Orthopedic Implants

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Introduction

People suffering from accidents and joint diseases such as osteoarthritis, rheumatoid arthritis, and post-traumatic arthritis may need surgery requiring implants such as total hip and knee replacements. Orthopedic implants also include temporary fracture fixation devices and components such as plates, screws pins, wires, and nails. Since orthopedic implants must function under different working conditions *in vivo*, a good understanding of the fundamental requirements of orthopedic materials and subsequent biological response is crucial to the design and optimization of implants under physiological conditions in the human body. Metallic alloys, ceramics, and polymers are commonly used in orthopedic implants. These materials possess different physical, chemical, and biological properties that cater to specific applications. For instance, owing to the good mechanical properties, metallic alloys are widely used in load-bearing joint prostheses and devices to fix bone fracture, whereas ceramic materials with excellent wear resistance and bioactivity are often used as articulating components or bioactive coatings on implants. In general, polymers serve as a cushion between joints to reduce friction and fixation devices. The demand for better orthopedic materials has spurred significant advance in this field leading to the design and production of orthopedic implants with better performance and novel properties.

Classification of Orthopedic Implants

Permanent Orthopedic Implants

Different types of total joints including the hip, knee, ankle, shoulder, elbow, wrist, and finger joints are used as replacements clinically (Park and Lakes, 2007). These permanent orthopedic implants are expected to serve in the human body throughout the life span of the patients, and metals, ceramics, and polymers are commonly used. In particular, hip and knee joint prostheses have experienced rapid development and clinical acceptance in recent years. However, the delicate articulation and complicated load transfer make such prostheses very difficult to design. A typical total hip replacement is composed of a stem, femoral head, liner, and acetabular cup (Fig. 1A). The femoral head is mounted on the stem and placed in the liner that is fixed with a cup. The liner protects the cup from wear damage, and the cup provides protection for the host bone. The development of the knee joint replacement has

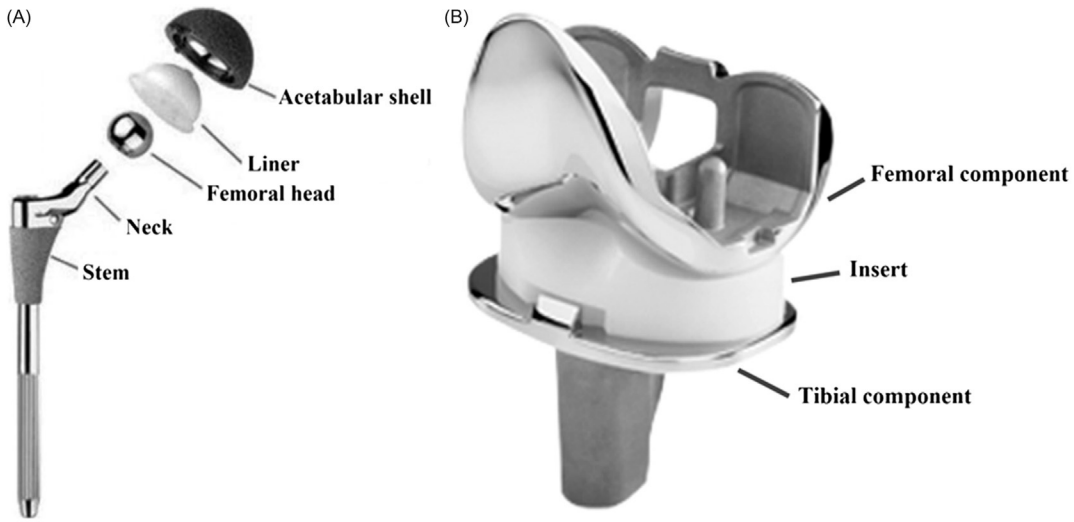

Fig. 1 (A) Total hip (Geetha et al., 2009) and (B) knee (http://www.zimmer.com/medical-professionals/products/knee/nexgen-lps-flex-mobile-lps-mobile.html) replacements.

been slower than that of hip joint prostheses because the geometry and biomechanics of knee movement are more complicated compared with that of the hip. A total knee replacement typically consists of a femoral component, insert, tibial tray, and patellar component (Fig. 1B). The insert between the femoral and tibial components is to create a smooth gliding surface.

Temporary Orthopedic Implants

Another type of orthopedic implants is temporary ones that are needed to fix broken or fractured bones during the healing process. Temporary orthopedic implants including plates, screws, pins, wires, and intramedullary nails (Park and Lakes, 2007) (Fig. 2) are supposed to serve for a relatively short time just long enough to let bones heal. Plates with different types and sizes containing holes

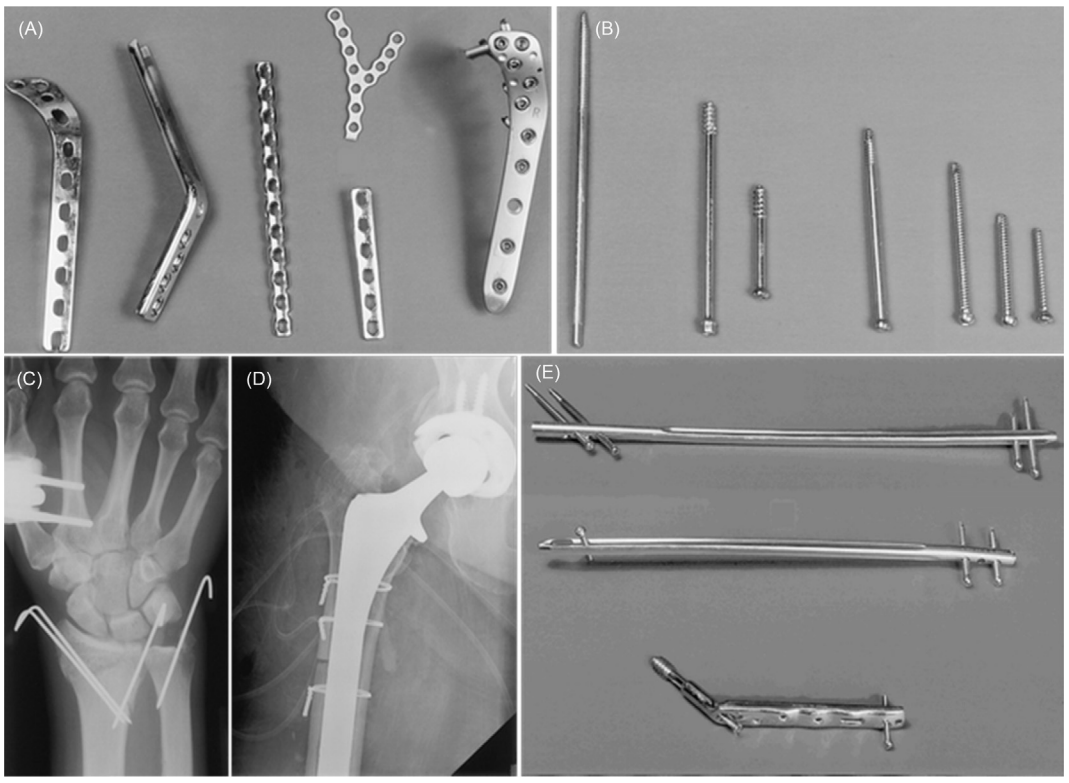

Fig. 2 (A) Plates, (B) screws, (C) pins, (D) wires, and (E) intramedullary nails used in fracture fixation (Taljanovic et al., 2003).

for screws and pins are used to fix plates to bones, so that bone fragments can be compressed together during the healing process of load-bearing bones. Screws are commonly used together with other devices especially plates to fix the associated devices to bones and also as stand-alone components to fix fractured fragments. Pins can be used as adjunctive fixation devices along with other devices in complex fracture of bones in order to withstand a large loading force or used alone to fix bones under a relatively weak force. As simple but versatile implants, wires are commonly employed to hold bone fragments together and also in reattachment of the greater trochanter in hip replacements or long oblique/spiral fracture of long bones. An intramedullary nail is a surgical rod forced into a long bone of the extremities through the medullary canal to act as an immobilization device to hold the two ends of the fractured long bone in position. To anchor the nails in place inside a piece of bone, other implants such as pins or screws are generally inserted into predrilled holes in the rods.

One major concern of bone healing fixed with rigid plates is bone weakening due to mainly the stress-shielding effect that may result in refracture after plate removal. The stress-shielding effect is also an important concern for permanent prosthetic implants. The osteogenic (bone generating) and osteoclastic (bone removing) processes are known to be normal activities in the human bones. As described by Wolff's law (Ruff et al., 2006), if a larger load is applied to the bone over a period of time, the osteogenic process is more effective and produces better support for the load. Conversely, a reduced load to the bone induces loss of bone mass. Nowadays, biodegradable materials such as metallic alloys (Zheng et al., 2014) and polymers (Middleton and Tipton, 2000) with similar Young's modulus to that of the bone have attracted a lot of attention as potential bone fixation devices. Free from the shielding effect, bone healing is promoted, and natural degradation of the implants enables gradual reduction of strength of the implant and final absorption by the human body. Thus, a follow-up surgery is not needed to remove the fixation devices after healing, thereby alleviating patient suffering and reducing health-care cost. However, this approach is still in the infancy stage due to issues such as rapid degradation, and more work is needed to make the materials clinically acceptable.

Consideration of Orthopedic Implant Materials

The aim of orthopedic implants is to restore the structural integrity and functionality of damaged joints and bones. To produce safe implants with a long lifetime and without inducing rejection, the biomaterials should possess desirable characteristics including mechanical properties, wear resistance, corrosion resistance, biocompatibility, and sometimes osseointegration (Davis, 2003; Chen and Thouas, 2014).

Mechanical Properties

The mechanical requirements for orthopedic implant materials are related to the intended working conditions and specific applications. The Young's modulus, yield strength, ultimate tensile strength, fracture toughness, and elongation at break are five important mechanical properties. Other special mechanical properties such as resistance against fatigue can be correlated to and predicted from these five properties. If the implant materials have a larger Young's modulus than the human bone, stress transfer to the adjacent bone will be prevented leading to bone resorption and implant loosening. Therefore, the orthopedic implant materials should preferably have a Young's modulus similar to that of the human bones. The fatigue strength is another important mechanical requirement for orthopedic implants from the perspective of cyclic load siting, especially joint replacements of the hips, knees, and ankles. With regard to total hip replacements, the loading stress level can be several times higher than that of the patient's body weight. Total hip replacements are expected to function for 20 years translating into maintaining the physical integrity after loading to $\sim$300 MPa for $\sim 1 \times 10^7$ times (Milne et al., 2003). In the human body, fatigue can occur along with stress and corrosion, which is known as fretting fatigue or fretting corrosion fatigue. Fatigue fracture becomes the major cause of premature failure of biomedical implants, and hence, materials with excellent fatigue strength are preferred for orthopedic implants. Generally, materials with high ultimate tensile strength, Young's modulus, and yield strength tend to have excellent fatigue resistance. Furthermore, the fracture toughness of implant materials, especially brittle ceramics, should be assured to resist crack propagation under a load, and a certain amount of elongation is also required to improve the manufacturability.

Wear Resistance

Wear is an inevitable problem for joint replacements where there is relative motion between the two surfaces under loading resulting in successive release of wear debris into the areas surrounding an implant. Polymer, metal, or ceramic particles with a certain size can cause adverse biological reactions in the human body. The most prevalent wear debris are polyethylene particles generated primarily from the articulating interface of the prostheses since the softer of the two bearing materials experiences more rapid wear. The wear particles enter the periprosthetic tissue and are phagocytosed by macrophages. The macrophages then release cytokines and other mediators of inflammation leading to development of inflamed granulomatous tissues surrounding the bone. Eventually, osteoclasts are activated to resorb the bone resulting in osteolysis (Ingham and Fisher, 2005). As a consequence of the cellular response to particulate wear debris from implant materials, osteolysis is a major cause of long-term failure of prostheses (Harris, 2001). Even though the overall density of particles may be small, local accumulation of wear debris may produce local osteolysis. The occurrence of osteolysis increases with the wear rate, and it has been shown that osteolysis is rarely observed below

the wear threshold (Dumbleton et al., 2002). Therefore, high wear resistance is desired for joint replacements and fixation devices to avoid loosening and premature failure of implants. The wear resistance depends on the yield strength, tensile strength, and Young's modulus. Hardness is also an important mechanical property for orthopedic implants such as the femoral head and materials with large values of these three attributes generally exhibit high hardness and good wear resistance.

Corrosion Resistance

Corrosion is of great concern particularly for metallic materials *in vivo*. The human body presents an aggressive environment because human body fluids consist of various cations such as sodium, potassium, calcium, and magnesium (Mg) ions; anions like chloride (Cl) ions, phosphate, and bicarbonate ions; amino acids; proteins; and dissolved oxygen (O_2). Electrochemical corrosion tends to occur on the metallic implant surfaces in the corrosive physical environment over time. As commonly observed from metals in the ambient environment, corrosion is accelerated by aqueous ions especially aggressive Cl^-. Proteins can promote corrosion by binding to metal ions to transport them away from the implant surface or absorbing on the implant surface to reduce diffusion of O_2. Although the pH of human fluids is normally maintained at 7.2–7.4, this value may fall to 3 if infection occurs after surgery or injury. Acidification in the physiological environment can accelerate corrosion of the implant materials (Manivasagam et al., 2010). The released degradation products such as metallic ions particularly chromium (Cr), cobalt (Co), and nickel (Ni) ions in excess amounts can be toxic to the human body (Hallab et al., 2005). Hence, the corrosion resistance is an important property that must be considered for metallic biomaterials. For orthopedic implants, especially permanent prosthetic implants, the degradation process may reduce the structural integrity of the implants, resulting in premature failure and needing a revision surgery. Thus, development of implant materials with excellent corrosion resistance in the physiological environment is essential to achieving the expected longevity of implants in the human body. Materials with controllable degradation rates have great potential as temporary orthopedic implants because they are supposed to function during the healing stage and then not needed after complete healing.

Biocompatibility

As the most important aspect of implant materials, biocompatibility is defined by the US Food and Drug Administration as the effect that the materials induce no measurable harm to the host. It is critical that implant materials are nontoxic and do not produce harmful effects to the host because of the intimate contact with human tissues. Metal ions released from metallic implants to adjacent tissues may activate the immune system and induce hypersensitivity response, which may eventually contribute to implant failure (Hallab et al., 2005). Hence, nontoxic elements should be selected as alloying elements during the design and fabrication of new metallic implants. When trace elements naturally existing in the human body are included, it is important to control the released amounts to reasonably low levels during the life span of the implants by enhancing the corrosion resistance. Meanwhile, large amounts of particulates generated by wear, corrosion, or a combination of these two processes from polymer, metal, or ceramic implants are foreign bodies engulfed by the local immune system and induce subsequent inflammation to invade the bone–implant interface. This can lead to aseptic osteolysis and eventual implant loosening, thereby hampering the normal functions of orthopedic implants (Hallab and Jacobs, 2009). Therefore, the selected alloying elements in the implant materials and excellent resistance to corrosion and wear in the physiological environment are crucial to good biocompatibility.

Osseointegration

Osseointegration is defined as the direct structural and functional connection between the living bone and surface of a load-bearing implant (Branemark, 1983). The ability of an implant to bond with the surrounding host bone is another fundamental requirement for permanent orthopedic implants. Insufficient osseointegration can lead to the formation of fibrous tissues and ensuing loosening of the prostheses. Factors such as the design, chemical composition, surface roughness, and surface chemistry of the implants and loading conditions are important to good osseointegration of implants (Mavrogenis et al., 2009). Hence, a surface integrating well with the adjacent bone is vital to the prevention of osteolysis. Nevertheless, bone bonding is not necessary or sometimes desired for short-term temporary devices because they are removed after healing.

Metals in Orthopedic Implants

Metallic materials are commonly used in orthopedic implants including permanent prosthetic implants and temporary fixation devices because of the proper strength, ductility, fracture toughness, hardness, corrosion resistance, and biocompatibility, which are necessary for load-bearing components such as total joint prostheses and fracture fixation devices. Common biomedical metallic alloys can be categorized into several groups: stainless steels, Co-based alloys, titanium (Ti)-based alloys, and biodegradable alloys such as Mg-based alloys. The metallic materials in the first three groups have been approved by the US Food and Drug Administration and are routinely used in orthopedic implants. Materials of the last group are being developed, and their unique properties may meet special clinical requirements.

Stainless Steels

Stainless steels are iron (Fe)-based alloys that typically contain a certain amount of Cr and Ni. Molybdenum (Mo) and manganese are two other major alloying elements in medical-grade stainless steels and small amounts of carbon, nitrogen, phosphorus, sulfur, and silicon may also be present. The mechanical properties and corrosion resistance are affected by the introduced elements due to alteration of the microstructure. The corrosion resistance of stainless steels is enhanced by alloying with Cr, Ni, Mo, and nitrogen by different mechanisms. Cr with a minimum concentration of 11 wt% plays a major role in the corrosion resistance of stainless steels. Cr with a large affinity for O allows the formation of a Cr-rich oxide (Cr_2O_3) surface film that is adhesive and protective with the ability of self-healing in the presence of O_2 (Brooks et al., 1986). Ni enhances the corrosion resistance by the formation of the closely packed face-centered cubic structure and a protective Ni oxide film on the steel surface. Mo minimizes pitting corrosion by decreasing the formation of Cr carbide because of its strong tendency to form Mo carbide. Similar to Ni, nitrogen improves the resistance against pitting and crevice corrosion by stabilizing the closely packed face-centered cubic structure.

Stainless steels were the first metals used in orthopedic implants in 1926. On account of the good corrosion resistance, asthenic stainless steels, particularly 316 and its variants listed in Table 1, are widely used in orthopedic implants. 316L stainless steel (ASTM F138) with better resistance in Cl^- solutions is developed by reducing its maximum carbon content to 0.03 wt% to retard the formation of Cl carbides in the 1950s. High-nitrogen stainless steel (Orthinox, ASTM F1586) with higher fatigue strength and resistance to pitting and crevice corrosion was further developed in 1986. The emergence of high-nitrogen, Ni-free austenitic stainless steels such as ASTM F2229 for biomedical implants is driven by the adverse effects of released Ni ions on the human body. Better overall properties including strength, resistance to corrosion and wear, and biocompatibility are achieved by ASTM F2229 stainless steel. However, its fracture toughening is compromised because the large nitrogen content makes the steel brittle.

Generally, the biocompatibility of stainless steels is good but less satisfactory than other traditional metals due to faster corrosion in the physiological environment leading to release of toxic Cr^{3+} (Barceloux, 1999a) and Ni^{2+} (Barceloux, 1999b). The fatigue strength and wear resistance of stainless steels are also relatively poor causing possible failure after a while. These are the typical reasons restricting their application to permanent implants at load-bearing sites. Owing to the low cost compared with other alloys, 316L stainless steel has maintained its popularity in fixation devices such as the bone plates, bone screws, and intramedullary nails, which temporarily join two pieces of tissues together and are removed after healing. 316L stainless steel can be used in permanent implants under a small loading such as shoulders. Meanwhile, Orthinox stainless steel has been used as stem materials in permanent hip replacements because the fatigue strength is above the maximal loading stress value of hip stems and also due to enhanced pitting and crevice corrosion resistance (Chen and Thouas, 2014).

Co-Based Alloys

Co-based alloys are usually referred to as CoCr alloys. Similar to stainless steels, Cr at a high concentration leads to spontaneous formation of a passive Cr_2O_3 layer in the human body fluid environment. Mo and Ni are responsible for the further enhanced corrosion resistance. Addition of tungsten (W) enhances the strength by solid solution strengthening and control of the size and distribution of carbides, but impairs the corrosion resistance and corrosion fatigue strength of Co-based alloys. As a result of the chemical composition, the corrosion resistance of Co-based alloys is more than an order of magnitude higher than that of stainless steels.

The first application of Co-based alloy to hip prosthetic implants was in 1939. CoCrMo and CoNiCrMo alloys are now commonly used in implants, and the four main types are cast CoCrMo alloy (ASTM F75), wrought CoCrMo alloy (ASTM F799), wrought CoCrWNi alloy (ASTM F90), and wrought CoNiCrMo alloy (ASTM F562) (Davis, 2003). Some typical mechanical properties of cast and wrought Co-based alloys are presented in Table 2. F75 has a long history in the biomedical implant industry due to the excellent corrosion resistance and high wear resistance, while the casting process results in a coarse grain size compromising the mechanical properties. Wrought alloys coming into use under cold-working conditions later can increase the strength of the alloy. A modified F75 version, F799, is mechanically processed by hot forging to the final shape. The hot-forged F799 alloy has a finer grain structure than as-cast F75 and forms a hexagonal close-packed phase. F799 has approximately twice the fatigue, yield, and ultimate tensile strength than as-cast F75. F90 is a wrought CoCrWNi alloy with addition of W and Ni to improve the machinability and fabrication properties. After F90 is subjected to 44% cold working, the fatigue, yield, and ultimate tensile strengths are remarkably enhanced. The tensile strength of the cold-worked and aged F562 reaches 1795 MPa, which is the highest strength of any implant alloys. The superior fatigue resistance of the wrought Co-based alloys makes them popular in load-bearing components such as total joint replacements. Long-term clinical use shows that Co-based alloys exhibit good biocompatibility and

Table 1 Compositions (wt%) of 316L stainless steel (ASTM F138) and variants (Davis, 2003)

ASTM code/UNS No.	*Cr*	*Ni*	*Mo*	*Mn*	*Si*	*Cu*	*N*	*C*	*P*	*S*
F138/S31673	17.00–19.00	13.00–15.00	2.25–3.00	2.00	0.75	0.50	0.10	0.030	0.025	0.010
F1314/S20910	20.50–23.50	11.50–13.50	2.00–3.00	4.00–6.00	0.75	0.50	0.20–0.40	0.030	0.025	0.010
F1586/S31675	19.50–22.00	9.00–11.00	2.00–3.00	2.00–4.25	0.75	0.25	0.25–0.50	0.08	0.025	0.010
F2229/S29108	19.00–23.00	0.10	0.50–1.50	21.00–24.00	0.75	0.25	$>$ 0.90	0.08	0.03	0.010

Table 2 Typical mechanical properties of cast and wrought cobalt-based alloys (Ratner et al., 2004)

ASTM code	*Condition*	*Young's modulus (GPa)*	*Yield strength (MPa)*	*Tensile strength (MPa)*	*Fatigue endurance limit (at 10^7 cycles, R = −1) (MPa)*
F75	As-cast/annealed	210	448–517	655–889	207–310
	P/M HIP	253	841	1277	725–950
F799	Hot forged	210	896–1200	1399–1586	600–896
F90	Annealed	210	448–648	951–1220	Not available
	44% cold worked	210	1606	1896	586
F562	Hot forged	232	965–1000	1206	500
	Cold worked, aged	232	1500	1795	689–793

provide a long lifetime in the human body. However, there are still concerns about the stress-shielding effect arising from the high Young's modulus (210–230 GPa) and biological response to potential release of toxic Co, Cr, and Ni ions or wear debris from the implants. Co-based alloys are mainly used as articular parts such as the femoral head in permanent orthopedic implants.

Ti-Based Alloys

Pure Ti exists as a hexagonal close-packed structure (α phase) up to 885°C and body-centered cubic structure (β phase) above this temperature. Based on the microstructure, Ti alloys are categorized into four groups: α alloys, near-α alloys, α–β alloys, and β alloys. So far, α and near-α alloys have not been used in orthopedic applications due to their low strength and bending ductility compared with the α–β or β alloys. Tables 3 and 4 list the composition and mechanical properties of orthopedic Ti-based alloys. Aluminum (Al), vanadium (V), niobium (Nb), Fe, zirconium (Zr), molybdenum (Mo), and tantalum (Ta) are common alloying elements in biomedical Ti-based alloys. Among the first-generation α–β alloys, Ti–6Al–4V containing 6 wt% Al and 4 wt% Al are the most extensively used orthopedic alloys. Al is added to stabilize the α phase by increasing the transformation temperature from the α to β phases, and V stabilizes the tougher β phase by lowering the temperature of the β transformation. The fatigue strength of Ti–6Al–4V is higher than that of 316L stainless steel and comparable with that of Orthinox and Co-based alloys. To avoid the toxicity stemming from V, Ti–6Al–7Nb and Ti–5Al–2.5Fe alloys are designed. The elastic modulus of these alloys is about 110 GPa, which is half of those of stainless steels and Co-based alloys. However, this value is still much larger than that of the human bone, and the materials are prone to the stress-shielding effect.

The second-generation Ti-based alloys, β alloys, underwent rapid development in the 1990s. The major merit of β-Ti alloys is the Young's modulus being closer to that of cortical bone, thus minimizing the stress-shielding effect. Some β-structure stabilizing elements such as Nb, Fe, Zr, Mo, and Ta are used as alloying elements to avoid the deleterious effects of Al and V and β–Ti alloys that are considered to be relatively safe in the human body. The ultimate tensile strength and yield strength of α–β and β–Ti alloys are comparable with those of 316L stainless steels but lower than those of Co-based alloys. The fatigue strength of α–β–Ti alloys is higher than that of 316L stainless steels and quite similar to that of Co-based alloys. However, β–Ti alloys have lower fatigue strength than Co-based alloys, and this is one of the drawbacks of β–Ti alloys, and the other drawback is the poor wear resistance.

A stable, protective, strongly adherent oxide film forms naturally and instantly if a fresh surface is exposed to air or moisture because of the strong affinity between Ti and O. As a result, Ti-based alloys possess excellent resistance against general localized attack under the physiological conditions. Compared with stainless steels and Co-based alloys, Ti-based alloys have superior biocompatibility with surrounding tissues arising from not only their excellent corrosion resistance in saline solutions but also the nonreactivity of Ti. First-generation Ti alloys have been reported to cause toxicity and allergic reactions *in vivo* due to Al (Yokel,

Table 3 Typical mechanical properties of orthopedic Ti-based alloys (Davis, 2003)

Alloys	*Young's modulus (GPa)*	*0.2% yield strength (MPa)*	*Ultimate tensile strength (MPa)*
α–β alloys			
Ti–6Al–4V	110	860	930
Ti–6Al–7Nb	105	795	860
Ti–5Al–2.5Fe	110	820	900
β alloys			
Ti–13Nb–13Zr	79–84	836–908	973–1037
Ti–12Mo–6Zr–2Fe	74–85	1000–1060	1060–1100
Ti–15Mo	78	655	795
Ti–15Mo–5Zr–3Al	75–88	870–968	882–975
Ti–15Mo–2.8Nb–0.2Si–0.26O	83	945–987	979–999
Ti–16Nb–10Hf	81	736	851
Ti–35.5Nb–7.3Zr–5.7Ta	55–66	793	827

Table 4 Smooth fatigue strength of orthopedic implant alloys (Davis, 2003)

Alloys	*Test conditions*	*Fatigue limit at 10^7 cycles (MPa)*	*Fatigue limit/yield strength*
α–β alloys			
Ti–6Al–4V	Axial ($R = -1$/292 Hz)	500	0.6
	Axial ($R = -0.1$/292 Hz)	330	0.4
	RBF ($R = -1$/60 Hz)	610	0.7
Ti–6Al–7Nb	RBF ($R = -1$ Hz)	500–600	0.7
Ti–5Al–2.5Fe	RBF ($R = -1$ Hz)	580	0.8
β alloys			
Ti–15Mo–5Zr–3Al	RBF ($R = -1$/100 Hz)	560–640	0.5
Ti–13Nb–13Zr	Axial ($R = -0.1$/60 Hz)	500	0.6
Ti–12Mo–6Zr–2Fe	RBF ($R = -1$/167 Hz)	525	0.5
Ti–15Mo–2.8Nb–0.2Si–0.26O	RBF ($R = -1$/60 Hz)	490	0.5
Ti–35.5Nb–7.3Zr–5.7Ta	RBF ($R = -1$/60 Hz)	265	0.5
Ti–35.5Nb–7.3Zr–5.7Ta–0.4O	RBF ($R = -1$/60 Hz)	450	0.5
Stainless steels and Co-based alloys			
316L stainless steel	RBF ($R = -1$/100 Hz)	440	0.6
CoCrMo alloys	RBF ($R = -1$ Hz)	400–500	0.4–0.5
	RBF ($R = -1$/100 Hz)	500–580	

2000) and V (Barceloux, 1999c). V-free Ti–6Al–7Nb and Ti–5Al–2.5Fe are safer than Ti–6Al–4V as implants, and second-generation Ti alloys developed for the orthopedic implants are alloyed with other elements such as Nb, Zr, Mo, Fe, and Ta, but there is a lack of long-term follow-up data concerning the biocompatibility of β–Ti alloys.

Ti-based implants show good integration with host bone tissues by forming bone minerals on the surface to promote bone attachment. The use of Ti-based alloys as orthopedic implants also benefits from the higher corrosion resistance, lower Young's modulus, and much smaller weight than both stainless steels and Co-based alloys (Long and Rack, 1998). However, compared with stainless steels and CoCrMo alloys, Ti-based alloys exhibit larger wear rates that may be caused by the easier breakdown of the surface-passivated oxide layer in the presence of external stress, and consequently, α–β–Ti alloys are mainly used as femoral stems rather than articular components in hip joint prostheses. The compromised tribological properties of the second-generation β alloys have hampered their application as stems in joint replacements. Owing to the poor bending performance, Ti-based alloys are not recommended for fixation devices such as fracture plates that experience severe bending stress. Nevertheless, Ti-based implants are the best choice for patients who have metal allergy.

Biodegradable Metals

As bone fixation devices such as plates, screws, and pins, the nondegradable metallic biomaterials aforementioned must be removed by a second surgery after sufficient tissue healing. A new class of biodegradable metals has emerged as an alternative to traditional fixation implants. Biodegradable metals are expected to corrode gradually *in vivo* with an appropriate host response elicited by released corrosion products and then dissolve completely upon fulfilling the mission to assist tissue healing with no implant residues (Zheng et al., 2014). The three main types of biodegradable metals are Mg-based alloys, Fe-based alloys, and zinc (Zn)-based alloys (Li et al., 2014). Among them, Mg-based alloys have been most extensively studied *in vitro*, *in vivo*, and clinically. Fe-based alloys are still in the animal test stage and need more long-term clinical trials. One drawback of Fe-based alloys is the slow degradation in the physiological environment because the standard electrode potential at 25°C is −0.44 V for $Fe^{2+}(aq) + 2e^- \rightarrow Fe$ (s) (Peuster et al., 2006). Another problem is that ferromagnetism may interfere with magnetic resonance imaging (Hermawan et al., 2010). Zn-based alloys have a degradation rate between Mg-based alloys and Fe-based alloys, but only a few reports on Zn-based alloys as temporary implants have been found, and the materials are in the initial development stage needing further research. Here, Mg-based alloys that are more mature are described in details.

Although pure Mg does not contain and release alloying elements, the large corrosion rate caused by the unavoidable presence of impurities and relatively low strength drive the development of Mg alloys. Al has a high maximum solubility of 12.7 wt% in Mg and is a commonly alloying element in Mg alloys to enhance both the mechanical properties and corrosion resistance. Typical Mg–Al-based alloys such as AZ31, AZ61, AZ91, and AM60 have been extensively investigated as biodegradable metals. However, Mg–Al-based alloys are not recommended for orthopedic implants due to the potential neurotoxicity induced by Al (Yokel, 2000). Hence, some nontoxic or low toxic elements including Zn, Zr, Ca, strontium (Sr), and tin (Sn) are chosen to form Mg–Zn, Mg–Zr, Mg–Ca, Mg–Sr, and Mg–Sn alloys for orthopedic applications (Zheng et al., 2014). Zn with a relatively high solubility in Mg of 6.2 wt% enhances the mechanical properties such as the ultimate tensile strength and elongation of the Mg alloys due to solid solution strengthening and aging strengthening effects. Zr is used to refine the grain size of Mg-based alloys so that better mechanical properties and lower degradation rate are achieved. Ca and Sr improve the strength and creep properties, but at high concentrations, they may reduce the ductility and corrosion resistance. Addition of Sn improves the tensile strength, yield strength, elongation, and corrosion resistance. Rare earth (RE) elements such as yttrium (Y), gadolinium (Gd), and neodymium (Nd) are also incorporated in

Mg to strengthen the materials and enhance the corrosion resistance. The common Mg-RE-based alloys for biomedical implants include the Mg–Y, Mg–Gd, and Mg–Nd series, and among them, WE43 (Mg–Y–RE) has been extensively investigated for its excellent mechanical properties and corrosion resistance. One important point concerning the alloying elements is that the amount should be controlled to a suitable range; otherwise, negative influence may be observed. Of course, the biosafety of the RE elements is another concern in practice.

In temporary orthopedic implants, Mg-based alloys are more promising than traditional alloys such as stainless steels, Co-based alloys, and Ti-based alloys. First of all, Mg degrades spontaneously in the human body so that a follow-up surgery to remove the implants can be obviated after tissue healing. Mg tends to degrade in the saline environment of human tissues. This natural degradation behavior enables Mg and its alloys to be effectively applied as absorbable implant materials. Secondly, Mg has a Young's modulus (41–45 GPa) close to that of cortical bone (3–20 GPa), thereby mitigating the stress-shielding effect. As a lightweight metal, Mg has a density of 1.74 g cm^{-3} that is also very similar to that of the bone (1.8–2.1 g cm^{-3}). Thirdly, as a vital element in the human body, Mg has good biocompatibility (Staiger et al., 2006). The major concern of Mg alloys in biomedical implant applications is the rapid corrosion in the physiological environment during healing due to the low standard electrode potential of $Mg^{2+}(aq) + 2e^- \rightarrow Mg\ (s)$ (−2.37 V at 25°C). The large degradation rate of Mg alloys has so far limited wider clinical adoption because of mechanical integrity loss before sufficient healing, released alloying elements, excessive hydrogen evolution, and local alkaline environment. Hence, a suitable degradation rate of Mg alloys to allow complete healing is desirable, and surface treatment is a practical and economical strategy to enhance the corrosion resistance of Mg-based alloys (Wu et al., 2013; Zhao et al., 2014).

Ceramics in Orthopedic Implants

The two main types of orthopedic ceramics are bioinert ceramics and bioactive ceramics. Bioinert ceramic materials possess excellent wear resistance, high compressive strength, inherent chemical inertness, and biocompatibility. Bioinert ceramics elicit minimal response from the living tissues because they undergo little physical or chemical alteration over a long time inside the human body. Bioinert ceramic materials are commonly used as articular components in total joint replacements but generally have not been applied to fracture fixation applications mainly due to their poor ductility. Among the various types of bioinert ceramics, Al_2O_3 (alumina), ZrO_2 (zirconia), and silicon nitride (Si_3N_4) have been extensively investigated. Bioactive ceramic materials can bond directly with surrounding living bone tissues. These ceramic materials are often applied as coatings on metal bone implants rather than load-bearing components due to the low mechanical strength. As an example, hydroxyapatite with a porous structure allows bone tissues to grow inside the pores leading to a better integration between the implants and adjacent tissues.

Alumina

Al oxide (Al_2O_3), commonly known as alumina, exists in different metastable phases that irreversibly transform into α-Al_2O_3 if heated above 1200°C. α-Al_2O_3 with a close-packed hexagonal structure has high thermodynamically stability. Alumina has merits such as the high melting point, high hardness, high compressive strength, low friction coefficient, and remarkable resistance against the attack of strong inorganic acids. The high chemical stability of alumina is the basis of the excellent biocompatibility in the physiological environment. These properties arise from the strong ionic and covalent bonds between Al^{3+} and O^{2-} in alumina (Kokubo, 2008). Alumina has high surface wettability due to chemisorption of hydroxyl groups stemming from easy bonding with water molecules and proteins. Boasting high hardness and wettability, alumina implants have excellent scratch resistance and low wear rate. Medical-grade alumina is produced by pressing and high-temperature sintering (1500–1700°C), and small amounts of sintering additives mainly Mg and Cr oxides up to 0.5 wt% are used to obtain high-quality alumina. Mg oxide (MgO) is introduced to promote sintering and limit grain growth resulting in a denser ceramic, and Cr_2O_3 is added to compensate for the decrease in hardness caused by addition of MgO (Ducheyne et al., 2011).

Based on the high compressive strength, superior wear resistance, excellent resistance, and biocompatibility, alumina has been used as articulating surfaces in total hip replacements (Rahaman et al., 2007; Hannouche et al., 2005). Since its first application to orthopedics in 1970, improvements have been made to increase the longevity of alumina components. The purity, porosity, and grain size are three important factors affecting the mechanical properties of ceramics. A decrease in the chemical purity and increase in the average grain size and porosity can undermine the mechanical properties. To increase the long-term reliability of alumina in orthopedic applications, high-purity and high-density alumina with fine and homogeneous grains should be used. During the production of high-quality alumina, high-purity raw materials are selected, and the manufacturing process must be controlled properly. A significant development impacting the production of medical-grade alumina is hot isostatic pressing (HIP). HIP allows shaping at a high pressure with temperature just below the sintering temperature, thereby limiting grain growth and assuring a large density (Yaszemski, 2003). However, the strong ionic covalent bond brings some disadvantages. The major limitation of alumina as bearing components in orthopedic applications is brittle fracture. Meanwhile, like most ceramics (Table 5), alumina has a moderate tensile and flexural strength, and hence, it has not been used commercially in fracture fixation devices.

Table 5 Mechanical properties of orthopedic ceramics (Kokubo, 2008)

Ceramics	*Density ($g\ cm^{-3}$)*	*Young's modulus (GPa)*	*Fracture toughness (MPa)*	*Compressive strength (MPa)*	*Tensile strength (MPa)*
Alumina	3.98	420	3–5.4	4400	282–551
Zirconia (TZP)	5.74–6.08	210	6.4–10.9	1990	800–1500
Si_3N_4 (HPSN)	3.3	304	3.7–5.5	3700	700–1000
Hydroxyapatite (3% porosity)	–	7–13	3.05–3.15	350–450	38–48

Zirconia

From low to high temperature, pure zirconia exhibits three common phases including the monoclinic, tetragonal, and cubic structures that are of interest to biomedical applications. The monoclinic structure exists below 1170°C, and as the temperature is increased, it transforms to a tetragonal phase at approximately 1170°C and then a cubic phase at about 2370°C up to the melting point of 2716°C. The volume and shape changes on cooling associated with the transformation make pure zirconia unsuitable for biomedical applications (Ducheyne et al., 2011). The discovery of transformation toughening has promoted engineering applications of zirconia. In the early stage, metallic oxides such as CaO, MgO, and Y_2O_3 are added to zirconia to stabilize the tetragonal phase under the metastable conditions at room temperature (Kelly and Denry, 2008).

In 1975, Garvie et al. (1975) reported MgO-partially stabilized zirconia (Mg-PSZ) with increased toughness arising from the spontaneous transformation of the metastable tetragonal phase into the more stable monoclinic phase under mechanical stimuli such as stress at crack tips with a large volume expansion. This volume expansion induces compressive stress and retards crack propagation to increase the strength and toughness. Mg-PSZ was the first zirconia used in orthopedics, but no large-scale clinical application has occurred. Yttrium-stabilized zirconia, known as yttria (Y_2O_3)-tetragonal zirconia polycrystal (Y-TZP), was developed using Y_2O_3 as the stabilizer and almost completely formed by tetragonal grains at room temperature (Gupta et al., 1977). In the 1990s, Y-TZP became the materials of clinical choice for large-scale industrial production of hip joint femoral heads because of the higher strength and toughness than Mg-PSZ and CaO-PSZ. Because of the transformation-toughening mechanism, Y-TZP has enhanced mechanical properties including high fracture strength that is twice that of high-density alumina and high flexural strength. Meanwhile, the finer grain size with less residual porosity of Y-TZP delivers better tribological performance than alumina. Young's modulus of zirconia is smaller than that of alumina and similar to those of Ti-based alloys enabling the metal-to-ceramic design transformation. Like alumina, zirconia has excellent biocompatibility, and as a popular alternative to alumina due to the superior properties, Y-TZP is a better candidate in femoral heads in total hip prostheses (Park and Lakes, 2007). However, the metastability of the tetragonal phase at room temperature leads to the transformation of the surface in the presence of water causing microcracking and roughening known as aging or low-temperature degradation (Ducheyne et al., 2011). Fracture of the zirconia ceramic femoral heads occurs after short-term implantation, and thousands of zirconia ball heads have been recalled. The failure is caused by accelerated aging of the balls that are not sufficiently densified during sintering. Intrinsic aging of Y-TZP cannot be eliminated but can be minimized by controlling the manufacturing process to refine the grain size, increase the density, reduce the porosity, and avoid residual stress.

On the heels of the failure of Y-TZP, the need for ceramics with high mechanical performance required by femoral heads and other orthopedic components has led to the development of alumina–zirconia composites. This may benefit from zirconia transformation toughening, thus avoiding the major drawback associated with the transformation in the presence of body fluids. Owing to the aging problem of zirconia, alumina-toughed zirconia has not matched the success of zirconia-toughened alumina (ZTA) (Kurtz et al., 2014). The most successful representative of ZTA is the BIOLOX® delta components (Fig. 3), which are considered the ceramic golden standard in total hip replacements. BIOLOX® delta is composed of α-alumina with fine and homogeneously dispersed zirconia grains (Fig. 4). Alumina provides the wear resistance and hardness, and zirconia together with other additives and high density and small grain size provide the improved mechanical properties. ZTA has played a more important role in the market of ceramic heads. Compared with Y-TZP, ZTA heads offset the higher price with the better mechanical performance and stability. However, the *in vivo* study on the stability of ZTA composites versus aging is not complete, and no critical effects have been demonstrated yet.

Si_3N_4

Si_3N_4 exists in two major crystalline phases, namely, α and β, exhibiting a hexagonal structure with different stacking sequences (Hardie and Jack, 1957). The β phase is the major form of Si_3N_4 ceramics since the α phase is chemically unstable compared with the β phase. β-Si_3N_4 is a well-known tough ceramic due to the unique interlocking microstructure composed of elongated grains that act as a reinforcing phase to enhance the fracture toughness by triggering various toughening effects. In sintering, Si_3N_4 powders mixed with sintering aids such as Y_2O_3, Al_2O_3, and MgO are compacted and heated in a nitrogen atmosphere at high temperature. The amounts of introduced additives affect the properties such as fracture toughness, strength, and oxidation resistance. In contrast to alumina and zirconia, Si_3N_4 ceramic materials usually exhibit higher fracture toughness and are more resistant to crack

Fig. 3 BIOLOX® delta ceramic (A) femoral heads, (B) cup inserts, and (C) knee joint components (https://www.ceramtec.com/biolox/).

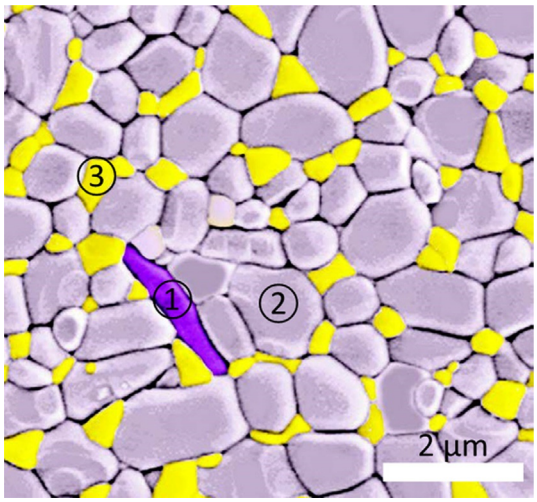

Fig. 4 Microstructure of BIOLOX® delta: ① platelets with crack-stopping function, ② Al_2O_3 particle, and ③ ZrO_2 particle (https://www.ceramtec.com/biolox/).

propagation (Mazzocchi and Bellosi, 2008). Meanwhile, silicon nitride (Si_3N_4) has been commercially employed in many industrial components due to the high hardness, thermal shock resistance, wear resistance, strength, and fracture toughness. Si_3N_4 is also considered to be biocompatible and promotes cell growth (Neumann et al., 2004). These favorable properties make Si_3N_4 ceramics suitable for orthopedic applications such as bearing components in prosthetic hip and knee joints (Mazzocchi and Bellosi, 2008; Bal et al., 2009; Mazzocchi et al., 2008; Bal and Rahaman, 2012). The major concern of Si_3N_4 is surface oxidation particularly in the presence of moisture leading to the formation of a silicon-oxide-rich layer several nanometers thick on the Si_3N_4 surface. Those oxidized layers may chip off over time resulting in third-body wear. Some studies have demonstrated that surface oxidation can be controlled by doping Si_3N_4 with selected additives (Klemm, 2010), but nevertheless, surface oxidation should be properly eliminated before clinical studies.

Hydroxyapatite

Biological fixation is defined as the process in which prosthetic components become firmly bonded to the host bone by ongrowth or ingrowth without the use of bone cement (Jaffe and Scott, 1996). In the late 1960s, the concept of biological fixation of load-bearing implants using bioactive hydroxyapatite coatings was proposed as an alternative to cemented fixation. Apatite represents a family of phosphate minerals with the formula $M_{10}(RO_4)_6X_2$, where M is usually calcium, R is usually phosphorus, and X is hydroxide or a halogen. Among the various forms, hydroxyapatite ($Ca_{10}(PO_4)_6(OH)_2$) is the major mineral constituent in natural bones. Although pure hydroxyapatite has a Ca/P molar ratio of 1.67, biological hydroxyapatite is usually Ca-deficient and carbonated (Vallet-Regi and Gonzalez-Calbet, 2004).

Since hydroxyapatite is chemically similar to the mineral part of natural bones, synthetic hydroxyapatite has strong chemical affinity to hard tissues and offers many advantages in orthopedic applications compared with bioinert ceramics. Based on the excellent biocompatibility, bioactivity, osteoconductivity, and direct bonding to bones, hydroxyapatite can form bond directly with surrounding tissues, thereby promoting new bone growth via the osteoconduction mechanism without causing local or systemic toxicity or inflammation (O'Hare et al., 2010). However, the relatively low strength and toughness of hydroxyapatite make it unsuitable for load-bearing implants. Currently, hydroxyapatite is employed as bioactive coatings on prosthetic implants such as total hip prostheses. It is advantageous to combine the strength of metallic implants such as Ti-based alloys with a bioactive hydroxyapatite coating (Habibovic et al., 2002; Sun et al., 2001). When a hydroxyapatite-coated prosthesis is implanted, a fibrous tissue-free layer containing carbonated apatite forms on the surface and enhances the bonding between the implant and living bone, resulting in early prosthesis stabilization and superior fixation of the prosthesis to the surrounding tissues (Habibovic et al., 2008). The synthetic hydroxyapatite can be made either dense or porous. Synthetic hydroxyapatite with a porous structure is commonly fabricated on metallic implants since the porosity contributes to tissue growth. The commercial method to deposit hydroxyapatite coatings on metallic implants is plasma spraying (Ong and Chan, 2000), but although clinical success of hydroapatite-coated implants produced by plasma spraying has been reported, some problems, particularly the poor coating-metal adhesion strength, are commonly found. Other methods such as thermal spraying, sputter coating, pulsed laser deposition, dip coating, sol-gel deposition, electrophoretic deposition, and ion-beam-assisted-deposition (Mohseni et al., 2014; Yang et al., 2005) have been employed to fabricate high-quality hydroxyapatite coatings on metallic implants. Improvement of the adhesion strength between hydroxyapatite coatings and metallic implants is an essential requirement regardless of the techniques used.

Polymers in Orthopedic Implants

Polymers are commonly used in orthopedics as interpositional cementing materials between the implant surface and bone and as bearing insert of hip and knee joint replacements. Poly(methyl methacrylate) (PMMA) is universally used as a major ingredient in bone cement for the anchorage of cemented orthopedic prostheses. Low friction and high wear resistance are required for polymers as the articulating surface due to the existence of the articulating contact with the opposing surface is usually made of metals. Ultrahigh-molecular-weight polyethylene (UHMWPE) is commonly utilized as a load-bearing surface in total joint replacements. The predominant problem of the metal-on-polymer articulating surface is the production of wear particles such as polymer debris. New methods to enhance the wear resistance without impairing other essential properties of UHMWPE are desired for UHMWPE as orthopedic components. Polyetheretherketone (PEEK) and biodegradable polymers have also emerged as components in joint prostheses and temporary fixation devices.

Poly(Methyl Methacrylate)

Charnley (1960) developed the first bone cement using PMMA for hip joint prostheses. Although alternative materials are available and continue to be designed, PMMA remains the major ingredient in bone cement for anchorage of cemented orthopedic prostheses, particularly for femoral stem fixation to bone in total hip replacements and femoral and tibial components fixation to bone in total knee replacements. PMMA provides mechanical anchoring of the prosthesis and as an elastic buffer allows gradual transfer of loading from the prosthesis to the bone due to the low Young's modulus. The cement location in a total hip replacement is shown in Fig. 5 (Yaszemski, 2003).

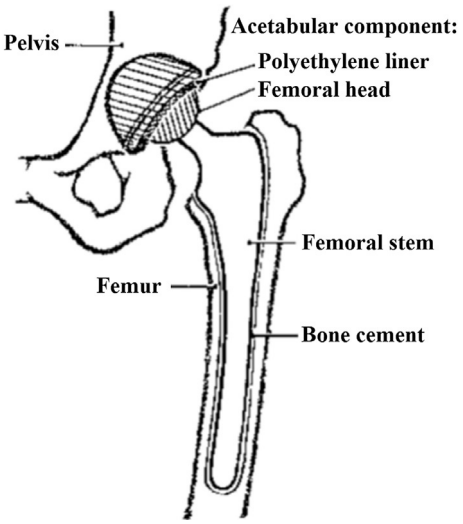

Fig. 5 Schematic of bone cement and implant in hip arthroplasty.

Commercial bone cement is primarily composed of the PMMA powder and monomer methyl methacrylate (MMA) liquid. The powder generally contains an initiator (benzoyl peroxide) and radiopacifier (barium sulfate). An activator (*N*,*N*-dimethyl-*p*-toluidine) and stabilizer (hydroquinone) are also added to the liquid. When the powder and liquid are mixed, *N*,*N*-dimethyl-*p*-toluidine reacts with benzoyl peroxide to produce free radicals, and the monomer MMA liquid is polymerized. During this process, the monomer liquid wets the particle surface of the polymer powder and links them together after polymerization. Finally, the polymerization process ceases. The viscosity of the bone cement gradually changes into a doughlike state and eventually hardens into a rigid polymer with chain propagation. The time from the beginning of mixing of the bone cement components to formation of bone cement with enough viscosity to be handled as a cohesive mass is the dough time. The place of the cement in the cavity should be operated from this point until the bone cement is too hard to mold anymore.

In spite of the aforementioned advantages, bone cement has some drawbacks (Passuti and Gouin, 2003; Lewis, 1997). The polymerization reaction releases a large amount of heat and raises the temperature of the bone cement causing necrosis of the adjacent bone. Release of the unreacted monomer MMA can induce allergic reactions and hypotension due to potential cardiotoxic effects in the human body. Another concern is that shrinkage of the cement volume during curing produces gaps and less contact between the cement and prosthesis and between the cement and the bone. During curing of the bone cement, porosity may develop, and large pores are detrimental to the mechanical properties. Cement wear fragmentation produced during long-term usage can interact with the surrounding tissues to evoke inflammatory tissue response and increase bone destruction. Nonetheless, despite these disadvantages, bone cement is quite prevalent clinically.

Ultrahigh-Molecular-Weight Polyethylene

UHMWPE is a linear polyolefin with a repeating unit of $-CH_2CH_2-$. Medical-grade UHMWPE has long chains with a molecular mass of 2×10^6–6×10^6 g mol^{-1} and is a semicrystalline polymer with a set of ordered regions embedded in a disordered amorphous phase (Turell and Bellare, 2004). UHMWPE has low friction, high wear resistance, good toughness, high impact strength, high resistance to corrosive chemicals, excellent biocompatibility, and low cost. UHMWPE has been used clinically in joint implants for over 40 years, particularly as an articular liner in a total hip replacements and tibial insert in total knee replacements. In 1962, UHMWPE was first used as acetabular components and has become the dominant bearing materials in total hip replacements since the 1970s. However, the wear of UHMWPE in contact with harder components made of metals or ceramics was a major problem in orthopedics in the 1980s mainly due to continuous reorientation of the polymer chains. The wear debris may induce osteolysis leading to loosening of implants and weakening of the bone structure. There was a major breakthrough in the development of highly cross-linked UHMWPE in the late 1990s. Cross-linking of UHMWPE is widely implemented by radicalizing the side chains with radiation such as gamma ray, electron beam, or chemicals such as peroxide to improve the wear resistance due to the decreased mobility of the polymer chains after cross-linking (Lewis, 2001). To improve the oxidation resistance, the cross-linked UHMWPE is thermally treated. Highly cross-linked UHMWPE has been successfully used in load-bearing joints and becomes the standard in total hip replacements.

Prior to implantation, orthopedic implants are generally sterilized by gamma irradiation in ambient air. Gamma ray induces the formation of free radicals through chain cleavage. After gamma irradiation, free radicals may still exist in the polymer and react with available O species during storage or *in vivo* induce detrimental oxidation of UHMWPE (Premnath et al., 1996). Although highly cross-linked UHMWPE has enhanced wear resistance, other properties such as the ductility, fracture toughness, fatigue resistance,

and tensile strength may be compromised by gamma irradiation (Lewis, 2001; Premnath et al., 1996). Nonionizing methods such as sterilization using ethylene oxide gas or gas plasma emerge, and some stabilization treatment has also been conducted after cross-linking to eliminate the deleterious influence mentioned earlier (Kurtz et al., 1999). The antioxidant vitamin E is also incorporated into cross-linked UHMWPE to suppress oxidation by reacting with free radicals (Bracco and Oral, 2011). There is still no clinical history in joint replacement components even though vitamin E exhibits safety and biocompatibility. Therefore, methods to enhance the wear resistance without impairing any other essential properties of UHMWPE and long-term clinical application are desired for UHMWPE in orthopedic applications.

Polyetheretherketone

PEEK is a member of the polyaryletherketone (PAEK) family consisting of an aromatic molecular backbone with combinations of ketone and ether functional groups between the aryl rings. Two members of the PAEK families considered for implants are polyetherketone and PEKK with the latter being dominant in implants (Kurtz, 2011). PEEK is a polyaromatic semicrystalline thermoplastic polymer composed of a crystalline phase and an amorphous phase. The typical crystalline content of injection-molded biomedical PEEK ranges from 30 to 35%, which has a close relationship with the mechanical properties. The stabilized chemical structure of PEEK confers its stability at high temperature, resistance to chemical and irradiation degradation, and higher strength than many metals. PEEK is extremely resistant to attack by all substances apart from concentrated sulfuric acid and is also biocompatible. Thermal degradation is not a concern for PEEK during clinical applications. Unlike UHMWPE, PEEK has notable resistance against gamma and electron beam irradiation, and the implant components can be effectively sterilized by gamma irradiation in air without degradation of the mechanical properties. The chemical inertness, biocompatibility, mechanical properties, and radiolucency render PEEK suitable biomaterials in orthopedic implants (Kurtz, 2011).

In the 1990s, PEEK emerged as a leading thermoplastic candidate as a substitution for metals in orthopedic implants. Although PEEK-based biomaterials are now widely accepted in the spine field, they continue to be clinically investigated as hip stems and articular bearing components in joint prostheses and fracture fixation devices (Kurtz and Devine, 2007). PEEK composites have been developed by adding certain additives to enhance the strength and stiffness. Among the various additives, carbon and glass fillers were first used as reinforcement additives in PEEK, and carbon-fiber-reinforced PEEK (CFR-PEEK) is currently studied as components in orthopedic implants (Li et al., 2015). The strength and elastic modulus of the CFR-PEEK composite depend on the percentage, size, length, and orientation of the fibers. Young's modulus of PEEK is 3–4 GPa, which is near but not identical to that of the human bone. With increasing percentage of carbon fibers, Young's modulus and tensile strength rise, but the elongation at break decreases, indicating that more robust PEEK is obtained. The CFR-PEEK composite with a certain composition may have mechanical strength and elasticity close to those of cortical bone, thus providing support and minimizing the stress-shielding effect in fracture fixation applications. The CFR-PEEK composite also has excellent fatigue resistance, wear resistance, and durability making them suitable for long-term joint implants. No adverse side effects have been observed from PEEK, but it is still considered bioinert because of the slow reaction with surrounding tissues. To overcome this problem, increasing efforts have been made to enhance bone growth around the implants to improve fixation of PEEK components with bone by incorporating bioactive materials such as hydroxyapatite and Ti dioxide as a filler or applying a surface coating on PEEK (Abdullah et al., 2015). Overall, the short-term effective performance of the PEEK composites as orthopedic implants is supported by preliminary clinical data. However, long-term clinical trials should be done to prove the superiority over traditional orthopedic materials.

Biodegradable Polymers

Among the different kinds of synthetic biodegradable polymers, polyesters are the earliest and most extensively used. Polyesters are generally synthesized from a wide range of monomeric units via ring opening polymerization. Degradation of polyesters occurs though a nonspecific hydrolytic scission of the ester bonds, which leads to a reduction in the molecular weight, strength, and mass of the polymer. The most widely investigated polyesters for orthopedic applications are polyglycolide (PGA), polylactide (PLA), and their copolymer (Middleton and Tipton, 2000). PGA has high crystallinity (45–55%), large elastic modulus, and small solubility in organic solvents. PGA has a melting point of $> 200^{\circ}C$ and glass transition temperature of 35–40°C. On account of the excellent fiber-forming ability of PGA, self-reinforced PGA with a Young's modulus of 12.8 GPa that is close to that of the bone and remarkably higher than those of other degradable clinical polymers can be formed. PGA has been investigated as bone internal fixation devices because of the higher rigidity. PGA is broken down into glycine that can be excreted in urine or converted into carbon dioxide and water via the citric acid cycle. However, PGA degrades rapidly due to the hydrophilic nature, and the acidic products with low solubility may cause inflammation in surrounding tissues, thereby hampering orthopedic applications (Maurus and Kaeding, 2004).

Lactide is a chiral molecule existing in two optical forms: L-lactide and D-lactide. Polymerization of lactide produces four materials: poly(L-lactide) (PLLA), poly(D-lactide), poly(D,L-lactide) (PDLLA), and meso-PLA. Among them, PLLA is commonly studied for orthopedic fixation devices. PLLA has approximately 37% crystallinity depending on the polymer processing parameters and molecular weight. PLLA has a glass transition temperature of 60–65°C and melting temperature of 173–178°C (Maurus and Kaeding, 2004). The degradation product of PLA is lactic acid, which is incorporated into the tricarboxylic acid cycle and finally processed into carbon dioxide and water. PLLA has good tensile strength, low extension, and elastic modulus of approximately 4.8 GPa. PLLA has been used in orthopedic fixation devices (Nair and Laurencin, 2007). However, being more hydrophobic than

PGA, PLLA tends to degrade more slowly than PGA. Meanwhile, as an amorphous polymer, PDLLA has a rapid degradation rate. Therefore, copolymers of L-lactide with glycolide or D,L-lactide have been developed to achieve better comprehensive properties for orthopedic devices (Middleton and Tipton, 2000). The degradation rate of the copolymer depends on various factors such as the ratio of the monomers, molecular weight, and structure of the copolymer.

Conclusion

Selection of the proper materials for orthopedic implant relies on the specific applications. Despite the success of traditional materials such as Ti-based alloys, ZTA, and UHMWPE, loosening of acetabular and femoral components is the most significant problem in total joint replacements. To avoid premature failure of orthopedic prostheses and assure long-term performance, desirable physical, chemical, and biological characteristics are needed especially for permanent implants. Fracture fixation devices can be made with traditional metals, but biodegradable metals and polymers have attracted much attention. To satisfy the ever-increasing demand, new and better biomaterials are being developed continuously.

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