BIOCERAMICS
This chapter gives an overview of and re-examines key issues which concern both the processing and applications of ceramics as biomaterials.

Biomaterial, by definition, is “a non-drug substance suitable for inclusion in systems which augment or replace the function of bodily tissues or organs”. Surgeons and engineers identified materials and design problems that resulted in premature loss of implant function - such as through mechanical failure, corrosion, or inadequate biocompatibility of the component. Key factors in a biomaterial usage are its biocompatibility, biofunctionality, as well as availability (to a lesser extent).

Ceramics are ideal candidates with respect to all the above criteria, except for their brittle behavior.

In this chapter, we shall revisit the presently available and currently investigated bioceramics, their preparation methods, properties, and their applications in comparison to biogenic, metallic and polymeric biomaterials.
General Concepts in Bioceramics

When a man-made material is placed within the human body, tissue reacts towards the implant in a variety of ways depending on the material type. The mechanism of tissue interaction (if any) depends on the tissue response to the implant surface. In general, tissue responses to a biomaterial may be described or classified as bioinert, bioresorbable, or bioactive.

1. **Bioinert** refers to any material that once placed within the human body has minimal interaction with its surrounding tissue. Examples of bioinert materials are stainless steel, titanium, alumina, partially stabilized zirconia, and ultra high molecular weight polyethylene. Generally a fibrous capsule forms around a bioinert implant. Hence the implant’s biofunctionality relies on tissue integration through the capsule (Figure 6-1 (a)).

2. **Bioactive** refers to a material, which upon being placed within the human body interacts with the surrounding bone and in some cases, even with the soft tissue. This occurs through a time-dependent kinetic modification of the surface, Prime examples of bioactive materials are synthetic hydroxyapatite \([\text{Ca}_{10}(\text{PO}_4)_6(\text{OH})_2]\), glass-ceramic A-W , and Bioglass (Figure 6-1(b,c)).
3. **Bioresorbable** refers to a material that upon placement within the human body starts to dissolve (resorbed) and is slowly replaced by advancing tissue (such as bone). Common examples of bioresorbable materials are tricalcium phosphate \([\text{Ca}_3(\text{P}O_4)_2]\) and polylactic-polyglycolic acid copolymers. Calcium oxide, calcium carbonate (coral), and gypsum are other common materials that have been used during the last three decades (Figure 6-1(d)).

**Figure 6-1** Classification of bioceramics according to their responses at the bone-implant interface: (a) bioinert alumina dental implant; (b) bioactive hydroxyapatite \([\text{Ca}_{10}(\text{P}O_4)_2(\text{OH})_2]\) coating on a metallic dental implant; (c) surface active bioglass; (d) bioresorbable tricalcium phosphate implant \([\text{Ca}_3(\text{P}O_4)_2]\)
Bioceramics and Production Methods

In clinical practice, four basic classes of material are used for biomedical implants and devices. They are bioceramics, bio-metals (metals that could be used as biomaterials), bio-polymers, and composites. Each material class has combinations of properties determined by material composition and production methods used, while each set of properties has its own benefits and limitations. Recently a fifth group of inorganic-organic composites is introduced. They are appropriately named as hybrids since some natural materials are used. Ceramics are the hardest of solids. Table 6-1 gives some of the mechanical properties of natural and synthetic biomaterials.

Metals are intrinsically soft and ductile due to their appropriate slip systems which allow yielding and metallic bonding, and where electrons are free to move around the ion cores. Metals can be shaped easily by machining or by casting from molten state without any major difficulties.
Table 6-1 Mechanical properties of biomaterials (modified from Lutton and Ben-Nissan)

<table>
<thead>
<tr>
<th></th>
<th>Young’s Modulus GPa</th>
<th>Compressive Strength MPa</th>
<th>Tensile Strength MPa</th>
<th>Density g/cm³</th>
<th>Fracture Toughness MPam ½</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>METALS</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Titanium (Ti–6Al–4V)</td>
<td>114</td>
<td>450–1850</td>
<td>900–1172</td>
<td>4.43</td>
<td>44–66</td>
</tr>
<tr>
<td>Cr–Co–Mo</td>
<td>210</td>
<td>480–600</td>
<td>400–1030</td>
<td>8.3</td>
<td>120–160</td>
</tr>
<tr>
<td>Stainless Steel (316L)</td>
<td>193</td>
<td>-</td>
<td>515–620</td>
<td>8.0</td>
<td>20–95</td>
</tr>
<tr>
<td><strong>CERAMICS</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Alumina</td>
<td>420</td>
<td>4400</td>
<td>282–551</td>
<td>3.98</td>
<td>3–5.4</td>
</tr>
<tr>
<td>Zirconia (TZP)</td>
<td>210</td>
<td>1990</td>
<td>800–1500</td>
<td>5.74–6.08</td>
<td>6.4–10.9</td>
</tr>
<tr>
<td>Silicon Nitride (HPSN)</td>
<td>304</td>
<td>3700</td>
<td>700–1000</td>
<td>3.3</td>
<td>3.7–5.5</td>
</tr>
<tr>
<td>Hydroxyapatite (3% porosity)</td>
<td>7–13</td>
<td>350–450</td>
<td>38–48</td>
<td>-</td>
<td>3.05–3.15</td>
</tr>
<tr>
<td><strong>HUMAN TISSUE</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>Cortical Bone</td>
<td>3.8–11.7</td>
<td>88–164</td>
<td>82–114</td>
<td>1.7–2.0</td>
<td>2–12</td>
</tr>
<tr>
<td>Cancellous Bone</td>
<td>0.2–0.5</td>
<td>23</td>
<td>10–20</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Cartilage</td>
<td>0.002–0.01</td>
<td>-</td>
<td>5–25</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td><strong>OTHERS</strong></td>
<td></td>
<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Bone Cement (PMMA)</td>
<td>2.24–3.25</td>
<td>80</td>
<td>48–72</td>
<td>-</td>
<td>1.19</td>
</tr>
<tr>
<td>UHMWHD Polyethylene</td>
<td>0.69</td>
<td>20</td>
<td>38–48</td>
<td>0.94</td>
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</tr>
</tbody>
</table>
However, most ceramics are intrinsically hard due to their ionic, covalent, and/or mixed bonding which presents an enormous lattice resistance to the motion of dislocations. Hence, ceramics cannot be shaped by melting and casting. These properties of hardness and strength of ceramics are exploited in areas where wear resistance is required. In general, crack-tip plasticity gives metals their high toughness. Energy is absorbed in the plastic zone, generating a tortuous path that makes crack propagation much more difficult. Although some plasticity can occur at the tip of a crack in a ceramic too, the energy. Due to their strong bonding, ceramics have very high melting — or more appropriately, dissociation — temperatures. Hence, ceramics can be produced only through high temperature sintering. Sintering is a process of densification where powders are heated up usually to two-thirds of their melting temperature, and with the aid of a driving force such as diffusion, they consolidate (Figure 6-2).

**Figure 6—2 SEM of a sintered bioceramic**
Contact Angle Methods

Sintering rate is controlled by diffusion. However, large driving force shortens the sintering time and increases the final density. Various methods used in producing advanced ceramics are given in Table 6-2. Hot isostatic pressing is most commonly used.

Table 6-2 Some common bioceramic production methods

<table>
<thead>
<tr>
<th>PRESSING</th>
<th>CASTING</th>
</tr>
</thead>
<tbody>
<tr>
<td>Uniaxial</td>
<td>Slip Casting</td>
</tr>
<tr>
<td>Cold Isostatic Pressing (CIP)</td>
<td>Thixotropic Casting</td>
</tr>
<tr>
<td>Hot Pressing (HP)</td>
<td>Gel Casting</td>
</tr>
<tr>
<td>Hot Isostatic Pressing (HIP)</td>
<td>Soluble Mold Casting</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>PLASTIC FORMING</th>
<th>COATINGS</th>
</tr>
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<tbody>
<tr>
<td>Extrusion</td>
<td>Sol-Gel Coating</td>
</tr>
<tr>
<td>Injection Molding</td>
<td>Electrodeposition</td>
</tr>
<tr>
<td>Compression Molding</td>
<td>Flame / Plasma Spray</td>
</tr>
<tr>
<td></td>
<td>SBF</td>
</tr>
<tr>
<td></td>
<td>PVD / CVD</td>
</tr>
</tbody>
</table>
Bioinert Ceramics in Articulation

Ceramics are considered hard, brittle materials with relatively poor tensile properties. Other characteristics include excellent compressive strength, high resistance to wear, and favorably low frictional property in articulation. The low frictional property is enhanced by the fact that ceramics are hydrophilic with good wettability (Figure 6-4).

Bioceramics used singularly or with additional natural, organic, or polymeric materials are amongst the most promising of all biomaterials for hard and soft tissue applications. Ceramics that are used in implantation and clinical purposes include aluminum oxide (alumina), partially stabilized zirconia (PSZ) (both yittria [Y-TZP] and magnesia stabilized [Mg-PSZ]), Bioglass®, glass-ceramics, calcium phosphates (hydroxyapatite and p-tricalcium phosphate), and crystalline or glassy forms of carbon and its compounds.

Figure 6-4 Hydrophilichydrophobic behavior shown by wetting angles of different orthopedic biomaterials: (a) Polyethylene (RCH-1000); (b) FeCrNiMo, AISI-3 16L stainless steel; (c) CoCrMo alloy (Protasul-2); (d) Alumina ceramic (Biolox)
1- Alumina Ceramics

High-purity alumina bioceramics have been developed as an alternative to surgical metal alloys for total hip prostheses and tooth implants. The high hardness, low friction coefficient and excellent corrosion resistance of alumina offers a very low wear rate as the articulating surface in orthopedic applications. Medical grade alumina has a very low concentration of sintering additives (< 0.5 wt%), relatively small grain size (< 7 pm) and a narrow grain-size distribution. Such a microstructure is capable of inhibiting static fatigue and slowing crack growth while the ceramic is under load. The average grain size of current medical grade aluminas is 1.4 µm, and surface finish is usually controlled to a roughness of less than 0.02 pm. However, unless its surface is modified or used directly in articulating areas, alumina has a fundamental limitation as an implant material in that, like other “inert” biomaterials, a non-adherent fibrous membrane may develop at the interface. Currently alumina is used for orthopedic and dental implants, and which can be polished to a high surface finish and high hardness. The mechanical behavior of alumina ceramics At 112 MPa stress level, the probability for medical grade alumina to survive 50 years is 99.9 percent.
2- Partially Stabilized Zirconia (PS Z)

Compared to alumina, PSZ has high Weibull modulus - hence better reliability, higher flexural strength and fracture toughness, lower Young’s modulus (Table 6-1), and the ability to be polished to a superior surface finish [20,2 11. The higher fracture toughness is of particular importance in femoral heads due to the tensile stresses induced by the taper fit onto the femoral stem. Partially stabilized zirconia femoral heads make up about 25 percent of the total number of operations per year in Europe. Although not quite as hard as alumina, PSZ still possesses excellent wear resistance and has been used for similar orthopedic applications as alumina.

Clarke et al. (2000) conducted a study on the articulation of the femoral head in total hip replacement (THR), using hip simulators with alpha-calf serum as a lubricant. In this study, wear rates of alumina/alumina, zirconia/alumina, and zirconia/zirconia couples were investigated. Results revealed that wear rates using zirconia/zirconia exhibit a mild run-in phase, as compared to a more evident run-in phase for alumina/alumina articulation.
Following the run-in phase, weight changes of all couple samples were observed. Zirconia/zirconia wear offered little observable weight change. Alumina/alumina wear (although very low) revealed a steady weight loss trend after the run-in phase. In the case of zirconia/alumina (where the head was made of zirconia and the liner made of alumina), the zirconia head showed little weight loss but the alumina liner revealed a typical run-in phase followed by a steady state weight loss.

Zirconia ceramic implants have had a somewhat controversial history. Bologna, Italy, reported that the wear of zirconia-on-zirconia is 5,000 times worse than that of alumina-on-alumina.

It has been shown that when evaluating total hip joint replacements (THR), the lubricant used in these laboratory evaluations significantly influences the wear results. Various studies have been conducted successfully when alumina-on-alumina bearings were lubricated using water, saline, and serum. However, zirconia-on-zirconia tests in water have consistently shown catastrophic results while those in serum have demonstrated good results. Widely available is a number of clinical studies which showed excellent short-term results when zirconia balls were combined with alumina cups. Given these somewhat contradictory results between laboratory and clinical studies, Clarke et al studied the wear of zirconia-on-zirconia bearings in both water and serum.
The water lubrication test resulted in a wear about 10,000 times greater than with serum lubrication. The wear of both zirconia femoral head and cup in water lubrication showed a high weight loss of 28 mg after only 6,100 cycles. In contrast, the zirconia wear with serum lubrication had a weight loss of only 0.74 mg after 20 million cycles. This difference between the two lubricants was also distinct in the micro-wear of the ball surfaces. With serum, there were still some original machine tracks to be seen; with water there was total surface deterioration (Figure 6-5).

**Figure 6-5** SEM images of zirconia (TZP) femoral head deterioration in hip simulator studies showing comparative wear surfaces: (a) zirconia with serum lubrication; (b) zirconia with water lubrication.
Therefore it was proposed that the simulator with water lubrication created extensive zirconia transformation. However under serum lubricant conditions, this transformation did not occur. Based on this difference in lubricant performance, Clarke et al postulated that the serum proteins either formed an effective solid lubricant film between the zirconia surfaces. Their study revealed the dramatic effect that serum proteins had at the bearing surface, thus greatly reducing the wear rate of zirconia.

3- New Modified Zirconia Implants

Recently, new degradation-free zirconia-alumina composites have been reported: TZP/alumina composite (80%TZP of [90mol%ZrO₂-6mol%Y₂O₃-4mol %Nb₂O₅ composition] and 20%Al₂O₃). Another potential composite – which comprises 70vol%TZP (stabilized with 10mol%CeO₂) and 30vol%Al₂O₃ and 0.05 mol 5TiO₂ - is currently being investigated in Japan.
Bioresorbable and Bioactive Ceramics

1-Calcium Phosphates for Bone Replacement Applications

In 1926, apatite was identified as the only recognizable mineral phase. Numerous other studies have suggested the existence of other mineral phases in the bone. These include amorphous calcium phosphate (ACP), brushite, and octacalcium phosphate (OCP). The presence of substantial amounts of ACP or brushite has not been yet experimentally proved, and nuclear magnetic resonance studies support the conclusion that bone is composed essentially of carbonate-substituted hydroxy apatite (CHAP).

It was not until the 1970s that synthetic hydroxyapatite \([\text{Ca}_{10}(\text{PO}_4)_2(\text{OH})_2]\) was accepted as a potential biomaterial. Synthetic hydroxyapatite forms a strong chemical bond with bone in vivo, and it is also able to remain stable under the harsh conditions encountered in the physiologic environment. Bone-like carbonated apatite is a non-stoichiometric mineral found in hard tissues of all mammals. Synthetic hydroxyapatite \([\text{Ca}_{10}(\text{PO}_4)_6(\text{OH})_2]\) has been an attractive material for chromatographic separation catalysis, ion exchange, bone and tooth implants.
Albee and Morrison proposed the use of calcium phosphate ceramics for biomedical applications, after observing accelerated bone growth when tricalcium phosphate was injected into bone defects. Pure tricalcium phosphate (TCP) Ca$_3$(PO$_4$)$_{2}$ is more soluble in physiological environment than other phosphate ceramics (bioresorbable). Consequently, it can be used in situations where accelerated bone growth is desired. p-tricalcium phosphates have been used successfully as fillers for bone defects to stimulate new bone formation. Their study also showed that after a '12-month period, P-tricalcium phosphate was totally absorbed. It was targeted that these materials will be used to fill voids in bone structure since they will dissolve over a period of time. It was also targeted that while the dissolution takes place, bone regrowth or advancement takes place at a similar rate.

Dissolution rates of some materials under simulated physiological conditions have been investigated:

Tetracalcium phosphate > P-Tricalcium Phosphate > Hydroxyapatite

It has been stated that hydroxyapatite is “scarcely resorbable”, hence justifying the use of hydroxyapatite for osseous implants.
2- Simulated Body Fluid (SBF)

Introduced by Kokubo and co-workers [43-441, the simulated body fluid (SBF) is welcomed as a promising method in bioceramics production. This synthetic body fluid is highly supersaturated in calcium and phosphate - in relationship to apatite formation - even under normal conditions. Therefore if a material has a functional group effective for apatite nucleation on its surface, apatite can be formed spontaneously. For an artificial material to bond to the living bone, it is widely accepted that a bone-like apatite layer must form on its surface. Formation of the bone-like apatite layer on bioactive materials can be produced in a simulated body fluid (SBF) which has ion concentrations almost equal to those of the human blood plasma. SBF is a metastable solution.
3- Coralline Apatites

Coralline apatites can be derived from the sea coral. Coral is composed of calcium carbonate in the form of aragonite. As coral is a naturally occurring structure, it has optimal strength and structural characteristics. The pore structure of coralline calcium phosphate produced by certain species is similar to human cancellous bone, making it a suitable material for bone graft applications (Figure 6-7). Coral and converted coralline hydroxyapatite have been used as bone grafts and orbital implants since the 1980s. This is because the porous nature of the structure allows ingrowth of blood vessels to supply blood for bone, which eventually infiltrates the implant. Size and interconnectivity of pores are of utmost importance when hard and soft tissue ingrowth is required. The hydrothermal method was first used by Roy and Linnehan [46] in 1974 to form hydroxyapatite directly from corals.
Figure 6-7 Comparison of the Australian coral: (a) in original state; and (b) after hydrothermal conversion.

The resulting material is known as coralline hydroxyapatite, be it in the porous coralline structure or in powdered form. Alternatively, aragonite can be converted to carbonate hydroxyapatite using microwave processing technique.
4- Calcium Phosphate Coatings

Due to its unfavorable mechanical properties, it has been accepted that porous hydroxyapatite cannot be used under load bearing conditions. Instead hydroxyapatite has been used as thin film coatings on metallic alloys. Of the metallic alloys investigated, titanium-based alloys have shown to be the preferred material for thin film coatings. Being lightweight and with high strength-to-weight ratios, titanium alloys possess good mechanical strength and fatigue resistance under load bearing conditions. Of the coating techniques utilized, thermal spraying (plasma, and to a lower extend flame spraying) tends to be the most commonly used and analyzed.

While striving towards this target, thermally sprayed coatings are being improved continuously by using different compositions and post-heat treatments which convert amorphous phases to crystalline calcium phosphates. Currently plasma coating of macrotextured orthopedic implants is used commercially, and other techniques involving less soluble fluorapatite compositions are also being investigated. Techniques that are capable of producing thin coatings include pulsed-laser deposition and sputtering which, like thermal spraying, involve high-temperature processing.
Other techniques such as electrodeposition and sol-gel use lower temperatures and avoid the challenge associated with the structural instability of hydroxyapatite at elevated temperatures. The advantages of sol-gel technique are numerous, namely: It results in a stoichiometric, homogeneous, and pure coating due to mixing on a molecular scale; Firing temperatures are reduced due to small particle sizes with high surface areas; It is able to produce uniform fine-grained structures (Figure 6-8); It uses different chemical routes (alkoxide or aqueous based); It is easy to be applied to complex shapes using a range of coating techniques such as dip, spin, and spray coating; Due to its lower processing temperature, it avoids the phase transition (\(-156^\circ\text{K}\)) observed in titanium-based alloys used for biomedical devices.

**Figure 6-8** SEM and AFM images of sol-gel (alkoxide) derived hydroxyapatite coatings
5- Synthetic Bone Graft Ceramics

Bone grafting is currently used in orthopedic and maxillofacial surgeries for these treatments: diaphyseal defects bridging, non-union, metaphyseal defects filling, and mandibular reconstruction. Autogenous bone graft has the following characteristics:

**Osteogenic** (that is, able to form bone due to living cells such as osteocytes or osteoblasts);

**Osteoconductive** (that is, no capacity to induce or form bone but provides an inert scaffold upon which osseous tissue can regenerate bone);

**Osteoinductive** (that is, able to stimulate cells to undergo phenotypic conversion to osteoprogenitor cell types capable of bone formation).

There are no substitutes for autogenous bone; there are, however, synthetic alternatives. Allografts have been used as an alternative. However this alternative offers low or no osteogenicity, increases immunogenicity, and resorbs more rapidly than autogenous bone.

Allograft bone is a useful material for patients who require bone grafting of a non-union but have inadequate autograft bone. Bulk allografts can be utilized to treat segmental bone defects. Calcium sulfate (plaster of Paris) or its composites are one of the oldest osteoconductive materials available. They have been used to fill bony defects.
Demineralized bone matrix (DBM) The members of this group are responsible for the morphogenetic events involved in tissues and organs development. Urist also isolated a protein from the bone matrix, which was termed as bone morphogenic protein (BMP) Bovine collagen mixed with hydroxyapatite is marketed as a bone graft substitute, which can be combined with bone marrow aspirated from the fracture site.

6- Bioglasses and Glass- Ceramics
Since Hench and Wilson discovered the bioglasses which bond to living tissue (Bioglass), various kinds of bioactive glasses and glass-ceramics with different functions - such as high mechanical strength, high machinability, and fast setting ability - have been developed. The glasses that have been investigated for implantation are primarily based on silica (SiO₂) which may contain small amounts of other crystalline phases. Hench and Vrouwenvelder et al. suggested that 45S5 Bioglass has greater osteoblastic activity as compared to hydroxyapatite. The reasoning behind this was due to a rapid exchange of alkali ions at the surface, which then leads to the formation of a silica-rich layer over a period of time.
Bioglasses has been used successfully in clinical applications as artificial middle ear bone and in alveolar ridge maintenance implants. Bioglass, when subjected to heat treatment, will result in reduced alkaline oxide content and precipitated crystalline apatite in the glass.

Nano-bioceramics, Composites and Hybrids

1- Nanoapatite-polymer Fiber Composites

Bone is a composite in which nanosized apatite platelets are deposited on organic collagen fibers. If three-dimensional synthetic organic fibers can be fabricated into a composite structure and then modified with functional groups effective for apatite nucleation, morphologically similar bone structures could perhaps be prepared in such a manner. The resultant composite could be expected to exhibit bioactivity as well as mechanical properties analogous to those of the living bone.
2- Bioceramics in In Situ Radiotherapy and Hyperthermia

One of the most common approaches in cancer treatment is the removal of the diseased parts. Unfortunately recovery of or return to full function is seldom achieved. Non-invasive treatment techniques — where only the cancer cells are Destroyed In 1987, microspheres of 17Y$_2$O$_3$-19Al$_2$O$_3$-64SiO$_2$ (mol%) glass, 20-30 um in diameter were shown to be effective for in situ radiotherapy of liver cancer. The glass microspheres are usually injected into diseased liver through the hepatic artery. Once entrapped in small blood vessels, they block the blood supply to the cancer and directly irradiate the cancer with P-rays. Since the (3-ray transmission is only 2.5 mm in diameter on living tissue, and since the glass microspheres have high chemical durability the surrounding normal tissue is hardly damaged by the p-rays.

Figure 6-9 SEM image of Y2O3 microspheres for radiotherapy applications
Oxygen is known to be poorly supplied to cancer cells to produce lactic acid. Hence cancer cells can be destroyed at around 43°C, whereas normal living cells are kept alive even at 48°C. If ferri- or ferromagnetic materials are implanted around cancer cells and placed under an alternating magnetic field, it is expected that through magnetic hysteresis loss of the ferri-or ferromagnetic materials, cancer cells locally heated can be destroyed.

3- Bone Cement Composites

During the last five years, bone cement materials have grown in popularity and are slated to be the osteoconductive substitutes for bone graft. They are prepared like acrylic cements, where a range of powders such as monocalcium phosphate, tricalcium phosphate, and calcium carbonate is mixed in a sodium phosphate solution. These composites are currently used in orthopedics for fractures management. It has been suggested that these materials could improve the compressive strength of vertebral bodies in osteoporosis. Injection of calcium phosphate cement has been shown to be feasible and effective: the cement indeed improved their compressive strength. Attempts have been made to prepare hydroxyapatitekeramic composites by adding various ceramic reinforcements: metal fibers, Si$_3$N$_4$ or hydroxyapatite whiskers, Al$_2$O$_3$ platelets, and ZrO$_2$ particles.
7- Design with Bioceramics

Five important factors should be considered in any implantation work:

1- Production method of the material used;

2- Biocompatibility, tissue-implant interface reactions, and choice of the material used;

3- Applied stresses and biomechanics of the joint;

4- Physical well-being and age of the patient; and

5- Surgical technique.

Various criteria exist when selecting a biomaterial for a particular biomedical application. After biocompatibility, mechanical properties are the second decisive criterion when it comes to selecting a material for a particular orthopedic or maxillofacial application. Selected mechanical properties of a range of synthetic and natural biomaterials are shown in Table 6-1.