

Composite Biomaterials



Lecture Note (October 2013)

This lecture note is strictly for Biomedical Engineering Student in STEI – ITB attending EL 3004 and not for commercial purpose. Students are advised to read in advance “Mengenal Sifat Material” by Sudaryatno Sudirham and Ning Utari S.

Adopted from: Joseph D. Bronzino, “Biomedical Engineering Fundamentals”, CRC Press, third edition, 2006, Section V, CHAPTER 41, by Roderic Lakes.

Prepared by
Sudaryatno Sudirham (Lecturer)

www.darpublic.com

COMPOSITE BIOMATERIALS

[Adopted from: Joseph D. Bronzino, “*Biomedical Engineering Fundamentals*”, CRC Press, third edition, 2006, Section V, **CHAPTER 41**, by Roderic Lakes.]

Composite materials are solids which contain two or more distinct constituent materials or phases, on a scale larger than the atomic. The term ‘composite’ is usually reserved for those materials in which the distinct phases are separated on a scale larger than the atomic, and in which properties such as the elastic modulus are significantly altered in comparison with those of a homogeneous material. Accordingly, reinforced plastics such as fiberglass as well as natural materials such as bone are viewed as composite materials, but alloys such as brass are not. A foam is a composite in which one phase is empty space. Natural biological materials tend to be composites. Natural composites include bone, wood, dentin, cartilage, and skin. Natural foams include lung, cancellous bone, and wood. Natural composites often exhibit hierarchical structures in which particulate, porous, and fibrous structural features are seen on different micro-scales [Katz, 1980; Lakes, 1993]. In this segment, composite material fundamentals and applications in biomaterials (Park and Lakes, 1988) are explored. Composite materials offer a variety of advantages in comparison with homogeneous materials. These include the ability for the scientist or engineer to exercise considerable control over material properties. There is the potential for stiff, strong, lightweight materials as well as for highly resilient and compliant materials. In biomaterials, it is important that each constituent of the composite be biocompatible. Moreover, the interface between constituents should not be degraded by the body environment. Some applications of composites in biomaterial applications are (1) dental filling composites, (2) reinforced methyl methacrylate bone cement and ultra-high molecular weight polyethylene, and (3) orthopedic implants with porous surfaces.

41.1 STRUCTURE

The properties of composite materials depend very much upon structure. Composites differ from homogeneous materials in that considerable control can be exerted over the larger scale structure, and hence over the desired properties. In particular, the properties of a composite material depend upon the shape of the heterogeneities, upon the volume fraction occupied by them, and upon the interface among the constituents. The shape of the heterogeneities in a composite material is classified as follows. The principal inclusion shape categories are (1) the particle, with no long dimension, (2) the fiber, with one long dimension, and (3) the platelet or lamina, with two long dimensions, as shown in Figure 41.1. The inclusions may vary in size and shape within a category. For example, particulate inclusions may be spherical, ellipsoidal, polyhedral, or irregular. If one phase consists of voids, filled with air or liquid, the material is known as a cellular solid. If the cells are polygonal, the material is a honeycomb; if the cells are polyhedral, it is a foam. It is necessary in the context of biomaterials to distinguish the above structural cells from biological cells, which occur only in living organisms. In each composite structure,

se may moreover make the distinction between random orientation and preferred orientation.

41.2 BOUNDS ON PROPERTIES

Mechanical properties in many composite materials depend on structure in a complex way, however for some structures, the prediction of properties is relatively simple. The simplest composite structures are the idealized Voigt and Reuss models, shown in Figure 41.2. The dark and light areas in these diagrams represent the two constituent materials in the composite. In contrast to most composite structures, it is easy to calculate the stiffness of materials with the Voigt and Reuss structures, since in the Voigt structure the strain is the same in both constituents in the Reuss structure the stress is the same. The Young's modulus, E , of the Voigt composite is:

$$E = E_i V_i + E_m [1 - V_i] \quad (41.1)$$

in which E_i is the Young's modulus of the inclusions, and V_i is the volume fraction of inclusions, and E_m is the Young's modulus of the matrix. The Voigt relation for the stiffness is referred to as the rule of mixtures.

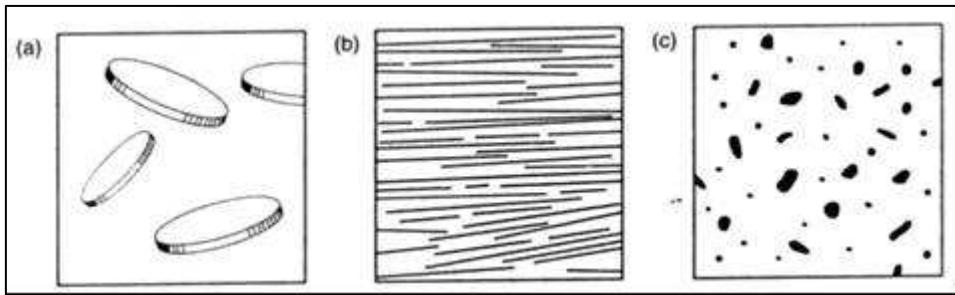


FIGURE 41.1 Morphology of basic composite inclusions. (a) Particle, (b) fiber, and (c) platelet.

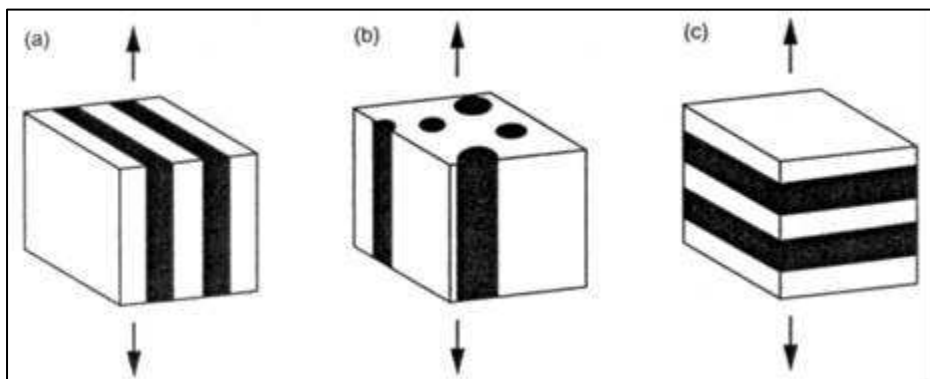


FIGURE 41.2 Voigt (a. laminar; b. fibrous) and Reuss (c) composite models, subjected to tension force indicated by arrows.

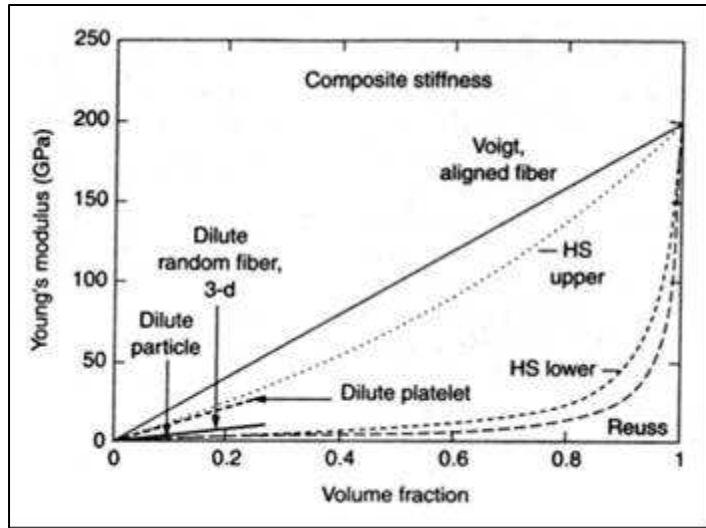


FIGURE 41.3 Stiffness vs. volume fraction for Voigt and Reuss models, as well as for dilute isotropic suspensions of platelets, fibers, and spherical particles embedded in a matrix. Phase moduli are 200 and 3 GPa.

The Reuss stiffness E ,

$$E = \left[\frac{V_i}{E_i} + \frac{1 - V_i}{E_m} \right]^{-1} \quad (41.2)$$

is less than that of the Voigt model. The Voigt and Reuss models provide upper and lower bounds respectively upon the stiffness of a composite of arbitrary phase geometry [Paul, 1960]. The bounds are far apart if, as is commonplace, the phase moduli differ a great deal, as shown in Figure 41.3. For composite materials which are isotropic, the more complex relations of Hashin and Shtrikman [1963] provide tighter bounds upon the moduli (Figure 41.3); both the Young's and shear moduli must be known for each constituent to calculate these bounds.

41.3 ANISOTROPY OF COMPOSITES

Observe that the Reuss laminate is identical to the Voigt laminate, except for a rotation with respect to the direction of load. Therefore, the stiffness of the laminate is anisotropic, that is, dependent on direction [Lekhnitskii, 1963; Nye, 1976; Agarwal and Broutman, 1980]. Anisotropy is characteristic of composite materials. The relationship between stress σ_{ij} and strain ϵ_{kl} in anisotropic materials is given by the tensorial form of Hooke's law as follows:

$$\sigma_{ij} = \sum_{k=1}^3 \sum_{l=1}^3 C_{ijkl} \epsilon_{kl} \quad (41.3)$$

Here C_{ijkl} is the elastic modulus tensor. It has $3^4 = 81$ elements, however since the stress and strain are represented by symmetric matrices with six independent elements each, the number of independent modulus tensor elements is reduced to

36. An additional reduction to 21 is achieved by considering elastic materials for which a strain energy function exists. Physically, C_{2323} represents a shear modulus since it couples a shear stress with a shear strain. C_{111} couples axial stress and strain in the 1 or x direction, but it is not the same as Young's modulus. The reason is that Young's modulus is measured with the lateral strains free to occur via the Poisson effect, while C_{111} is the ratio of axial stress to strain when there is only one nonzero strain value; there is no lateral strain. A modulus tensor with 21 independent elements describes a triclinic crystal, which is the least symmetric crystal form. The unit cell has three different oblique angles and three different side lengths. A triclinic composite could be made with groups of fibers of three different spacings, oriented in three different oblique directions. Triclinic modulus elements such as C_{2311} , known as cross-coupling constants, have the effect of producing a shear stress in response to a uniaxial strain; this is undesirable in many applications. An orthorhombic crystal or an orthotropic composite has a unit cell with orthogonal angles. There are nine independent elastic moduli. The associated engineering constants are three Young's moduli, three Poisson's ratios, and three shear moduli; the cross-coupling constants are zero when stresses are aligned to the symmetry directions. An example of such a composite is a unidirectional fibrous material with a rectangular pattern of fibers in the cross-section. Bovine bone, which has a laminated structure, exhibits orthotropic symmetry, as does wood. In a material with hexagonal symmetry, out of the nine C elements, there are five independent elastic constants. For directions in the transverse plane the elastic constants are the same, hence the alternate name transverse isotropy. A unidirectional fiber composite with a hexagonal or random fiber pattern has this symmetry, as does human Haversian bone. In cubic symmetry, there are three independent elastic constants, a Young's modulus, E , a shear modulus, G , and an independent Poisson's ratio, ν . Cross-weave fabrics have cubic symmetry. Finally, an isotropic material has the same material properties in any direction. There are only two independent elastic constants, hence E , G , ν , and also the bulk modulus B are related in an isotropic material. Isotropic materials include amorphous solids, polycrystalline metals in which the grains are randomly oriented, and composite materials in which the constituents are randomly oriented.

Anisotropic composites offer superior strength and stiffness in comparison with isotropic ones. Material properties in one direction are gained at the expense of properties in other directions. It is sensible, therefore, to use anisotropic composite materials only if the direction of application of the stress is known in advance.

41.4 PARTICULATE COMPOSITES

It is often convenient to stiffen or harden a material, commonly a polymer, by the incorporation of particulate inclusions. The shape of the particles is important [see Christensen, 1979]. In isotropic systems, stiff platelet (or flake) inclusions are the most effective in creating a stiff composite followed by fibers; and the least effective geometry for stiff inclusions is the spherical particle, as shown in Figure 41.3. A dilute concentration of spherical particulate inclusions of stiffness E_i and

volume fraction V_i , in a matrix (with Poisson's ratio assumed to be 0.5) denoted by the subscript m , gives rise to a composite with a stiffness E :

$$E = \frac{5(E_i - E_m)V_i}{3 + 2(E_i/E_m)} + E_m \quad (41.4)$$

The stiffness of such a composite is close to the Hashin.-Shtrikman lower bound for isotropic composites. Even if the spherical particles are perfectly rigid compared with the matrix, their stiffening effect at low concentrations is modest. Conversely, when the inclusions are more compliant than the matrix, spherical ones reduce the stiffness the least and platelet ones reduce it the most. Indeed, soft platelets are suggestive of crack-like defects. Soft platelets, therefore result not only in a compliant composite, but also a weak one. Soft spherical inclusions are used intentionally as crack stoppers to enhance the toughness of polymers such as polystyrene (high impact polystyrene), with a small sacrifice in stiffness.

Particle reinforcement has been used to improve the properties of bone cement. For example, inclusion of bone particles in PMMA cement somewhat improves the stiffness and improves the fatigue life considerably [Park et al., 1986]. Moreover, the bone particles at the interface with the patient's bone are ultimately resorbed and are replaced by ingrown new bone tissue. This approach is in the experimental stages.



FIGURE 41.4 Microstructure of a dental composite. Miradapt^o (Iobnson & Iohnsonj) 50% by volume filler: barium glass and colloidal silica [Park and Lakes, 1992].

Rubber used in catheters, rubber gloves, etc. is usually reinforced with very fine particles of silica (SiO_2) to make the rubber stronger and tougher.

Teeth with decayed regions have traditionally been restored with metals such as silver amalgam. Metallic restorations are not considered desirable for anterior teeth for cosmetic reasons. Acrylic resins and silicate cements had been used for anterior teeth, but their poor material properties led to short service uk and clinical failures. Dental composite resins have virtually replaced these materials and are very commonly used to restore posterior teeth as well as anterior teeth [Cannon, 1988].

The dental composite resins consist of a polymer matrix and stiff inorganic inclusions [Craig, 1981]. A representative structure is shown in Figure 41.4. The particles are very angular in shape. The inorganic inclusions confer a relatively high stiffness and high wear resistance on the material. Moreover, since they are translucent and their index of refraction is similar to that of dental enamel, they are cosmetically acceptable. Available dental composite resins use quartz, barium glass, and colloidal silica as fillers. Fillers have particle size from 0.04 to 13 μm , and concentrations from 33 to 78% by weight. In view of the greater density of the inorganic filler phase, a 77% weight percent of filler corresponds to a volume percent of about 55%. The matrix consists of a polymer, typically BIS-GMA. In restoring a cavity, the dentist mixes several constituents, then places them in the prepared cavity to polymerize. For this procedure to be successful the viscosity of the mixed paste must be sufficiently low and the polymerization must be controllable. Low viscosity liquids such as triethylene glycol dimethacrylate are used to lower the viscosity and inhibitors such as BHT (butylated hydroxytoluene) are used to prevent premature polymerization. Polymerization can be initiated by a thermochemical initiator such as benzoyl peroxide, or by a photochemical initiator (benzoin alkyl ether) which generates free radicals when subjected to ultraviolet light from a lamp used by the dentist.

Dental composites have a Young's modulus in the range 10 to 16 GPa and the compressive strength from 170 to 260 MPa [Cannon, 1988]. As shown in Table 41.1, these composites are still considerably less stiff than dental enamel, which contains about 99% mineral. Similar high concentrations of mineral particles in synthetic composites cannot easily be achieved, in part because the particles do not pack densely. Moreover, an excessive concentration of particles raises the viscosity of the unpolymerized paste. An excessively high viscosity is problematical since it prevents the dentist from adequately packing the paste into the prepared cavity the material will then fill in crevices less effectively.

The thermal expansion of dental composites, as with other dental materials, exceeds that of tooth structure. Moreover, there is a contraction during polymerization of 1.2 to 1.6%. These effects are thought to contribute to leakage of saliva, bacteria, etc., at the interface margins. Such leakage in some cases can cause further decay of the tooth.

Use of colloidal silica in the so-called "microfilled" composites allows these resins to be polished, so that less wear occurs and less plaque accumulates. It is more difficult, however, to make these with a high fraction of filler. All the dental composites exhibit creep. The stiffness changes by a factor of 2.5 to 4 (depending on the particular material) over a time period from 10 sec to 3 h under steady load [Papadogianis et al, 1985]. This creep may result in indentation of the restoration, but wear seems to be a greater problem. Dental composite resins have become established as restorative materials for both anterior and posterior teeth. The use of these materials is likely to increase as improved compositions are developed and in response to concern over long term toxicity of silver-mercury amalgam fillings.

41.5 FIBROUS COMPOSITES

Fibers incorporated in a polymer matrix increase the stiffness, strength, fatigue life, and other properties [Agarwal and Broutman, 1980; Schwartz, 1992]. Fibers are mechanically more effective in achieving a stiff, strong composite than are particles. Materials can be prepared in fiber form with very few defects which concentrate stress. Fibers such as graphite are stiff (Young's modulus is 200-800 GPa) and strong (the tensile strength is 2.7—5.5 GPa). Composites made from them can be as strong as steel but much lighter, as shown in Table 41.1. The stiffness of a composite with aligned fibers, if it is loaded along the fibers, is equivalent to the Voigt upper bound, Equation 41.1. Unidirectional fibrous composites, when

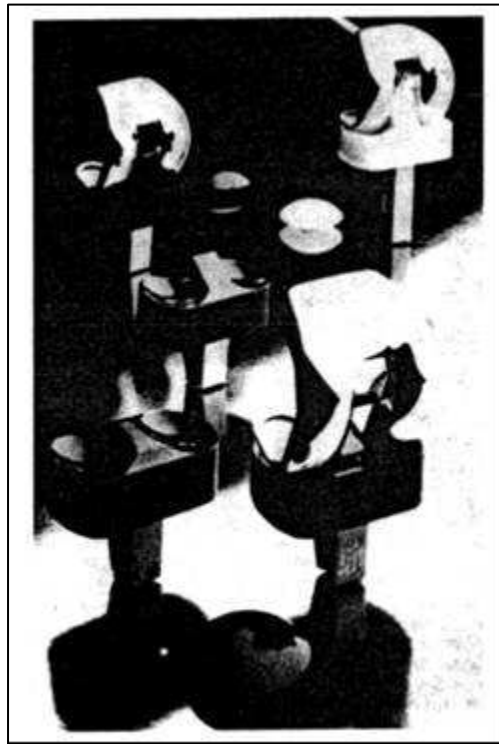


FIGURE 41.5 Knee prostheses with polyethylene tibial components reinforced with carbon fiber.

loaded along the fibers, can have strengths and stiffness comparable to that of steel, but with much less weight (Table 41.1). However if it is loaded transversely to the fibers, such a composite will be compliant, with a stiffness not much greater than that of the matrix alone. While unidirectional fiber composites can be made very strong in the longitudinal direction, they are weaker than the matrix alone when loaded transversely, as a result of stress concentration around the fibers. If stiffness and strength are needed in all directions, the fibers may be oriented randomly. For such a three-dimensional isotropic composite, for a low concentration of fibers,

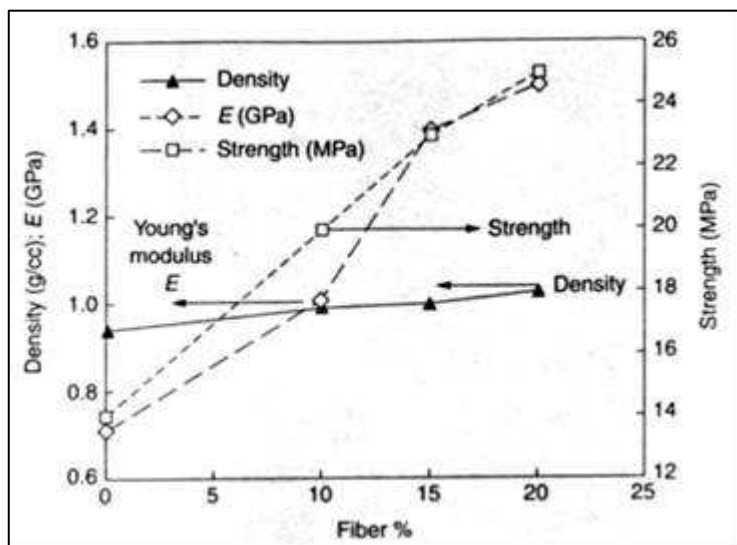
$$E = \frac{E_i V_i}{6} + E_m \quad (41.5)$$

so the stiffness is reduced by about a factor of six in comparison with an aligned composite as illustrated in Figure 41.3. However if the fibers are aligned randomly in a plane, the reduction in stiffness is only a factor of three. The degree of anisotropy in fibrous composites can be very well controlled by forming laminates consisting of layers of fibers embedded in a matrix. Each layer can have fibers oriented in a different direction. One can achieve quasi-isotropic behavior in the laminate plane; such a laminate is not as strong or as stiff as a unidirectional one, as illustrated in Table 41.1. Strength of composites depends on such particulars as the brittleness or ductility of the inclusions and the matrix. In fibrous composites failure may occur by (1) fiber breakage, buckling, or pullout, (2) matrix cracking, or (3) debonding of fiber from matrix.

Short fiber composites are used in many applications. They are not as stiff or as strong as composites with continuous fibers, but they can be formed economically by injection molding or by in situ polymerization. Choice of an optimal fiber length can result in improved toughness, due to the predominance of fiber pull-out as a fracture mechanism.

Carbon fibers have been incorporated in the high density polyethylene used in total knee replacements (Figure 41.5). The standard ultra high molecular weight polyethylene (UHMWPE) used in these implants is considered adequate for most purposes for implantation in older patients. A longer wear-free implant lifetime is desirable for use in younger patients. It is considered desirable to improve the resistance to creep of the polymeric component, since excessive creep results in an indentation of that component after long term use. Representative properties of carbon reinforced ultra high molecular weight polyethylene are shown in Figure 41.6 [Sclippa and Piekarski, 1973]. Enhancements of various properties by a factor of two are feasible.

FIGURE 41.6
Properties of
carbon fiber
reinforced ultra
high molecular
weight
polyethylene.
(Replotted from
Sclippa, E. and
Piekarski, K. 1973.
*J. Biomed. Mater.
Res.*,7. 39—70.
With permission.)



Polymethylmethacrylate (PMMA) used in bone cement is compliant and weak in comparison with bone. Therefore several reinforcement methods have been attempted. Metal wires have been used clinically as macroscopic “fibers” to reinforce PMMA cement used in spinal stabilization surgery [Fishbane and Pond, 1977]. The wires are made of a biocompatible alloy such as cobalt—chromium alloy or stainless steel. Such wires are not currently used in joint replacements owing to the limited space available. Graphite fibers have been incorporated in bone cement [Knoell et al., 1975] on an experimental basis. Significant improvements in the mechanical properties have been achieved. Moreover, the fibers have an added beneficial effect of reducing the rise in temperature which occurs during the polymerization of the PMMA in the body. Such high temperature can cause problems such as necrosis of a portion of the bone into which it is implanted. Thin, short titanium fibers have been embedded in PMMA cement [Topoleski et al. 1992]; a toughness increase of 51% was observed with a 5% volumetric fiber content. Fiber reinforcement of PMMA cement has not found much acceptance since the fibers also increase the viscosity of the unpolymerized material. It is consequently difficult for the surgeon to form and shape the polymerizing cement during the surgical procedure.

Metals are currently used in bone plates for immobilizing fractures and in the femoral component of total hip replacements. A problem with currently used implant metals is that they are much stiffer than bone, so they shield the nearby bone from mechanical stress. Stress-shielding results in a kind of disuse atrophy: the bone resorbs [Engh and Bobyn. 1988]. Therefore composite materials have been investigated as alternatives [Bradley et al., 1991; Skinner, 1988]. Fibrous composites can deform to higher strains (to about 0.01) than metals (0.001 for a mild steel) without damage. This resilience is an attractive characteristic for more flexible bone plates and femoral stems. Flexible composite bone plates are effective in promoting healing [Jockish, 1992]. Composite hip replacement prostheses have been made with carbon fibers in a matrix of polysulfone and polyetherether ketone (PEEK). These prostheses experience heavy load with a static component. Structural metals such as stainless steel and cobalt chromium alloys do not creep significantly at room or body temperature. In composites which contain a polymer constituent, creep behavior is a matter of concern. The carbon fibers exhibit negligible creep, but polymer constituents tend to creep. Prototype composite femoral components were found to exhibit fiber dominated creep of small magnitude and are not expected to limit the life of the implant [Maharaj and Jamison, 1993].

Fibrous composites have also been used in external medical devices such as knee braces [Yeaple, 1989], in which biocompatibility is not a concern but light weight is crucial.

TABLE 41.1 Properties of Bone, Teeth, and Biomaterials

Material	Young's modulus E (GPa)	Density ρ (g/cm^3)	Strength (MPa)	References
Hard Tissue				
Tooth, bone, human compact bone, longitudinal direction	17	1.8	130 (tension)	Craig and Peyton, 1958; Reilly and Burstein, 1975; Peters et al., 1984; Park and Lakes, 1992
Tooth dentin	18	2.1	138 (compression)	Park and Lakes, 1992
Tooth enamel	50	2.9		
Polymers				
Polyethylene (UHMW)	1	0.94	30 (tension)	Park and Lakes, 1992
Polymethyl methacrylate, PMMA	3	1.1	65 (tension)	
PMMA bone cement	2	1.18	30 (tension)	
Metals				
316L Stainless steel (wrought)	200	7.9	1000 (tension)	Park and Lakes, 1992
Co-Cr-Mo (cast)	230	8.3	660 (tension)	
Co Ni Cr Mo (wrought)	230	9.2	1800 (tension)	
Ti6Al4V	110	4.5	900 (tension)	
Composites				
Graphite-epoxy (unidirectional fibrous, high modulus)	215	1.63	1240 (tension)	Schwartz, 1997
Graphite-epoxy (quasi-isotropic fibrous)	46	1.55	579 (tension)	Schwartz, 1997
Dental composite resins (particulate)	10-16		170-260 (compression)	Cannon, 1988
Foams				
Polymer foams	10^{-4} -1	0.002-0.8	0.01-1 (tension)	Gibson and Ashby, 1988

41.6 POROUS MATERIALS

The presence of voids in porous or cellular solids will reduce the stiffness of the material. For some purposes, that is both acceptable and desirable. Porous solids are used for many purposes: flexible structures such as (1) seat cushions, (2) thermal insulation, (3) filters, (4) cores for stiff and lightweight sandwich panels, (5) flotation devices, and (6) to protect objects from mechanical shock and vibration; and in biomaterials, as coatings to encourage tissue ingrowth. Representative cellular solid structures are shown in Figure 41.7.

The stiffness of an open-cell foam is given by (Gibson and Ashby, 1988)

$$E = E_s [V_s]^2 \quad (41.6)$$

in which E_s is the Young's modulus and V_s is the volume fraction of the solid phase of the foam; V_s is also called the relative density.

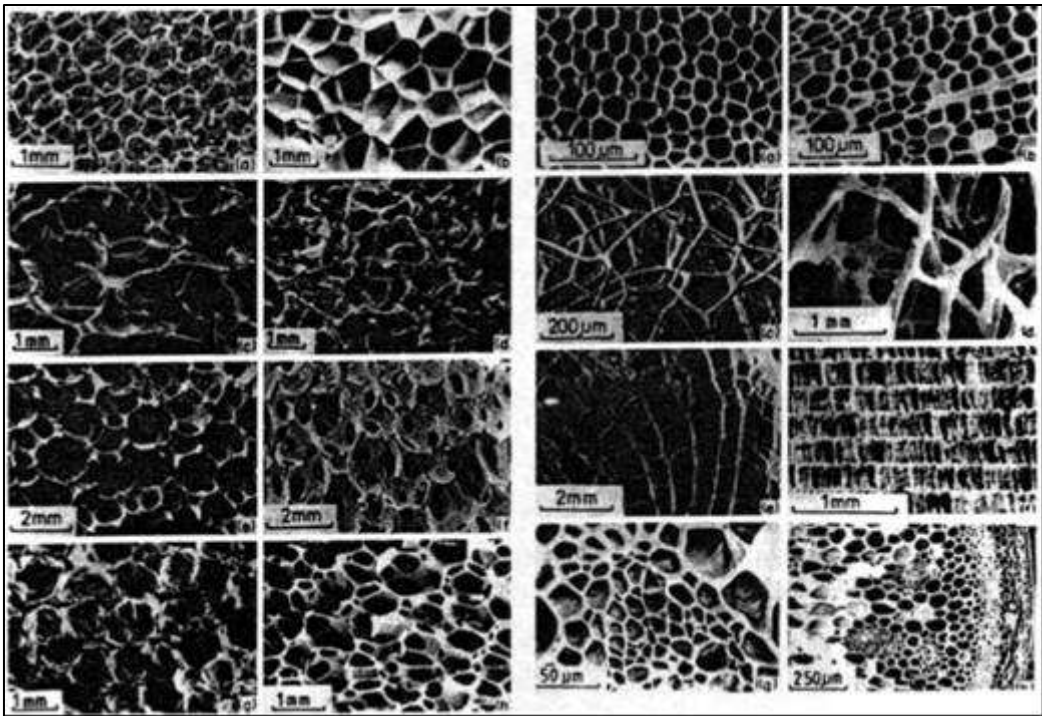


FIGURE 41.7 Cellular solids structures, after Gibson and Ashby (1988). Left: synthetic cellular solids: (a) open-cell polyurethane. (b) closed-cell polyethylene. (c) foamed nickel, (d) foamed copper, (e) foamed zirconia. (f) foamed mullite. (g) foamed glass. (h) polyester foam with both open and closed cells. Right: natural cellular solids: (a) cork, (b) balsa wood, (c) sponge. (d) cancellous bone, (e) coral, (f) cuttlefish bone. (g) iris leaf. (h) plant stalk.

The strength for crushing of a brittle foam and the elastic collapse of an elastomeric foam is given,

$$\sigma_{\text{crush}} = 0.65 \sigma_{fs} [V_s]^{3/2} \quad (41.7)$$

$$\sigma_{\text{coll}} = 0.05 E_s [V_s]^2 \quad (41.8)$$

Here σ_{fs} is the fracture strength of the solid phase. These strength relations are valid for relatively small density. Their derivation is based on the concept of bending of the cell ribs and is presented by Gibson and Ashby [1988]. Most man-made closed cell foams tend to have a concentration of material at the cell edges, so that they behave mechanically as open cell foams. The salient point in the relations for the mechanical properties of cellular solids is that the relative density dramatically influences the stiffness and the strength.

As for the relationship between stress and strain, a representative stress strain curve is shown in Figure 41.8. The physical mechanism for the deformation mode beyond the elastic limit depends on the material from which the foam is made. Trabecular

bone, for example, is a natural cellular solid, which tends to fail in compression by crushing. Many kinds of trabecular bone appear to behave mechanically as an open cell foam. For trabecular bone of unspecified orientation, the stiffness is proportional to the cube of the density and the strength as the square of the density (Gibson and Ashby, 1988), which indicates behavior dominated by bending of the trabeculae. For bone with oriented trabeculae, both stiffness and strength in the trabecular direction are proportional to the density, a fact which indicates behavior dominated by axial deformation of the trabeculae.

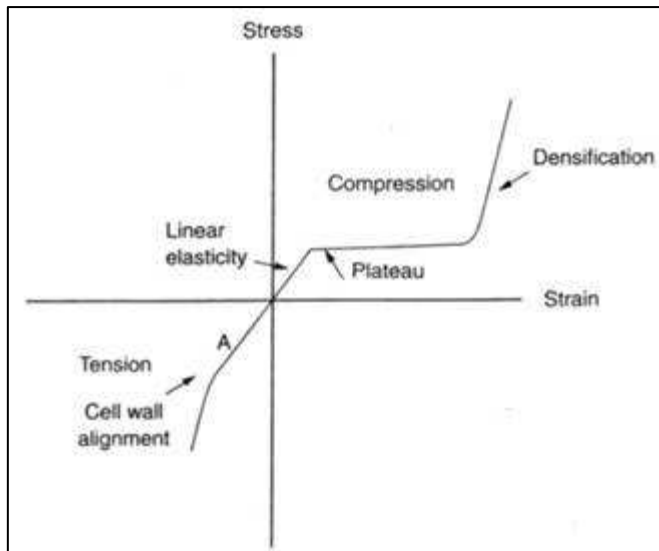


FIGURE 41.8 Representative stress-strain curve for a cellular solid. The plateau region for compression in the case of elastomeric foam (a rubbery polymer) represents elastic buckling; for an elastic-plastic foam (such as metallic foam), it represents plastic yield, and for an elastic-brittle foam (such as ceramic) it represents crushing. On the tension side, point 'A' represents the transition between cell wall bending and cell wall alignment. In elastomeric foam, the alignment occurs elastically, in elastic plastic foam it occurs plastically, and an elastic-brittle foam fractures at A.

Porous materials have a high ratio of surface area to volume. When porous materials are used in biomaterial applications, the demands upon the inertness and biocompatibility are likely to be greater than for a homogeneous material.

Porous materials, when used in implants, allow tissue ingrowth (Spector et al., 1988a, hI). The ingrowth is considered desirable in many contexts, since it allows a relatively permanent anchorage of the implant to the surrounding tissues. There are actually two composites to be considered in porous implants (1) the implant prior to ingrowth, in which the pores are filled with tissue fluid which is ordinarily of no mechanical consequence; and (2) the implant filled with tissue. In the case of the implant prior to ingrowth, it must be recognized that the stiffness and strength of the porous solid are much less than in the case of the solid from which it is derived.

Porous layers are used on bone compatible implants to encourage bony ingrowth [Galante et al., 1971; Ducheyne. 1984]. The pore size of a cellular solid has no influence on its stiffness or strength (though it does influence the toughness), however pore size can be of considerable biological importance. Specifically, in orthopedic implants with pores larger than about 150 μm , bony ingrowth into the pores occurs and this is useful to anchor the implant. This minimum pore size is on the order of the diameter of osteons in normal Haversian bone. It was found experimentally that pores $<75 \mu\text{m}$ in size did not permit the ingrowth of bone tissue. Moreover, it was difficult to maintain fully viable osteons within pores in the 75 to 150 μm size range. Representative structure of such a porous surface layer is shown in Figure 41.9. Porous coatings are also under study for application in anchoring the artificial roots of dental implants to the underlying jawbone. Porous hydroxyapatite has been studied for use in repairing large defects in bone [Meffert et al., 1985; Holmes et al., 1986]. Hydroxyapatite is the mineral constituent of bone, and it has the nominal composition $\text{Ca}_{10}(\text{P}_4)_6(\text{OH})_2$. Implanted hydroxyapatite is slowly resorbed by the body over several years and replaced by bone. Tricalcium phosphate is resorbed more quickly and has been considered as an implant constituent to speed healing.

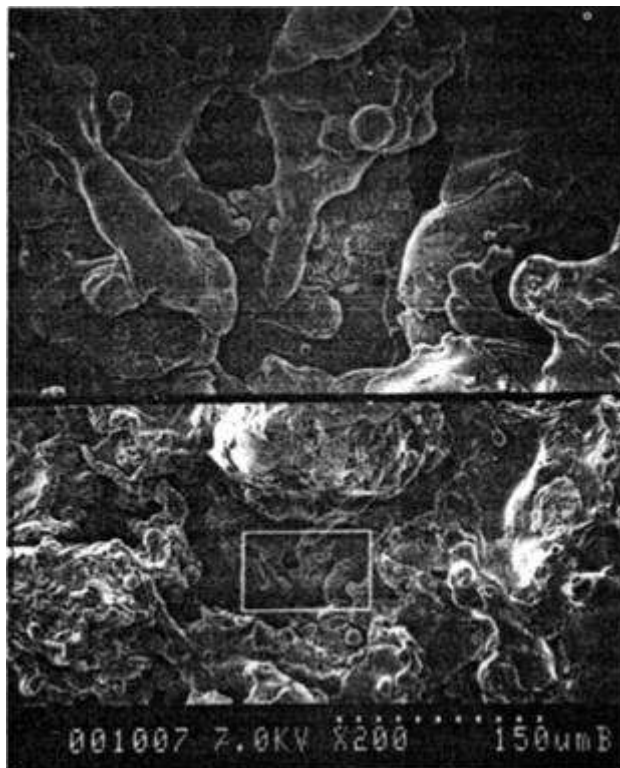


FIGURE 41.9 Irregular pore structure of porous coating in Ti5Al4V alloy for bony ingrowth. The top scanning electron microscopic picture is a 5 x magnification of the rectangular region of the bottom picture (200 x). (From Park, J.B. and Lakes, R.S. 1992. *Bionaterials*, Plenum, New York.)

When a porous material is implanted in bone, the pores become filled first with blood which clots, then with osteoprogenitor mesenchymal cells, then, after about 4 weeks, bony trabeculae. The ingrown bone then becomes remodeled in response to mechanical stress. The bony ingrowth process depends on a degree of mechanical stability in the early stages of healing. If too much motion occurs, the ingrown tissue will be collagenous scar tissue, not bone.

Porous materials used in soft tissue applications include polyurethane, polyamide, and polyester velours used in percutaneous devices. Porous reconstituted collagen has been used in artificial skin, and braided polypropylene has been used in artificial ligaments. As in the case of bone implants, the porosity encourages tissue ingrowth which anchors the device.

Blood vessel replacements are made with porous materials which encourage soft tissue to grow in, eventually forming a new lining, or neointima. The new lining consists of the patient's own cells. It is a natural nonthrombogenic surface resembling the lining of the original blood vessel. This is a further example of the biological role of porous materials as contrasted with the mechanical role.

Ingrowth of tissue into implant pores is not always desirable. For example, sponge (polyvinyl alcohol) implants used in early mammary augmentation surgery underwent ingrowth of fibrous tissue, and contracture and calcification of that tissue, resulting in hardened, calcified breasts. Current mammary implants make use of a balloon-like nonporous silicone rubber layer enclosing silicone oil or gel, or perhaps a saline solution in water. A porous layer of polyester felt or velour attached to the balloon is provided at the back surface of the implant so that limited tissue ingrowth will anchor it to the chest wall and prevent it from migrating.

Foams are also used externally to protect the human body from injury. Examples include knee pads, elbow pads, wrestling mats, and wheelchair cushions. Since these foams are only in contact with skin rather than any internal organs, they are not subject to rigorous biocompatibility requirements. They are therefore designed based on mechanical considerations. Foam used in sports equipment must have the correct compliance to limit impact force without bottoming out. Foam used in wheelchair cushions is intended to prevent pressure sores in people who suffer limited mobility. The properties of cushions are crucial in reducing illness and suffering in people who are confined to wheelchairs or hospital beds for long periods. Prolonged pressure on body parts can obstruct circulation in the capillaries. If this lasts too long it may cause a sore or ulcer called a pressure sore, also called a bed sore. In its most severe manifestation, a pressure sore can form a deep crater-like ulcer in which underlying muscle or bone is exposed [Dinsdale, 1974]. A variety of flexible cushion materials have been tried to minimize the incidence and severity of pressure sores [Garber, 1985]. Viscoelastic foam allows the cushion to progressively conform to the body shape. However, progressive densification of the foam due to creep results in a stiffer cushion which must be periodically replaced.

Porous materials are produced in a variety of ways. For example, in the case of bone compatible surfaces they are formed by sintering of beads or wires. Vascular and

soft tissue implants are produced by weaving or braiding fibers as well as by nonwoven “felting” methods. Protective foams for use outside the body are usually produced by use of a “blowing agent” which is a chemical which evolves gas during the polymerization of the foam. An interesting approach to producing micro-porous materials is the replication of structures found in biological materials: the *replarnineform* process [White et al., 1976]. The rationale is that the unique structure of communicating pores is thought to offer advantages in the induction of tissue ingrowth. The skeletal structure of coral or echinoderms (such as sea urchins) is replicated by a casting process in metals and polymers these have been tried in vascular and tracheal prostheses as well as in bone substitutes.

41.7 BIOCOMPATIBILITY

Carbon itself has been successfully used as a biomaterial Carbon based fibers used in composites are known to be inert in aqueous (even seawater) environments, however they do not have a track record in the biomaterials setting. *In vitro* studies by Kovacs [1993] disclose substantial electrochemical activity of carbon fiber composites in an aqueous environment. If such composites are placed near a metallic implant, galvanic corrosion is a possibility. Composite materials with a polymer matrix absorb water when placed in a hydrated environment such as the body. Moisture acts as a plasticizer of the matrix and shifts the glass transition temperature towards lower values [Delasi and Whiteside, 1978], hence a reduction in stiffness and an increase in mechanical damping. Water immersion of a graphite epoxy cross-ply composite [Gopalan et al., 1989] for 20 days reduced the strength by 13% and the stiffness by 9%. Moisture absorption by polymer constituents also causes swelling. Such swelling can be beneficial in dental composites since it offsets some of the shrinkage due to polymerization.

Flexible composite bone plates are effective in promoting healing I Jockish, 19921, but particulate debris from composite bone plates gives rise to a foreign body reaction similar to that caused by ultra high molecular weight polyethylene.

41.8 SUMMARY

Composite materials are a relatively recent addition to the class of materials used in structural applications. In the biomaterials field, the ingress of composites has been even more recent. In view of their potential for high performance, composite materials are likely to find increasing use as biomaterials.