Bio implant materials: Requirements, Types -and Properties – A review

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ABSTRACT

Bio implants are biosynthetic materials used in medical applications to fix fractures, to replace a bone or to support the healing process of a damaged bone. Although the primary function of implant material is to provide stability to bone fractures, they also indirectly assist in the biological aspects of bone healing by minimizing micro-motion at the bone-implant interface.

In the early years of implant surgeries pure metals were primarily employed due to their excellent specific strength, fatigue strength, corrosion resistance and wear resistance. However the issues that ensued relating to poor bio-compatibility, weak osseointegration, excessive rigidity and bone implant failures due to instability lead researchers to look for smart bio-materials to tackle these difficulties as they realized that mechanical and biological aspects of bone healing are closely inter-related and are crucial for obtaining a satisfactory clinical outcome.

This further spurred the evolution of bio implant materials paving way for Ceramic, Polymer and Hybrid implant materials. This article highlights the review of various bio implant materials that are in current use along with their advantages and disadvantages. The performance aspects of the different types of implants are also compared and illustrated.

KEY WORDS: Biomaterials, Bio Implants, Osseo-Integration, Bio-Compatibility.

1. INTRODUCTION

Bio materials are those materials that are accepted by living tissues and can be used for tissue replacements. On a macroscopic level these devices are used to fix or replace a bone and to support its healing process.

With the worldwide increase in the average age of population there is a subsequent increase in the number of surgical procedures which has in-turn urged researchers to improve and optimize bio materials.

Orthopaedic implants are used routinely worldwide for fixation of long bone fractures and non-unions, for correction and stabilization of spinal fractures and deformities, for replacement of arthritic joints, and for other orthopaedic and maxillofacial applications (Stuart, 2003).

Mechanical stabilization, facilitating pain relief, normal functioning by providing stability to bone fractures (for example, unwanted shear stress is reduced indirectly by orthopaedic implants thereby assisting in the biological aspects of bone healing (Carter, 1998)) and faster healing of degenerative body conditions are some of the functions that are required of a bio implant.

The above factors implement a stringent limitation on the available materials that can be employed to satisfy the requisite conditions.

From a surgical point of view elements such as patient characteristics (e.g. smoking, excessive alcohol use, diabetes, medications), local factors (e.g. difficult anatomical site, infection, poor vascular supply, irradiation), and surgical and implant factors (surgical technique, inadequate implant characteristics or application of the implant) are given high priority before employing a material as an implant for surgery (Bishop, 2012).

With a diverse set of factors to be taken into consideration, the search for the *ideal* bio implant material remains at large. This stimulated research along various directions which further presented numerous insights on determining the apt bio material.

As a result, compared to the early years of implantology where implants were just considered as mechanical devices (Stuart, 2003), we have come a long way in understanding the correlations that determine the success of an orthopaedic implant.

History: The long historical roots depicting the tree of implants can be traced back to the Egyptians and the Mayan civilization. Their early works focussed on replacing a broken tooth and thus can be rightfully referred to as the pioneers in the art of rehabilitation. People of the Incas Empire used pieces of sea shells to replace missing teeth by gently tapping them into bones (Anjard, 1981).

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However the notable breakthrough occurred by the dawn of the modern era between 1935-1978 where metal alloys, ceramics and synthetic polymers started emerging. This evolution slowly substituted the previously used naturally derived materials due to their high degree of sophistication which ensured superior efficient clinical results. This spur was led by the birth of screw fixtures made of an alloy named Vitallium (Cobalt-Chromium-Molybdenum) around mid-1930s by Drs. Alvin and Moses Strock (Tamal, 2015). The Strock brothers used vitallium screws and experimented on the anchorage and support of replacement of a missing tooth for which they were acknowledged in selecting a biocompatible metal to be used in the human dentition (Linkow, 1991). They were able to achieve an implant life of 15 years (Monika, 2016).

Their research led to further analytical understanding of bone physiology, drilling protocol, implant design and surface texture, initial implant stability, single-stage implant surgery, and immediate loading of implants (Tamal, 2015).

Subsequently S. Formiggini (known as the "Father of Modern Implantology") and Zepponi by 1940s created a stainless steel spiral screw implant which facilitated bone growth into the metal (Linkow, 1991). This marked a vital step in the progression of metallic materials as bio implants.

Basic requirements: As discussed earlier a bio implant is required to have a specific set of characteristic properties to enable them to perform its duty. Table1 gives an elaborate view of the various requirements to be considered while selecting a material to be used as a bio implant.

It indicates that all 3 of the factions of compatibility, mechanical properties and manufacturing criterion are necessary, meaning that the selection of the implant material itself is the principal criterion for proper functioning. No amount of design changes can help if the material is not biologically and mechanically compatible (Soumya and Rajarshi, 2012). Therefore the chosen material must display the following properties (Soumya and Rajarshi, 2012; ASM International, 2003):

- A biocompatible chemical composition to avoid adverse tissue reactions
- The human body must be compatible with the material used for the prosthesis.
- Excellent resistance to degradation (e.g., corrosion resistance for metals or resistance to biological degradation in polymers)
- Acceptable strength to sustain cyclic loading endured by the joint.
- High wear resistance to minimize wear debris generation.
- The implant should have the desired balance of mechanical and physical properties necessary to perform as expected.
- A low modulus to minimize bone resorption.
- The device under question should be relatively easy to fabricate, being reproducible, consistent, and conforming to all technical and biological requirements.

Table.1. Implant material requirements in orthopaedic applications
Source: Ref (Marc Long and Rack, 1998)
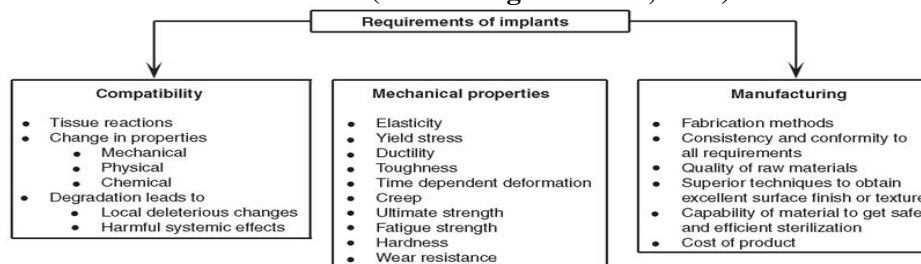
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Metallic implants: The use of stainless steels, cobalt based alloys and titanium and its alloys are well established due to their good biocompatibility, satisfactory mechanical strength and superior corrosion resistance. Nevertheless metallic implants are generally much stiffer than natural bones, leading to a major source for bone resorption namely stress shielding and eventual failure of such implants (Xiaojian, 2016).

In earlier days stainless steel was easily available and due to this it was frequented as an implant material. ASTM F138 also commonly known as 316L (an alloy mainly constituting of Chromium, Nickel, and Molybdenum) was the first form of stainless steel that was commercially used.

Table.2. Comparison of mechanical properties of commonly used orthopaedic alloys
(ASM International, 2003)

Alloy designation	Elastic modulus GPa	Yield strength MPa	Ultimate tensile strength MPa
Stainless steel	200	170–750	465–950
Co-Cr-Mo	200–230	275–1585	600–1795
Commercially pure Ti	102	692	785
Ti-6Al-4V	110	850–900	960–970

Since stainless steel is susceptible to crevice corrosion and pitting (Alex, 2014), researchers started looking for materials that provided better compatibility and corrosion resistance properties. Moreover from Table.2, it is evident that stainless steel has a high modulus of elasticity of 200 GPa which is not preferred (as it is nearly 10 times that of the modulus of elasticity of bone).

Following the use of stainless steel, the search for alloys with better compatibility continued and this resulted in cobalt-based alloys. Alloys like Co-Cr-Mo also faced the same problem of having a very high modulus of elasticity despite its excellent corrosion resistance, wear resistance, and fatigue strength (Table.3).

Table.3. Metal alloys along with their advantages and disadvantages
(Marc Long and Rack, 1998)

	Stainless steels	Cobalt-base alloys	Ti & Ti-base alloys
Designation	ASTM F-138 ('316 LDVM')	ASTM F-75 ASTM F-799 ASTM F-1537 (Cast and wrought)	ASTM F-67 (ISO 5832/II) ASTM F-136 (ISO 5832/II) ASTM F-1295 (Cast and wrought)
Principal alloying elements (wt. %)	Fe(bal.) Cr(17-20) Ni(12-14) Mo(2-4)	Co(bal.) Cr(19-30) Mo(0-10) Ni(0-37)	Ti(bal.) Al(6) V(4) Nb(7)
Advantages	cost, availability processing	wear resistance corrosion resistance fatigue strength	Biocompatibility corrosion minimum modulus fatigue strength
Disadvantages	long term behaviour high modulus	high modulus biocompatibility	power wear resistance low shear strength
Primary utilisations	Temporary devices (fracture plates, screws, hip nails) Used for THRs stems in UK (high Nitrogen)	Dentistry castings Prostheses stems Load-bearing components in TJR (wrought alloys)	Used in THRs with modular (CoCrMo or ceramic) femoral heads Long-term, permanent devices (nails, pacemakers)

Titanium and its alloys: The frequented usage of pure titanium and its alloys as materials for metallic transplants in the field of orthopaedics and dentistry can be attributed to its distinctive medley of physical, chemical and biological properties. Properties such as low elastic modulus close to that of human bone (thereby earning Ti implants the name "Osseointegrated implants"), excellent resistance to corrosion (almost equivalent to that of Platinum), superelasticity and low specific strength have made titanium the "golden standard" for bio implant materials (Adell, 1990; Jemt, 1996; Niinomi, 1998).

The American Society for Testing and Materials (ASTM) categorises the available titanium for implant bio materials into six distinct types. These six materials are further divided into four grades of commercially pure titanium (CpTi) and two titanium (Ti) alloys.

The stoichiometric composition of commercially pure titanium (CpTi) (ASTM F67) allows its classification into 4 grades (Grade I, Grade II, Grade III, Grade IV) that vary mainly in oxygen content, with Grade IV having the most (0.4%) and Grade I the least (0.18%) (Nikitas, 2000).

Although oxide properties are not affected, mechanical differences exist between the different grades primarily because of the contaminants that are present in minute quantities (Albrektsson, 1985). These contaminants

(which include trace elements of carbon, oxygen, nitrogen and iron) further contribute to a noticeable improvement in the mechanical properties of pure titanium.

Titanium alloys are classified as follows: α, near-α, α+β, metastable β or stable β depending upon their room temperature microstructure (Collings, 1984; Polmear, 1981).

Alpha titanium alloys have a hexagonal closely packed (hcp) crystallographic structure with a high melting point of (1650 ˚C). Pertinent to the above classification, α-stabilizers such as Sn, Al, O and β-stabilizers, such as Mo, Si, V are alloyed with titanium to obtain α and β forms.

The properties of various titanium alloys depend on their composition, the relative proportions of the α and β phases, and the alloy's prior thermal treatment and thermo-mechanical processing conditions. Depending on the composition and heat treatment, both the alpha and beta forms may coexist (Parr, 1985; McCracken, 1999).

For instance β alloys (be it metastable or stable) possess high strength, good formability and high hardenability whilst also offering the remarkable probability of combined superior corrosion resistance and low elastic modulus (Bania, 1993; Schutz, 1993).

On the other hand despite their low ambient temperature strength α and near-α titanium alloys are predominantly used as bio materials due to their superior corrosion resistance alone. In contrast, α+β alloys exhibit higher strength due to the presence of both α and β phases (Marc Long and Rack, 1998).

The mechanical properties such as Elastic modulus, Yield strength and Ultimate tensile strength along with their microstructure for different titanium alloys are represented in the table given below.

Table.4. Mechanical properties of materials used as orthopaedic implants
(Soumya and Rajarshi, 2012; ASM International, 2013; Marc Long and Rack, 1998;
Reham and Michael, 2015; Dewei, 2017).

Alloy designation		Microstructure	Elastic modulus GPa	Yield strength MPa	Ultimate tensile strength MPa
Commercially pure Ti	Cp Ti grade I	α	102	170	240
	Cp Ti grade II	α	102	275	345
	Cp Ti grade III	α	102	380	450
	Cp Ti grade IV	α	104	483	550
Ti-6Al-4V		α/β	110	850-900	960-970
Ti-6Al-4V ELI		α/β	113	795	860
Ti-6Al-7Nb		α/β	105	921	1024
Ti-5Al-2.5Fe		α/β	110	914	1033
Ti-12Mo-6Zr-2Fe		Metastable β	74-85	1000-1060	1060-1100
Ti-15Mo-5Zr-3Al		Metastable β Aged β+α	75 88-113	870-968 1087-1284	882-975 1099-1312
Ti-15Mo-2.8Nb-3Al		Aged β+α Metastable β Aged β+α	88-113 82 100	1087-1284 771 1215	1099-1312 812 1310
Ti-Zr		Cast α'/β	N/A	N/A	900
Ti-0/20Zr-0/20Sn-4/8Nb-2/4Ta+(Pd, N, O)		α/β	N/A	726-990	750-1200
Ti-15Zr-4Nb-2Ta-0.2Pd		α/β	94–99	693–806	715–919
Ti-13Nb-13Zr		α'/β	79	900	1030
Ti-15Mo-3Nb-0.3O (21SRx)		Metastable β + silicides	82	1020	1020
Ti-35Nb-7Zr-5Ta (TNZT)		Metastable β	55	530	590
Ti-29Nb-13Ta-4.6Zr		β	80	864	911
Ti-35Nb-7Zr-5Ta 0.4O (TNZTO)		Metastable β	66	976	1010

Ceramics: Bio materials must be distinguished from biological materials because the former are the materials that are accepted by living tissues and, therefore, they might be used for tissue replacements, while the latter are the materials being produced by various biological systems (Sergey, 2010). Thus bioceramics derives its name due to its ceramic origin under the biomaterials.

From the late 1960s the use of ceramics as biomaterials became more common in the field of implantology. This can be attributed to its superior bio compatibility, resistance to cytotoxicity and biodegradability which aids faster bone growth and rehabilitation (Kawahara, 1987). However the most distinctive property that differentiates bioceramics from other bioimplant materials is its excellent resistance to corrosion and wear when used in the bearing

of joint prostheses (Hayashi, 1993). Certain varieties of bio ceramics even exhibit drug-delivery capability (Salinas and Vallet-Regı, 2007).

The properties of bio ceramics can either be inert or active in nature with respect to its physiological environment. Accordingly bioceramics can be classified into three categories namely Inert Bioceramics, Surface Reactive Bioceramics and Resorbable Bioceramics.

Bioinert ceramics as the name suggest are inert in nature. The most commonly used ceramic of this kind is Al_2O_3 as load-bearing prostheses and dental implants.

The first usage of ceramics as bioimplant materials was around 1979 when Harms and Mausle two scientists, experimented the biocompatibility of alumina (Al_2O_3) on rat macrophages (Harms and Mausle, 1979). Direct cytotoxicity was not observed in their experimentation.

Ceramics of this kind in general have high strength, superior wear and corrosion resistance. The coefficient of friction of alumina-alumina surface is better than that of metal-polyethylene surfaces (Ratner, 2004).

The increasing usage of bioinert ceramics in the field of orthopaedic surgery is credited to the fact that the affinity of the bone towards ceramics is better than that of metal alloy (Matsuguchi, 1991).

Despite these advantage the primary drawback of using alumina-base, ball and socket joint is the relatively high elastic modulus of alumina (>300GPa) which can be responsible for stress-shielding effects. To tackle this problem zirconia (ZrO_2) - based products having low modulus elasticity (~200GPa) were introduced (U.S. Food and Drug Administration) this even led to recall of hip implants using zirconia femoral heads (Hulbert, 1982).

In case of controlled surface reactive ceramic group the primary material of interest is bio glass whose major application is serving as a coating material for prosthetic metals or alumina ceramics in which it serves as a carrier for chemically bonding the implant to the bone (Hulbert, 1982).

The crucial role of Surface reactivity is that it supplements the implant-bone bonding process while also adding on to the bone tissue formation. Time dependent changes are prevalent due to reactions at the material/tissue interface thereby ensuing alterations in surface characteristics of implanted materials and surrounding tissues (Ducheyne and Qiu, 1999).

Bio glass is defined as a glass defined to elicit specific physiological responses, including providing surface reactive silica, calcium, phosphate groups and alkaline pH at an interface with tissues (Hulbert, 1982).

Hydroxyapatite [$Ca_{10}(PO_4)6(OH)_2$], commonly known as HA is frequented in the field of dentistry and orthopaedics as a bio material because of its close chemical resemblance (Ca/P=1.67) with bone and teeth (Tsui, 1998; Hukovic, 2003; Chen, 2005).

In this way bio ceramics can also be classified in terms of their Ca/P ratio. For example: dicalcium phosphate, tricalcium phosphate and tetra calcium phosphate have calcium phosphate ratios of 1, 1.5 and 2 respectively and this ratio must be accurately maintained to prevent further decomposition to stable products during heat treatment.

The strength of these ceramics was found to increase with increase in Ca/P ratio, reaching the maximum value around Ca/P~1.67(i.e. HA) and it decreases suddenly with Ca/P>1.67 (Suchanek and Yoshimura, 1998).

HA due to its bio compatibility and bio activity has been used as coating material in dental implants, on repair, scaffoles and skeletal implants (Liang, 2004).

Factors such face composition, crystallinity and microstructure of HA coating are critical in designing its cell response and mechanical performance (Kantesh, 2006).

Dense sintered HA provides an apatite surface which enriches the implant material with superior biocompatibility that is especially absorbed with hard tissues (Lange, 1989). Although HA has the faces fatigue failure and has low tensile strength (Putter, 1984; Groat, 1983), they are widely used for bone substitution and tooth replacement (Putter, 1983; Osborn, 1985; Groat, 1982).

The bending compressive and tensile strength of dense HA are in the ranges 32-250MPa, 120-900MPa and 38-300MPa respectively.

Similarly the values for porous HA bio ceramics are in the ranges of 2-11MPa, 2-200MPa and ~3MPa respectively (Suchanek and Yoshimura, 1998).

Resorbable ceramics are those that contains elements such calcium and phosphorus that are easily processed and are biodegradable within our biological systems.

Initially around 1974 the US Army Medical Research development command funded a research project on bio ceramics leading to the development of a porous resorbable ceramic consisting of β– $Ca_3(PO_4)_2$ (β- phase calcium phosphate), β–phase tricalcium phosphate ceramics (β -TPC); this was biodegradable in nature and was overcome with new bone ingrowth when implanted in bone (Dee, 2003).

Many experiments over time have given concrete proof that β -TPC is a bio compatible matrix for bone regeneration and is also effective in bone re construction in any form: multi crystalline, porous, powdered and/or granular (Dee, 2003). However $Ca_3(PO_4)_2$ remains as the primary resorbable bio ceramic till date.

The notable functions of resorbable bio ceramics include serving as scaffold permitting tissue replacement. In present age they are widely used in the area of treatment of maxillofacial defects (Sergey, 2010).

Table.5. Fabricability and bioinertness of ceramics (Kawahara, 1987)

Materials	Fabrication Temperature (°C)	Fabrication method	Bioinertness	Bending strength (MPa)	Modulus of strength elasticity (GPa)
Bioactive ceramics					
(i) Bioglass			***	98	
(ii) Crystallized glass	1050-1450	Melting Sintering	***	137-150	98
(iii) Hydroxyapatite	900-1200	Sintering	***	105-215	41.2-121.0
Bioinert ceramics					
(iv) Vitreouscarbon			****	68.6-205	16.8-27.4
(v) LTI carbon	1500	Sintering	****	508	50-70
(vi) Zirconia	1500-2000	Sintering	****	780	225
(vii) Silicon nitrate	1800-2000	Sintering	****	490	294
(viii) Silicon carbide	2000	Sintering	****	490	392
(ix) Polycrystal alumina	1700-1800	Sintering	****	372	372
(x) Single crystal alumina	2050	Melting	*****	1274	392
Hard tissues					
Cortical bone				29.4-186	10.8-17.6
Cancellous bone				0.4	0.09-0.19
Enamel				3.2-12.7	41-56
Dentine				6-16	12-18

Polymers: Polymers are macromolecules consisting of a large number of covalently bonded monomers that act as repeating units (Dee, 2003).

Fundamentally polymers can be divided into two types namely natural and synthetic. Natural polymers are those that constitutes of naturally occurring monomers like cellulose, collagen and natural rubber. On the other hand synthetic polymers are those that are artificially synthesised in laboratories. Polyvinyl chloride (PVC), polyethylene (PE) and polymethyl methacrylate (PMMA) are examples of synthetic polymers.

Depending on the local environment natural polymers satisfy wide range of functions. For example proteins such as myosin and collagen function as catalyst; lipids function as energy storing units; polysaccharides such as cellulose and amylose act as intercellular communication and membrane support (Yu, 2009).

The greatest advantage of natural polymers is that they degrade over time by their metabolic processes that take place in our body leaving no trace of it after it completes its destined function, while the greatest disadvantage is their reproducibility (ASM International, 2003).

The second class of synthetic polymers can be obtained in laboratories by various polymerization reactions (addition polymerization yields polymers like PE and PVC and condensation polymerization gives us Nylon-PET). Despite the complexity in preparation in some of the synthetic polymers they are used for specific applications for example the products of condensation polymerisation gives materials that exhibit a range of hydrophobic and hydrophilic properties and the latter kind (2-Hydroxyethyl methacrylate commonly known as polyHEMA) can be used in contact lenses (ASM International, 2003).

Since wide ranges of diversified monomers are available, they can be integrated to create polymers of different characteristic properties or to overcome a specified drawback of a particular type of polymer (Advances in polymer sciences, 2007).

Table.6. Different polymers and its application (Soumya and Rajarshi, 2012)

Name of polymer	Application
Polyethylene	Joint replacement
Polypropylene	Sutures
PET	Sutures, vascular prosthesis
Polyamides	Sutures
PTFE	Soft-tissue augmentation, vascular prostheses
Polyesters	Vascular prostheses, drug- delivery Systems
Polyurethanes	Blood-contacting devices
PVC	Tubing
PMMA	Dental restorations, intraocular lenses, joint replacement (bone cements)
Silicones	Soft-tissue replacement, ophthalmology
Hydrogels	Ophthalmology, drug-delivery Systems

Table.6, represents the usage of polymers as bio materials in various applications. It can be observed that are widely used in the field of tissue engineering and as biomaterials in orthopaedic and dental applications. Polymeric implants; biodegradable and non-biodegradable have been explored for diverse central nervous system (CNS) bioengineering applications (Dan, 2011).

Although polymers that are bio degradable in nature have been developed and used, in general the materials used in orthopaedic fixation are limited to PLA and PGA because of their well-established bio compatibility (Ying, 2008).

Recently the usage of resorbable polymers like polyglycolic acid (PGA), polylactic acid and PLA-PGA copolymers in spine reconstructive surgery is gaining attention in the orthopaedic community (Jiya, 2007).This is because PLA and PGA show high level of biocompatibility particularly poly-L-lactic acid (PLLA) has demonstrated high compatibility to that of bone and neural tissues.

Most of the time the natural polymers used is very similar in nature to the biological environment in which they are employed and this facilitates biodegradation of the polymeric component. A combination of polylactic acids (PLA) and polyglycolic acid (PGA) of various forms constituting biodegradable polymers are being used in the human body as fixation devices in the form of pins, screws, etc. (Jiya, 2007).

2. CONCLUSION

Williams defined biocompatibility as the capability of a medical device (implant) to perform with an appropriate host response in a specific application (Williams, 1986). Bio response of individual materials and biocompatibility of the device are the two key factors that are usually concentrated on while searching for an optimum material for bio implant.

Due to the diversified characteristics required of implant materials, no material is suitable for all biomaterial applications.

Over the years scientists are taking a keen interest in the development of smart implant materials that can correspond well to the different physical, chemical and biological surroundings in which they are pitted. Although metallic implants especially titanium and its alloys are considered as the best fit for a long time now, in recent years polymers are increasingly being used as bio materials. Moreover surface finishing techniques are employed and various coatings are also given over the implant material thereby optimizing it for better efficient clinical results.

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