 10A) which are then taken into the circulatory system (Figure 10B). During deposition osteoblasts group on the deposition surface and build the collagen network of bone (Figure 11A). Mineralization takes place afterwards (Figure 11B).

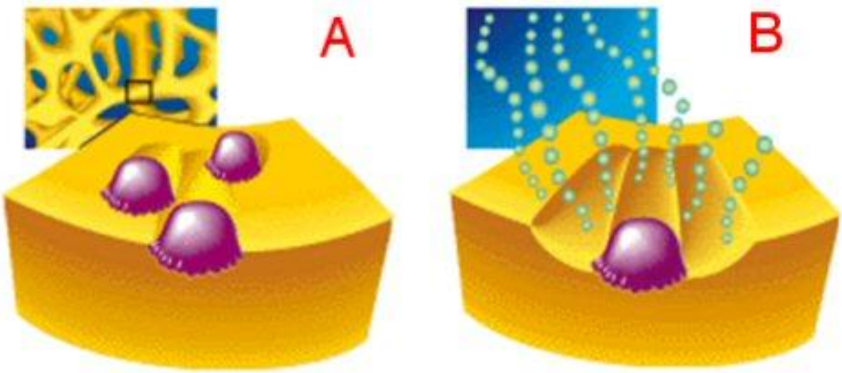

Figure 10: Bone resorption

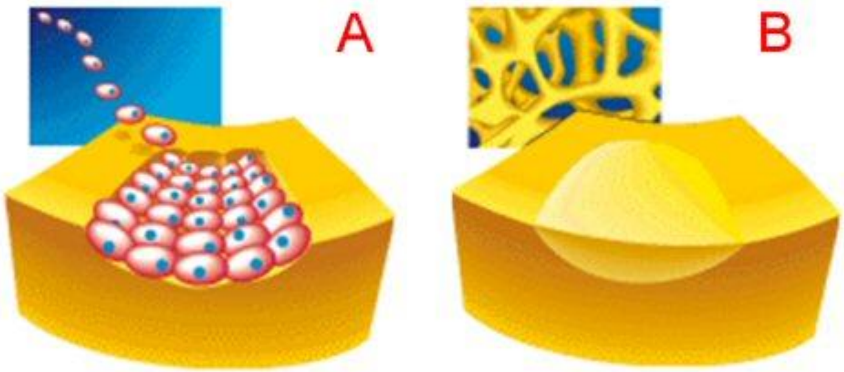

Figure 11: Bone deposition

Bone resorption and bone deposition processes are always active in bone. An equilibrium strain state exists in correspondence to which the two activities are perfectly balanced. When strain intensity is higher than the equilibrium strain deposition activity is more intense than resorption activity and net deposition occurs. When strain intensity is lower than the equilibrium strain deposition activity is less intense than resorption activity and net resorption occurs. Dynamical equilibrium between resorption and deposition is again achieved when the equilibrium strain state is newly established.

The cell-biology based *model of Davy and Hart* expresses functional dependence of bone remodeling on the strain field, based on cell activity (Figure 12). The load applied to the bone; together with geometric and material properties determine the local strain. Strain is detected by a transducer which generates the strain remodeling potential (SRP). This signal is modulated by genetic, hormonal and metabolic factors, generating the remodeling potential which regulates the recruitment rate and the activity of osteoblasts and osteoclasts, stimulating bone formation and bone resorption. The balance between bone deposition and bone resorption determines the net bone remodeling. Remodeling modifies bone geometric and material properties through a feedback loop.

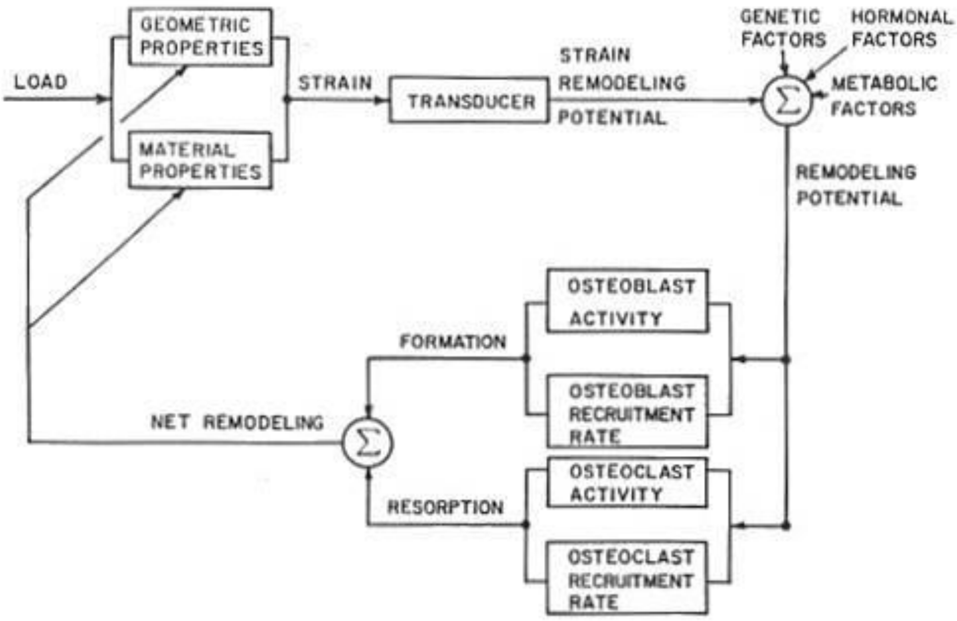

Figure 12: Schematic diagram of the Davy and Hart model for bone remodeling

10. Conclusion:

The mechanical properties of normal bone have been studied extensively [4]; but relatively little work has been published describing the GOD miracle that reveals how disease modifies these properties. Disorders increase the likelihood of fractures leading to a complete change in the mechanical properties of bones in the injured region after healing. This criteria activates the bone cells in the vicinity of defected zone to fill the interrications and adheres with the strange stem irregularities, this growth creates a set of mechanical trusses supporting the metallic stem and helps for regular distribution of loading, and leads to strengthening the joint producing in many cases fast healing and durable lasting hip.

References:

[1] Edwards, E.A., “Functional anatomy of the port systemic (porta-caval) communication”, Arch.Int.Med., 88-137, 1951.
[2] Gaynor Evans, F. “Mechanical properties of bones”, American lectures series No. 881, Springfield, Illinois, c.c. Thomas; 1973
[3] Stack, H.G., “The Palmer Fascia”, Edinburgh and London, Churchill Livingstone, New York, 1973.
[4] Reilly, D.T., Bursstien, A.H., “The mechanical properties of cortical bones”, J.of Bone joint surgery, 56-A, 1001-1022, 1974.

[5] Dickenson, R.P., Hutton, W.C. et al. " The Mechanical properties of bones in Osteoporosis", British Editorial Society of bone and joints surgery", Vol. 63-B, No.2, 1981.
[6] Cruess, R.L., Rennie, R.J., "Adult Orthopedics", Vol.1, 2, Churchill Livingstone, New York, 1984.
[7] Fish Fays, "Stress waves analysis"
[8] "Stress wave propagations in solids"