



Progress on Medical Implant: A Review and Prospects

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Received: 2 June 2022 / Revised: 16 September 2022 / Accepted: 29 September 2022 / Published online: 28 October 2022
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Abstract

Medical implant from different materials such as metals, ceramics, polymers and composites have gained a lot of research attraction due to wide applications in medical industry for treatment, surgical operations and preparing artificial body parts. In this work, we highlight a comprehensive review of medical implant mechanism, various types of implant materials, factors affecting the performance of implant and different characterization techniques. This review provides an overall summary of the state-of-the-art progress on various interesting and promising material-based medical implant. Finally, few new prospects are explained from the established theoretical and experimental results for real-life applications. This study is expected to promote extended interest of scientists and engineers in recent trend of modern biomaterials based medical implant.

Keywords Implant material · Biocompatibility · Hydroxyapatite · Dielectric constant

1 Introduction

As the life span of people in different part of the world increases, the number of age-related diseases is also increasing concurrently. Apparently, there is an urgent need to look beyond the existing medical treatments, medical implants, and prostheses to make the lives of people much more comfortable. Millions of patients have implanted medical devices in their bodies which help the bodies and in turn the person to function normally. These implanted medical devices involve surgical process for their inclusion into the human body ecosystem. The term medical implant refers to an artificial structure or part that is used to replace a fraction of the whole biological structure of the human body. Today, such medical implants can find their applications in orthopedics, cardiovascular stents, pacemakers, defibrillators, neural prosthesis, dentistry or drug delivery system that can enable a normal functionality of several human body parts [1–3]. However, an important step in tissue engineering (medical implant manufacturing) is the selection of an

appropriate biomaterial. The efficient biomaterial has to possess certain essential properties such as biocompatibility [4], bio-inertness [5], biofunctionality [6], and most importantly, the material must not cause inflammation or toxicity to the receptor [7].

The first metal alloy developed specifically to work with human tissues was “vanadium steel” in the early 1900s. Actually, the earliest successful implant was bone plates which were used to stabilize bone fractures and speed up the healing and recovery process for a patient. However, these early metal medical implants did not provide sufficient strength and were prone to corrosion. Later on, the stainless steel was introduced as a medical implant material which provided a far better corrosion resistance. Consequently, stainless steel immediately attracted the interest of majority of the medical fraternity working in this field. It created a great momentum and such metal implants experienced significant efforts towards their development and also in the deployment on several clinical use-cases [8]. Currently, different implant materials such as metals and alloys (e.g., Ti, Co, Ta, Nb, Zr, Mg and Fe based) [9, 10], ceramics (bioinert, bioglass and bioresorbable) [11, 12], carbon nanostructured implants [13], polymers, including polymer meshes and their composites [14–17] are used extensively in medicine. Further, composite material (e.g., zirconia-alumina and carbon fiber-reinforced polymer) [18, 19] is gaining importance as it is formed by combining two or more materials to create a new and superior material. These implant materials are used

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because of their high biological compatibility, variation of physical as well as mechanical properties, resistance to corrosion and other characteristics. Despite the variety, these materials do not provide solutions to all problems because of the limited strength characteristics, integrative properties, complexity of fixation, and biointegration properties. In particular, the high elastic modulus of metal implants is one of the reasons for bone resorption.

Through this review article, we aim to provide useful insights on various implant materials that are increasingly being developed for next-generation medical treatment in order to improve the life standard and life expectancy of people. This review is an attempt to explore the role of dielectric studies (dielectric constant, dielectric loss, and AC conductivity) on implant material as most of the reviews focused only on mechanical properties. Hence, this is the clear research gap exists till today and this review intends to give a complete overview from classification of implant and surface modification to few new interesting prospects from the established theoretical and experimental results for real-life applications.

2 Implant Based on Area of Application

Based on location where the medical implant has to be inducted in the human body, the implants can be classified as: cardiovascular implants, orthopedic implants, and implants for other use [3]. For instance, cardiovascular diseases include a wide range of conditions affecting both the heart and blood vessels. According to WHO, cardiovascular diseases continue to exist one of the prime cause of death all over the world, especially in developed countries. Annually, more than 17.9 million lives are lost due to such diseases [20]. Therefore, cardiovascular implants can contribute greatly in improving quality of human life as well as in reducing the overall cost of treating heart disease [21]. Lately, polymer-coated, and polymer-based cardiovascular implants such as shape memory polymers are being used extensively in modern medicine [22, 23]. Furthermore, with the changing demographics toward an aging society, they will gain even more importance. In addition, we have the orthopedic implants that capture the maximum of the implantable device market segment in terms of value. In India, approximately, 20 million bone fractures report annually [24]. Such market dominance also reflects the increasing prevalence of the sports and exercise injuries that is seen because of lifestyle changes and degenerative musculoskeletal disorders [25]. Apart from the cardiovascular and orthopedic implants, there are a bunch of implants that can be categorized together as other implants. For instance, evolution of cochlear implant technology helped to enhance hearing and speech abilities in a cost-effective way for children [26].

Moreover, development of perimodiolar electrodes, implantable microphones, and rechargeable batteries promise fully implanted devices for such applications in the coming future. Likewise, such implantable devices can also serve as a solution in treating vitreoretinal diseases by enabling intraocular sustained drug release [27]. Various subcategories under these classes are shown in Table 1.

3 Classification of Implant Based on Material

Generally, the material used for making implants and devices are divided into three categories such as metals, ceramics, and polymers. Furthermore, the metallic materials can be classified as either alloys or pure metals; ceramics can be categorized as glasses, glass-ceramics and carbons; and lastly the polymers can be divided into elastomers, thermosets, thermoplastics, and textiles. In addition, composite materials can also be treated as implant material efficiently. [3, 28].

3.1 Metals

Implants used in load-bearing applications are mostly prepared by metal, such as stainless steel, titanium, cobalt-based alloys, and titanium-based alloys. Such load-bearing medical implants can be simple structures such as wires and screws used for fixation of fracture plates to total joint prostheses (artificial joints) for the knees, shoulders, hips, and ankle. In addition, metallic implants are used as dental materials and maxillofacial cardiovascular surgeries [29]. Materials with mechanical properties closer to human bone can be generally used as long-lasting medical implants [30]. Porous material promotes cell growth and also provides stability [31]. Presently, with the rapid progress in field of medical implant, Ti and its alloys are commonly used as orthopedic implant in applications such as repair of joint, femoral fracture and skeleton structure fixation [32, 33]. Ti and Ti alloys are used because of two reasons. First, the elastic modulus of Ti and Ti alloys is almost same as that of the natural human bones, and second, Ti forms a naturally occurring oxide on its surface [34]. Callioglu et al. [35] reported that elastic modulus of Ti alloys are low compared to stainless steels [36] and CoCrMo alloys [37]. High elastic modulus of alloys can cause stress shielding effect which may not be suitable for hard tissue replacement [38]. In addition, density of Ti alloys [39] is almost half the density of stainless steel and CoCr alloys. Ti shows excellent biocompatibility, high strength, and good corrosion resistance. Light weight Ti is treated as a strong alloy [40], offers flexibility and can be coupled with other elements, such as Ag, Al, Ar, Cu, Fe,

Table 1 Implant based on area of application

Cardiovascular implants	Orthopedic implants	Other implants
Pacing devices <ul style="list-style-type: none"> • Cardiac resynchronization therapy devices • Implantable cardioverter-defibrillators • Implantable cardiac pacemakers • Pacing accessories—pacing leads, pacing batteries 	Reconstructive joint replacement <ul style="list-style-type: none"> • Knee replacements • Hip replacement implants • Other reconstructive joint replacement <ul style="list-style-type: none"> – Shoulder implants – Elbow implants – Ankle implants – Joint replacements 	Otolaryngeal implants <ul style="list-style-type: none"> • Cochlear implants • Airway and oesophageal stents • Cosmetic implants—breast implants • Dental implants
Cardiac stents and related implants <ul style="list-style-type: none"> • Coronary stents—drug-eluting stents, bare metal coronary stents • Stent-related implants Synthetic grafts, vascular grafts, peripheral grafts, vena cava filters 	Spinal implants <ul style="list-style-type: none"> • Thoracolumbar implants • Intervertebral spacers • Motion preservation devices • Cervical implants • Implantable spinal stimulators 	Ophthalmic implants <ul style="list-style-type: none"> • Intraocular lenses • Glaucoma and other lenses • Neurostimulators Gastroenterological implants <ul style="list-style-type: none"> • Gastric bands • Biliary stents • Urological implants Gynecological devices <ul style="list-style-type: none"> • Soft tissue repair • Intrauterine devices
Structural cardiac implants <ul style="list-style-type: none"> • Heart valves and accessories • Tissue heart valves • Ventricular assist devices • Implantable heart monitors 	Orthobiologics <ul style="list-style-type: none"> • Hyaluronic acid • Bone substitutes • Bone growth factors • Bone cement Trauma implants <ul style="list-style-type: none"> • Internal fixation devices • Craniomaxillofacial implants • Implantable trauma stimulators 	Drug implants <ul style="list-style-type: none"> • Hormonal implant • Brachytherapy products • Implantable drug pumps

Va, and Zn to generate other alloy forms. For example, most commonly used alloy for medical implant is Ti_6Al_4V [41].

Apart from Ti, magnesium and its alloys are used as implant materials since the first half of twentieth century [42]. Mg^{2+} and its alloys are mostly used in trauma surgery and orthopedics as they are non-toxic in nature inside the human body [43]. These alloys offers an inherent property by which they degrade safely in vivo after the healing process is completed. However, surface activation of magnesium alloys is still sought because the medical implants must be durable enough to allow for bone regrowth [44]. However, Kuhlmann et al. [45] reported that Mg^{2+} alloys corrode rapidly and form subcutaneous gas bubbles, thus limiting their use as implant material. Further, cobalt–chromium–molybdenum alloy (CoCrMo) is commonly used in total hip replacements [46]. CoCrMo has emerged as an alternative to Ti alloys as it offers better corrosion resistance and higher strength [47], although Hart et al. [48] found the existence of Cr^{3+} ions around the tissue surrounding CoCrMo implants. Cobalt in metallic form is also found around the implant, which can cause significant health

consequences in patients. Kop et al. [49] also reported corrosion in CoCrMo neck–stem junction.

However, implant materials with advanced features such as improved tribological properties and low-level costs are required today to meet the growing demand. Additive manufacturing (AM) is one of the techniques to meet these requirements. AM technology allows manufacturing of complex geometric shapes by placing layer upon layer [50, 51]. It is a computer-controlled process that creates three-dimensional objects. Tekdir et al. [52] placed Ti_6Al_4V layers on 316L stainless steel using laser powder bed fusion (LPBF) technique, which is one of the metal additive manufacturing methods. It has been reported that AM technology results in formation of a bimetallic implant material which is cost effective and have superior properties such as higher wear resistance.

3.2 Ceramics

Ceramic materials are formed when clay and their mixture with mineral additives, metal oxides and other inorganic

compounds undergo the process of sintering. Ceramic materials are generally used to repair rigid connective tissues in the body like bones. It is proved that the bone itself can be seen as a combination of organic and ceramic phases, where the ceramic phase is mostly calcium hydroxyapatite (HA) with a value of 1.67 for Ca/P ratio. In addition to the rigid connectivity of tissues, ceramics implants find their use as restorative materials for dentistry implants, such as crowns, cements, and dentures [53]. Bioceramic materials are mainly classified as—bioinert, bioactive and bioresorbable, according to response of the body to implant material.

In the second half of the twentieth century, ceramics started gaining importance in the field of medical implant as it offers better biocompatibility compared to metals [54]. Bioceramics have mainly evolved through three stages. In the beginning of 1960s, first-generation bioceramics (e.g., alumina, zirconia) were designed with the objective of bioinertness in which interaction of implant material with the surrounding tissue is as low as possible [55, 56]. In the manufacturing of prostheses components, alumina is used even today as it provides wear resistance and inertness [57]. Nowadays, zirconia implants is used as a substitute to Ti in implant dentistry because of its white color, high biocompatibility, low thermal conductivity and modulus of elasticity, and low bacterial attraction [58–60]. Wilmowsky et al. [61] mentioned that surface and microstructure modifications of zirconia improve healing of bone and thus can replace Ti implants. However, in the presence of moisture or hydrogen, zirconia losses its favorable properties such as its strength. This is known as low-temperature degradation [62]. The transformation rate is minimized due to reduction in grain size and/or increase in the concentration of stabilizing oxides [63]. In addition, zirconia fracture strength resistance is reduced during its cyclic loading [64]. Tang et al. [65] prepared zirconia-alumina composite using AM technology for dental implants. High toughness of zirconia is combined with with hardness, wear, and chemical resistance of alumina to overcome the drawbacks of zirconia.

In 1980s, second-generation bioceramics (e.g., calcium phosphates, bioactive glasses and glass-ceramics) were developed in which there is favorable interaction between implant material and surrounding tissue resulting in strong mechanical bond. These are called bioactive ceramics. They are mostly used as bone cements or in coating of implant material [66–69]. But due to brittle nature of bioceramics, it was felt that bioceramics alone could not meet the requirements of an implant material. This led to the development of third-generation bioceramics at the end of the twentieth century. These types of bioceramics focused on regeneration of tissues rather than replacement of tissues [70–72]. Second-generation bioceramics with porosity induced in it are studied for this purpose [73, 74]. Further, new types of advanced bioceramics, such as organic–inorganic hybrids

[75, 76], silica mesoporous materials [77, 78], so-called star-gels [79] or mesoporous ordered glasses [80, 81] are also studied. However, ceramic materials have poor fracture toughness which in turn limit their use for load-bearing applications [82].

3.3 Polymers

Both the above-presented classes of metals and ceramics practically lack in terms of versatility. This has attracted a lot of interest to look for the medical implants prepared from polymeric materials. Such polymeric implants due to their versatility can find their applications in wide varieties medical implant use-cases, such as facial prostheses, tracheal tubes, parts of liver and kidney, heart components and even for dentures as well as artificial joints for hip and knee [83, 84]. Polymers which are used as medical implants include poly(tetrafluoroethylene), polyamides (nylons), polysulfones, polyvinylchloride, polyesters, silicone, hydrogels, acrylic, chitosan, and collagen [3]. At present, polyaryletherketones (PAEKs) is among the most promising polymers. Polyetheretherketones (PEEK), a semi-crystalline polymer and poly(aryletherketone etherketoneketone) (PEKEKK), which offers high wear resistance are two most widely used PAEK polymers [2, 3]. Lately, conductive polymers such as polyaniline, polythiophene, polypyrrole and their composites are also gaining importance as they improve biocompatibility and mechanical properties [85]. As composite, conductive polymers also improve corrosion resistance of metallic implants [86]. Fonseca et al. [87] reported the use of bioabsorbable polymers (e.g., PLA, PGLA, and PCL) in cancer treatment. Basak [88] utilizes shape memory polymers such as polyethylene to manufacture medical devices.

3.4 Composites

They consist of both organic and inorganic constituents to imitate natural bone tissue. The organic part of implant material provides flexibility and improves its biocompatibility, whereas load-bearing strength and stiffness are executed by inorganic part. In composites, the organic matrix may be composed of natural polymers (e.g., chitosan, collagen) or synthetic polymers (e.g., polylactic acid (PLA), polycaprolactone (PCL)), whereas the inorganic part may be made of metal alloys and ceramics, such as hydroxyapatite (HA) [3, 89]. Desired mechanical properties such as strength, stiffness, toughness, and fatigue resistance can be obtained through development of composite material. Li et al. [90] fabricated carbon-reinforced PEEK material for orthopedic application. An important advantage offered by composites compared to traditional materials is its ability to design anisotropic properties as strengthening mechanism strongly depends upon the geometry of the reinforcement [3]. We

summarize each of above-mentioned categories of implants Table 2.

4 Surface Modification of Implant Material

When a medical implant is placed in the human body, different molecular interactions take place at the material surface. The types of interactions are affected by the surface properties of the implant material such as surface chemistry, wettability and surface roughness. At the same time, the environment inside the human body is physically and chemically different from ambient conditions which can lead to failure of implant material [91–93]. The major factors which can lead to implant failure are electrochemical degradation, in the form of corrosion, or mechanical degradation, in form of fatigue and wear. These damage mechanisms are explained as follows.

1. **Corrosion:** It is a natural process which results in gradual degradation of material through chemical or electrochemical reaction. Implant material faces severe corrosive environment inside the human body which includes blood, water, sodium, chlorine, proteins, plasma, and amino acids along with mucin in the case of saliva. The equilibrium of the corrosion reactions around the implant is disturbed by these biological molecules [94, 95]. Corrosion is also affected by changes in the pH values. Diseases, accidents, infections and other factors cause the pH value to change from 3 to 9 as opposed to 7, which is normal pH of human body [96]. Furthermore, the internal partial pressure of oxygen is about one quarter of atmospheric oxygen pressure. Hence, lower oxygen level increases the rate of corrosion as formation of protective passive oxide films on the surface of implant material slows down [97].
2. **Fatigue:** Fatigue refers to changes in properties when implant material is subjected to cyclic load [97]. It normally occurs at sites where stress concentrates. Imperfections in the form of impurities, holes and pits are bound to be in material as no material is perfect in the real world. So, when implant material is exposed to external loading, stress concentrates at these sites of imperfection leading to permanent defects in form of dislocations and microcracks. These defects cannot be removed even when unloading takes place and finally implant material suffers from catastrophic failure [98, 99]. Bone plates and orthopedics implants such as artificial hip joints are likely to suffer from fatigue [100].
3. **Wear:** Wear is one of the most common reasons for the failure of metal components. The problem of wear results in implant loosening as friction at joint replacement releases non-compatible metallic ions [101, 102].

In addition, mechanical loading results in corrosion fatigue which further accelerates wear processes. It lead to increase in particle levels over a period of time. These particles are ingested by immune system macrophages, which enhances their destruction. The dying macrophages release metabolites and enzymes resulting in harsh acidic environment in the region around implant material, which further result in erosion of the bone and implant [103, 104].

Thus, high fatigue strength, excellent corrosion and wear resistances are some of the important properties which determine the life span of implant material inside human body [97]. Currently, the most common practice of achieving improvement of implant properties is surface modification. The goal of surface modification is to generate specific chemical and physical surrounding which provide favorable cellular response to implant material. Surface modification of implants is done in order to reduce the problems caused by electrochemical and mechanical degradation. It helps in improving integration of implant with the surrounding tissue. It also decreases bacterial adhesion and inflammatory response. Surface modification affect surface energy, adhesion, biocompatibility, chemical inertness, etc. However, textured surfaces in some cases are counterproductive to the functioning of the device such as articulating surfaces or cardiovascular devices [105]. This review paper puts forward latest findings as shown in Fig. 1 regarding additive and subtractive surface modifications technique in order to improve bone tissue regeneration. One of the most commonly used additive modification method is coating [106].

4.1 Coatings

Coating can improve the surface modification of implant because implants inside a living organism may have to face high degrees of complications. Therefore, it is desired that the implant material must be non-reactive in nature both chemically and physically with the human body and its tissues. Moreover, it has been reported that the infection is one of the major causes of failure of prosthetic implants, so controlling infection inside the human body is a challenge after implantation. As such, functional coatings can be used to alter the substrate's surface attributes such as adhesion, wet ability, corrosion resistance, and wear resistance. Generally, coatings are used on porous metal surfaces for fixation of orthopedic prostheses which include biological and bioactive fixation [107, 108]. A simple mechanism depicting formation of biofilm on medical implant to protect from infection is illustrated in Fig. 2.

Moreover, the coating method can be classified as—physical vapor deposition and chemical vapor deposition.

Table 2 Classification of implant based on material

Categories	Advantages	Pitfalls	Applications
Metals 316L SS; CoCrMo; Ti; Ti ₆ Al ₄ V ; TiNi; Ti ₆ Al ₇ Nb; Au; Ti ₆ Al ₄ VCoCrMo	High strength, low density, high immunity to corrosion, inertness to body environment, enhanced biocompatibility, low modulus and high capacity to join with bone and other tissues	The biological and mechanical biocompatibility of metallic implants requires significant improvement: systemic toxicity of released metal ions due to corrosion, poor wear resistance under loading conditions fatigue, wearing of joint replacements due to movement	Fracture plates, screw, total hip replacement, dentistry casting, loads bearing components in joint replacement, Femoral head, pacemakers
Polymers PTFE; Polyamides (nylons); polysulfones; polyvinylchloride; polyesters; silicone; hydrogels; acrylic	Flexibility, resistance to biochemical effects, greater fatigue strength, good biocompatibility, lower modulus of elasticity, lightweight, variation of physical and mechanical properties by changing polymer composition and carbon content, resistance to radiation and chemical attacks, X-ray transparency	Highly specialized and poorly interchangeable. Biological inertia, low resistance to plastic deformation and the lower fracture load	Soft tissue replacement, vascular prostheses, Sutures, artificial heart components, blood bags, stents, ophthalmology
Ceramics Hydroxyapatite; zirconia; alumina sirolimus; everolimus; tacrolimus; angiopeptin; actinomycin; glasses	Biocompatible, brittle, hard, excellent strength, fracture toughness, stiff, low density, corrosion and wear resistance	Low durability, plastic deformation, brittle, poor tensile strength, prone to cracks and other defects	Dental and orthopedic implants, hip and knee replacements, bone plates, eyeglass, contact lenses
Composites Zirconia–alumina, carbon-reinforced material, HA-PLLA	Excellent strength, stiffness, toughness, fatigue resistance, anisotropic properties and light in weight	Complicate manufacturing, high cost, could be toxic	Dental implant, orthopedic implant, maxillofacial surgery, joint prostheses and bone graft

Fig. 1 Different techniques of surface modifications [106]

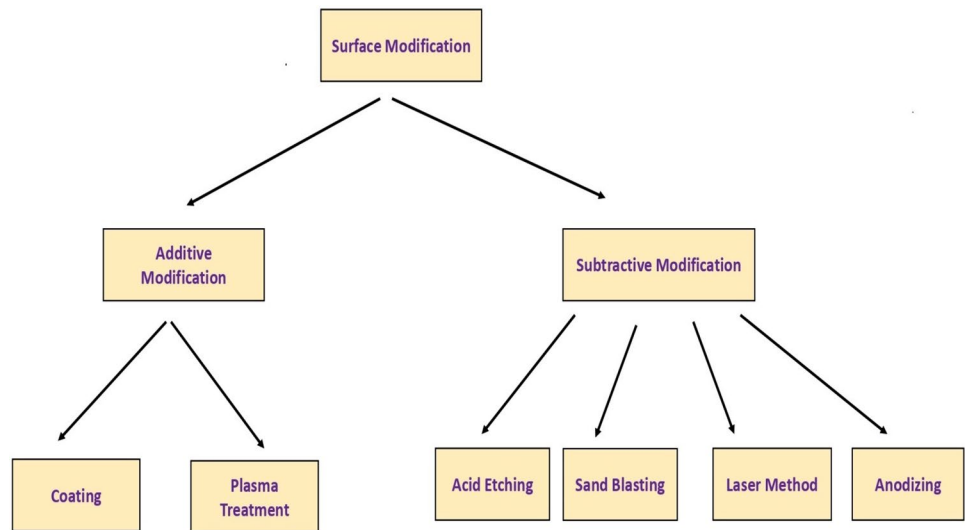
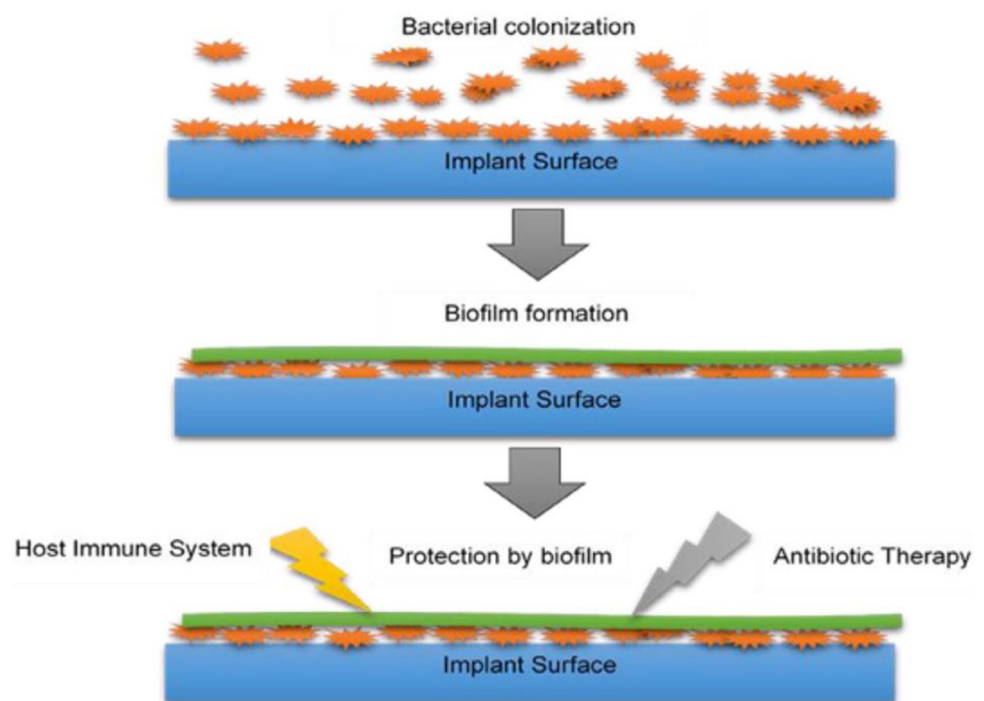


Fig. 2 Mechanism showing formation of biofilm to protect the implant from infection [37]



4.1.1 Physical Vapor Deposition

Substrate surface is deposited with thin film in physical vapor deposition (PVD) method. Film material in vaporized form is condensed and thereafter it is deposited on the target surface. Moreover, coating method is carried out in physical processes such as high-temperature, vacuum, evaporation followed by condensation. Some of the common coatings used with the help of PVD are zirconium nitride, titanium nitride, chromium nitride, and titanium aluminum nitride. PVD coatings ensure excellent purity of the films because of the high-vacuum condition used by evaporation coatings.

PVD coatings are sometimes harder and offer better resistant to corrosion and abrasion. It is also environment friendly than other existing conventional processes of coating, such as electroplating and painting [107–109]. Polymers such as polyethylene and polyvinylidene fluoride (PVDF) are used in PVD method [110, 111]. Some of the PVD coating processes applied in medical implant formation are described as follows.

In plasma-spray process, coating is usually formed by spraying a molten or heat softening material onto a substrate. The powdered material is injected into a very high-temperature plasma flame, which heats up the material quickly and

accelerates in to high speed. This hot material forms coating on substrate surface when cooled [112]. HA-coated metallic implants using plasma-spray method is widely used in orthopedic and dentistry fields since it is reliable, cost effective and protects the implant material surface from corrosion and wear [113]. It has been reported that composition, coating density and crystallinity of coating can be modified by changing process parameters such as spraying distance and particle size [114]. Sun et al. [115] coated HA using plasma-spray method and conclude structural and phase changes with variation in melting of particle, cooling rate and recrystallization. Singh et al. [116] coated HA and 10 wt % $80\text{Al}_2\text{O}_3 - 20\text{TiO}_2$ reinforced HA using plasma-spray method on $\text{Ti}_6\text{Al}_4\text{V}$ substrate and reported that implant properties enhanced when coated with reinforced HA compared to pure HA. Further, corrosion properties of nickel–chromium and chromium oxide is enhanced when plasma-spray method is used [117, 118]. However, defects such as cracks, unmolten particles and uneven thickness lead to poor adhesion strength in plasma-spray method [119].

Pulsed laser deposition (PLD) is a thin-film deposition method via a high-powered pulsed laser beam. Here, a high-power and focused laser beam strikes accurately on a material which is target for deposition inside a vacuum chamber. This material forms thin film on a substrate when deposited in vaporized form. This process can be performed in ultra-high vacuum or in the presence of a background gas such as oxygen, which is commonly used for oxide deposition to completely saturate the deposited film with oxygen [120]. Moreover, an argon atmosphere can also be preferred. Bio-functionality of implant materials is improved by the laser through surface modification [121]. PLD method is used to deposit high-quality hydroxyapatite thin film [122]. Crystallinity, morphology and phase of films deposited by pulsed laser technique depends on parameters such as wavelength of laser, duration of pulse and energy density [123]. Pelin et al. [124] deposited HA films using PLD on titanium and $\text{Ti}_6\text{Al}_4\text{V}$ substrates. They reported that with increase in deposition temperature, surface roughness is improved. Kuppaswamy et al. [125] deposited HA films on polyimide substrates successfully using PLD technique. Nasar [126] evaluated pulsed laser deposition of HA on $\text{Ti}_6\text{Al}_4\text{V}$ for dental applications. But, coating thickness is difficult to control in this technique [127].

In radio-frequency magnetron sputtering electromagnets, permanent magnets or both are used to create magnetic field, so that electrons can be trapped [128]. Sputtering deposit thin films by discharging of material from a source to a substrate. As a result, the ion with very high energy comes out in straight line from the target and impact on the substrate with very high energy [109, 129]. Biocompatible ceramic coating on implant material is deposited using this method [130]. This technique has excellent control over coating

composition, improves adhesion strength and requires low substrate temperature. Xu et al. [131] used radio-frequency magnetron sputtering to deposit HA coating on $\text{Ti}_6\text{Al}_4\text{V}$ substrate and reported biocompatibility of HA with the surrounding tissue. Further, zirconium nitride-reinforced HA enhances the bone strength of the coating. This composite coating is prepared with the help of radio-frequency magnetron sputtering [132]. Socol et al. [133] reported that polymethylmethacrylate (PMMA) substrate is used as cranio-spinal implant material. They deposited HA film using sputtering technique on PMMA substrate and observed compatibility with surrounding bone cells.

4.1.2 Chemical Vapor Deposition

Chemical vapor deposition (CVD) is a chemical process used to produce high-purity, high-performance solid material. Semiconductor industry and composite material frequently used CVD method for thin film production [109]. In CVD, substrates are exposed to one or more volatile precursors that react or decompose on the surface of substrate to form the desired deposit. CVD is highly useful for depositing thin layers of uniform thickness in complex geometries [134, 135]. CVD methods can be categorized as dip coating and spin coating.

In dip coating, the substrate is sequentially immersed in a solution and removed at a constant rate. This gives a layer of the non-volatile species above the drying line. Dip coating is one of the most ideal method for preparing thin layers in chemical solutions [136]. It is an inexpensive, waste-free process that scales easily and provides excellent thickness control. For this reason, it is gaining more and more importance for developing medical implant not only in research laboratories, but also in industrial processes. Dip coating is best suited to impregnate porosity, to prepare nanocomposites, and to do nano-casting [137]. Ritwik et al. [138] synthesized thin film of HA with intermediate layer of shellac. Dip coating is used to deposit film on $\text{Ti}_6\text{Al}_4\text{V}$ substrate and it was reported that the obtained adhesion strength is suitable for the orthopedic application. However, formation of cracks may occur on substrate surface as this method requires high sintering [139].

In spin coating, the substrate rotates around an axis that should be perpendicular to the coverage area. The spin on process has been developed for microelectronics so-called spin on glasses and rotationally symmetric substrates such as optical lenses or lenses for the eye. Moreover, the final thickness of a spin-coated layer will depend on the processing and material parameters such as angular velocity, viscosity and solvent evaporation rate [140]. Different coating techniques that we discussed above have minimal control over coating thickness and are also expensive. However, spin coating is an inexpensive process and shows compatible properties [141].

Spin coating technique is most commonly used for substrates having flat surface. However, it is examined that it can be beneficial for structure having 3D circular shapes such as dental screws and rods [142]. Soule et al. [143] deposited coating of glass on stainless steel 316L using spin coating. They reported enhanced bioactivity, improved antibacterial properties. In addition, it promotes regeneration of tissues around implant materials. Simple comparison between different coating method is depicted in Table 3.

4.2 Nanotechnology-Derived Surface Modification

However, emergence of nanotechnology led to the development of new techniques and methods of surface modification. Nanotechnology refers to materials in which one of their dimension is less than 100 nanometer [144]. It has been reported that osteoconductive and osteointegration properties of nano-modified implant material is improved as it offers large surface area to volume ratio [145]. As compared to conventional materials, nanocrystalline structure is harder, smoother and acts as good binder. Since nano materials have excellent wear resistance, it also reduces the problem of implant loosening [146]. This approach increases the implant material mechanical strength by imitating the composition of bone. Nanoparticles of hydroxyapatite, calcium triphosphate, bioactive glass, synthetic polymers and biodegradable polymers have been manufactured into three-dimensional porous scaffolds for tissue regeneration and bone repair [146–148]. Arumugam et al. [149] reported that more successful orthopedic implant materials can be developed by doping carbon nanotubes (CNTs) in small amount in polymer or ceramic matrices. High flexibility, high tensile strength and low density of CNTs make them an attractive material in field of nanotechnology. However, nanomaterial also possesses some threat and harmful effects as they can cause inflammatory response by depositing in the respiratory tract. As they are small in size, they can be transported easily in various parts of the body including heart and brain. The physical and chemical properties of nanoparticles including its shape, size and surface chemistry are also known to increase inflammation and tissue damage [146].

5 Characterization Techniques of Implant Material

During the last few years, development of implant material has gained momentum. In order to be used over a long period of time, quality and reproducibility of implant materials are essential. Further, characterization of physical property, chemical property and dielectric studies are necessary for increased acceptance and use of implant materials. These

Table 3 Comparison of different coating method

Method	Agents	Advantages	Pitfalls
Dip coating	HA, PTFE, forsterite, PLA, titania nanotube	Low cost, coating on complex shape and fast process	High temperature reduces adhesion strength, appearance of cracks and mismatch in thermal expansion
Plasma spray	HA, Ag, Sr, Mg, Cu, Ce.	High deposition rates, low cost, faster healing of bone and chance for coating degradation is less	Poor adhesion, HA layer may be decomposed due to high temperature, nonuniform coating density and transformation of phase
Sputtering deposition	Ag, Cu, Zn, HA.	On flat substrate forms dense coatings and with uniform thickness, good adhesion and homogenous coating	Low deposition rate, high cost, line of sight technique and low crystallinity increases rate of dissolution of film
Spin coating	PLLA, PCL, PGA, ZnO, HA	Substrates with complex shapes are suited, excellent control over coating thickness, inexpensive and requires low temperature	Not suitable for large substrate and lack of material efficiency.
Pulsed laser deposition	HA, Ag, carbon, lignin	Crystalline and highly adherent coatings, restore complex stoichiometry, excellent control on deposition parameters and forms porous coating	Line of sight technique, metal or alloy substrate may oxidize due to process temperature splashing or particle deposition and lacks uniformity

characterization techniques including dielectric studies are discussed briefly as follows.

5.1 X-Ray Diffraction (XRD)

X-ray diffraction (XRD) analysis is a method employed in the field of materials science subject to characterize the implant material crystallographic structure. For XRD analysis, target implant material is irradiated via incident X-rays and then measurement is performed on the basis of the intensities and scattering angles of the X-rays that are leaving the material [150]. It is based on Bragg's law:

$$2d \sin \theta = n\lambda, \quad (1)$$

where d denotes the interplanar spacing of the crystal, θ is angle of incidence, n is an integer which represent the order of reflection, and λ is wavelength of the incident X-rays. Nayak et al. [151] observed the crystallographic structure of gallium-doped hydroxyapatite and pure hydroxyapatite with help of XRD pattern as illustrated in Fig. 3.

5.2 Scanning Electron Microscope (SEM)

The surface morphology, particle size and chemical compositional analysis of implant material at different magnification were examined with scanning electron microscope [152]. SEM images of pure hydroxyapatite and strontium-doped hydroxyapatite [152, 153] are displayed in Fig. 4.

5.3 Vector Network Analyzer (VNA)

A VNA is a versatile device which reflects the electrical characteristics of any medical implant. It is normally used to measure transmission parameters, reflection parameters,

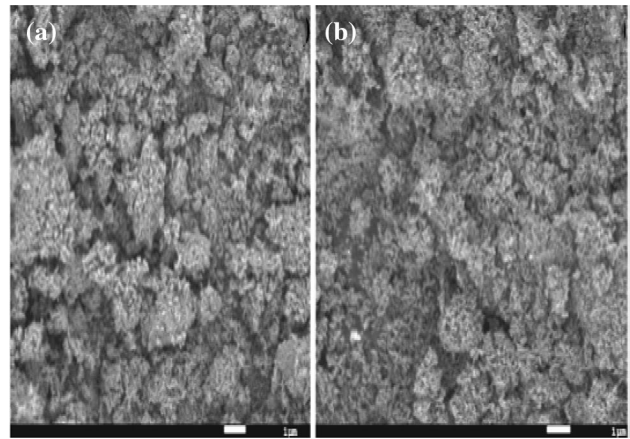


Fig. 4 SEM image. (a) Pure HA. (b) Sr-doped HA [153]

impedance parameters, scattering parameters, dielectric constant, dielectric loss, conductivity and loss tangent [154].

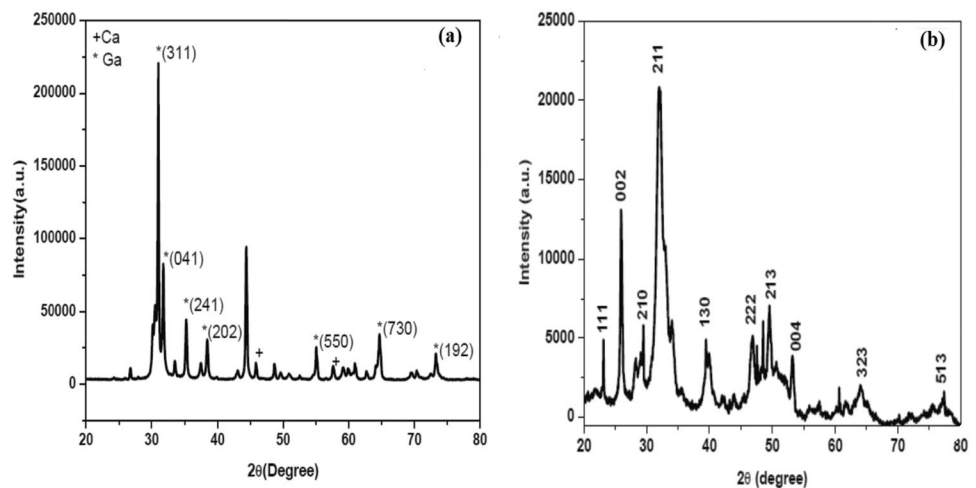
5.3.1 Dielectric Permittivity/Dielectric Constant (ϵ')

Variation of dielectric constant with frequency is significantly observed for different medical implant materials in Fig. 5. Dispersive region is observed in dielectric constant variation graph. The nature of dielectric constant exclusively dependent on implant material property [1, 151] and frequency band used.

5.3.2 Dielectric Loss (ϵ'')

Variation of dielectric loss or imaginary component of high-frequency complex permittivity with frequency is very important because it clearly signifies the lossy behavior [153, 155] of implant material which is shown in Fig. 6.

Fig. 3 XRD pattern. (a) HA doped with Ga. (b) Pure HA [151]



5.3.3 AC Conductivity (σ_{ac})

AC conductivity variation with frequency indicates the conducting nature and loss factor associated with implant material [151, 155] as shown in Fig. 7.

6 Current Useful Implant Materials

6.1 Hydroxyapatite (HA)

Biocompatibility is one of the most important property of a biomaterial used in medical implant. Due to injury when bone is lost, that lost bone can be filled by natural bone. But that natural bone may not be biocompatible with the surrounding tissue and can cause possible infection. HA, $Ca_{10}(PO_4)_6(OH)_2$ can be used as an alternative to natural

bone as its composition is similar to bone. Bioactivity and osteoconductive properties of HA made them suitable for bone regeneration and in improving osteointegration of implant. Thus, HA is ideal candidate for bone tissue engineering scaffolds [156]. However, HA will not be able to integrate with bone and confine within fibrous tissue if the gap between HA and bone is wide [157]. In addition, it is not suitable for major load bearing areas as it is brittle and have poor mechanical properties [1]. Therefore, HA is doped with various ions such as Zn^{2+} , Sr^{2+} , Ce^{3+} , Bi^{3+} , and Y^{3+} in order to improve its properties. Ca^{2+} ions in the HA structure is substituted by these ions, thus affecting the crystallinity, lattice parameters and morphology of HA. Hence, through these ionic substitution, characteristic properties of HA can be controlled as crystal structure and composition of HA is directly linked with its properties [158].

Fig. 5 Dielectric constant versus frequency plot of (a) BT in (HA-BT)/PVDF and (b) Ga-doped HA [1, 151], respectively

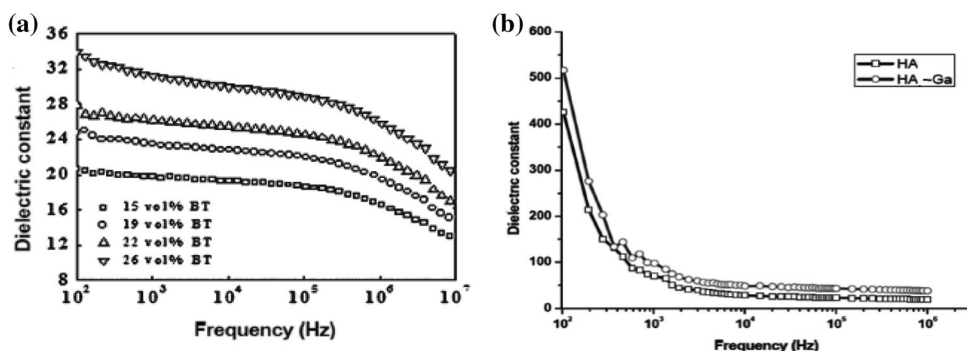


Fig. 6 Dielectric loss versus frequency plot of (a) Sr-doped HA and (b) Y-doped HA [153, 155], respectively

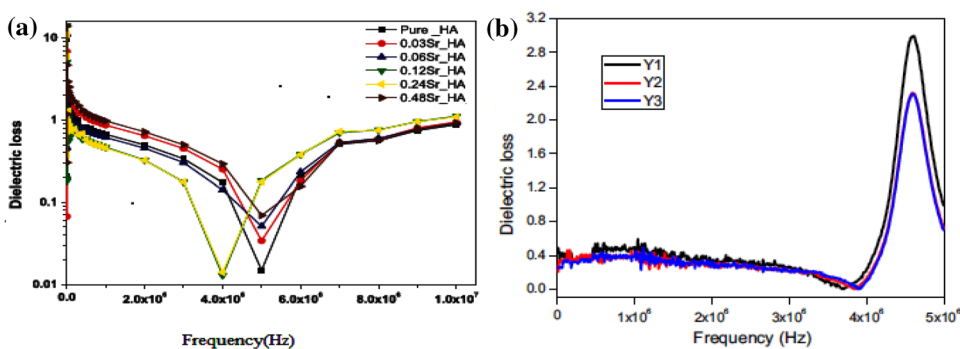
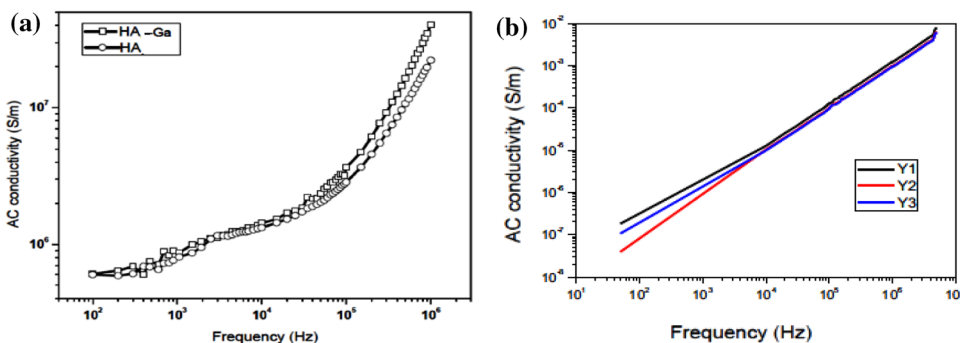


Fig. 7 AC conductivity versus frequency plot of (a) Ga-doped HA and (b) Y-doped HA [151, 155], respectively



Bioactivity and osteointegration of HA improves both *in vitro* and *in vivo* under the influence of electrical polarization. It has been reported that a surface with net charge (positive or negative) is more hydrophilic, further easing interfacial processes and protein adsorption as compared to neutral surface [157, 159]. It is also shown that bone growth is faster on negative charge surface [160]. It may be due to adsorption of Ca^{2+} ions. At the bone implant interface, surface charge also shows to increase early-stage bone mineralization [160]. Thus, surface charge modification provides a new direction in order to improve biological properties and performance of bone scaffold materials as shown in Fig. 8.

Direct current excitation can also accelerate bone healing process. Electroactive materials with high dielectric permittivity under the influence of external electric field results in formation of significant surface charge. High-dielectric permittivity makes material strongly polarized and thus shorten the bone regeneration time [1, 161]. The change in properties of HA doped with various ions was investigated by several researchers [162].

6.2 HA Doped with Different Materials

Yttrium (Y) use in medical applications include treatment of hepatocellular carcinoma [163, 164]. It has been reported that HA doped with Y using hydrothermal method not only helps in accelerating the bone growth but also restricts the oral bacterial growth [165]. Kaygili et al. [155] studied different doped HA samples with 0, 2 and 4 wt % of Y. Since Y^{3+} ions are smaller than Ca^{2+} ions, incorporation of Y cause size reduction. It was observed that change in Y content affect crystallinity degree and lattice parameters of the sample. Volume of unit cell decreases gradually as Y is added to the sample [166]. Electrical properties such as dielectric constant (ϵ'), dielectric loss (ϵ'') and AC conductivity

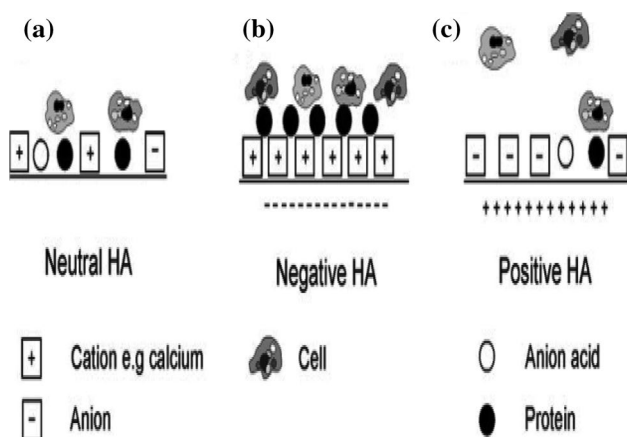


Fig. 8 Diagram showing the interaction between polarized charge and the material. (a) On neutral material. (b) On negative material. (c) On positive material [159]

(σ_{ac}) also vary with Y content as well as with frequency [167]. The dielectric loss (ϵ'') decreases gradually as content of Y is increased. The AC conductivity (σ_{ac}) increases linearly as frequency is increased but decreases with increasing content of Y. Composite showed insulator behavior as measured resistance values were in range of 1011–1012 Ω . However, addition of Y did not show any improvement in antimicrobial properties [168].

Bismuth (Bi) containing nanoparticles have recently gained a lot of research attraction in biomedical applications, because of their excellent properties, which include high stability, high surface area and strong diamagnetism [169]. Al-Hazmi [170] prepared $\text{Ca}_{(10-x)}\text{Bi}_x(\text{PO}_4)_6(\text{OH})_2$ ceramics using sol-gel technique and study the effect of Bi on dielectric and ferroelectric properties of composite. HA samples were doped with 0.1, 0.2, 0.3, 0.4 and 0.5 wt % of Bi. FTIR spectra showed that OH is substituted with Bi as Bi content increases [171]. Scanning electron microscopy (SEM) images indicate that the particle size of $\text{Ca}_{(10-x)}\text{Bi}_x(\text{PO}_4)_6(\text{OH})_2$ ceramics lies in nano-region and are randomly distributed. X-ray diffraction (XRD) technique further shows blending of Bi into HA. Bi dopant causes dielectric constant of the ceramics initially to increase and then as Bi content increases, dielectric constant decreases significantly. The reason behind this change in dielectric constant is electrical polarization in the ceramics [172]. AC conductivity shows linear behavior with frequency. It is also reported that due to introduction of Bi dopant, ferroelectric properties of the composite decreases. Further, Ahmed et al. [173] studied the antibacterial properties of Bi and reported that the antibacterial property enhanced when Bi is added with HA.

Strontium (Sr) is mainly found in human bone. It helps in treatment of osteoporosis, bone growth and development [174]. Reem et al. [153] used hydrothermal method to synthesize HA-doped Sr. Sr is added in different concentration, i.e., 0, 0.03, 0.06, 0.12, 0.24 and 0.48 in HA. It was reported that crystallinity value reduces since Ca^{2+} ions are being substituted by Sr^{2+} . XRD pattern of Sr-HA composite shows similar peaks with reduced intensity compared to pure HA [175]. Dielectric constant (ϵ') at lower frequency do not show much change while it increases with increase in frequency at higher frequency. In addition, dielectric constant increases as concentration of Sr increases in sample, mainly due to ionic polarization [176]. AC conductivity increases with increase in frequency [177].

Bowen et al. [178] synthesized HA-BaTiO₃ composite with BaTiO₃ (BT) composition varying from 0% to 100% by volume using pressure less sintering method. Effect of BaTiO₃ on electrical properties of the composite were studied and reported that with decreasing BaTiO₃ concentration, magnitude of dielectric constant as well as degree of dispersion decreases [179]. AC conductivity increases as

concentration of BaTiO₃ increases since BaTiO₃ has higher permittivity compared to HA. For higher frequency, AC conductivity increases almost linearly, while at low frequency, there is no dependence between them [180]. Dang et al. [1] prepared three-phase bio-electroactive nanocomposite and used PVDF with HA – BaTiO₃ composite material. BaTiO₃ nanoparticles which provide high dielectric permittivity and excellent bioactivity of HA nanoparticles is doped with PVDF to form three-phase bio-electroactive nanocomposites based ferroelectric polymer matrix. It is proved that three-phase (BT-HA)/PVDF nanocomposites exhibits higher dielectric permittivity which results in formation of surface charges due to an applied electrical field. Surface charge reduces bone regeneration time [1]. Three-phase (BT-HA)/PVDF nanocomposites can also be used as bio-dielectric materials since it has low dielectric loss [181]. However, molecular movement of PVDF is inhibited as volume fraction of BT increases because it would result in smaller areas of bulk PVDF phase which in turn result in relatively lower degree of crystallization of PVDF in three-phase nanocomposite.

Gallium (Ga) is used in treatment of osteoporosis and hypercalcemia [182]. It is an important component of composite implant material and shows antibacterial as well as antimicrobial properties when used with phosphate-based glasses (PBGs) [183]. It also helps in controlling drug delivery related issues [184]. HA doped with Ga bioceramics find application in bone formation and growth. Crystalline size increases as Ga³⁺ ion substitute Ca²⁺ ion in HA [185]. Nayak et al. [151] reported that Ga-doped HA-based bioceramics forms negative surface charge and HA–Ga experiences lower dielectric loss compared to pure HA establishes the efficiency of HA–Ga as a dielectric material. Further, high electrical polarization in HA–Ga bioceramics results in high value of dielectric constant. However, AC conductivity of HA–Ga is not changed significantly compared to pure HA.

Iron (Fe) is an essential part of human body and present in small amount in bones and teeth. Biomedical properties of HA enhance with addition of small amount of iron as it promote bone growth and had no toxic effect on the osteoblast cells [186]. Kaygili et al. [187] doped hydroxyapatite (HA) with Fe in different concentrations using sol–gel process and their properties were investigated. Crystalline size of the samples decreases gradually as Fe content increases thus affecting crystallinity degree, lattice parameters, unit cell volume and the phase composition [188]. As concentration of Fe increases, the dielectric permittivity of the HA increases for all frequency values. The value of AC conductivity increases as frequency increases and it was also reported that HA doped with Fe showed insulator behavior [189].

Chromium (Cr) improves the ability of insulin to convert glucose thus help cells in human body to gain energy

and maintain blood sugar level. Ibrahim et al. [190] doped hydroxyapatite with 0.5, 1.0, 2.0, and 3.0 wt % of chromium using wet precipitation method and studied dielectric properties. It has been reported that chromium improves biocommunication as it causes electrical conductivity to increase [191]. At low frequency, dielectric permittivity increases with increase in wt % of Cr. However, dielectric permittivity at higher frequency remains almost same for pure and doped ones [192]. Thus, HA bioconductivity and electrical properties improves with addition of Cr.

6.3 Piezoelectric-Based Biomaterials

Piezoelectric materials are smart materials as these materials convert the applied mechanical pressure into electrical signals which is called as direct piezoelectric effect. Since piezoelectric effect is a reversible process, these materials also convert electrical signals into mechanical signals [193]. There are two basic requirement for any material to show piezoelectricity. First, crystal lattice structure of the material and second, there should be lack of a center of symmetry [194]. Piezoelectric materials find application in tissue engineering such as in tissue repair and regeneration and they do not require any external power source to generate electrical signal as they are active transducer [195, 196]. Piezoelectric material produces electrical signal which can promote tissue formation by definite pathways and produce suitable bioelectrical signals like natural extracellular matrix (ECM), which is observed during remodeling phenomenon in bone and cartilage [197]. Tissues such as bone, cartilage, and tendon show direct piezoelectricity [198]. Collagen, one of the main constituent in bone and cartilage also shows piezoelectric property [199]. Piezoelectric materials can be used alone as well as a composite with other suitable material. It can either be polymers or ceramics.

6.3.1 Piezoelectric Polymers

Polymers are light in weight and are mechanically as well as electrically tough as compared to organic and inorganic materials. They are flexible and show excellent compatibility with other biomaterials [200]. Polymers possess piezoelectric property must have permanent dipole, the ability to sustain alignment and undergo large strains when mechanical stress is applied [201]. Some interesting piezoelectric polymer widely used as implant material are as follows:

Chitosan is biocompatible, biodegradable and non-toxic. It can be processed and manufactured in various forms such as fibers and films and thus have the ability to mould into the shape of the required tissue. As chitosan is polycationic in nature, it promotes bone growth [201, 202]. It is also used in regenerative medicine. Petrov et al. [203] synthesized chitosan–hydroxyapatite composite material and studied

electrical properties. It has been reported that dielectric permittivity increases with increase in frequency. Mechanical properties such as tensile modulus and strength is improved when CNT is added in small amount in chitosan as CNT have excellent mechanical properties [204]. However, chitosan degrades at higher temperature.

Piezoelectric coefficient of poly(3-hydroxybutyrate-co-3-hydroxyvalerate) (PHBV) is similar to human bone and is biocompatible as well as biodegradable [205]. It has been reported that composite material of PHBV/PCL/HA accelerate bone regeneration as PCL enhances surface roughness and hydrophilicity while HA improves bioactivity [206]. Wu et al. [207] fabricated a composite mixture of PHBV/Bioglass which can be used to prepare better engineered cartilage as compared to pure PHBV. But it has been reported that enzymatic degradation mechanism can degrade PHBV and thus failed to form strong interaction with the surrounding tissue [208].

Polyvinylidene fluoride (PVDF) has high flexibility and its non-toxic nature finds application in tissue engineering [209]. P(VDF-TrFE), which is a copolymer of vinylidene fluoride (VDF) and trifluoroethylene (TrFE) have positive effect on cell adhesion and cell growth. When mixed with natural polymers such as starch or cellulose, it finds application in tissue repair and regeneration as scaffold structure [210, 211]. The composite of PVDF-TrFE and barium titanate piezoelectric promotes bone regeneration [212]. However, PVDF is not suitable for biological environment as it tends to degrade in extreme alkaline medium [213]. Non-biodegradable nature of PVDF limits its use in tissue engineering [214].

Poly-L-lactic acid (PLLA) is biocompatible and biodegradable polymer. High value of piezoelectric coefficient and excellent mechanical properties of PLLA finds application in orthopedics such as screws and pins. Since it is degradable, it has been reported for rapid bone regeneration [215]. Chorsi et al. [216] have reported that due to piezoelectric polarization, PLLA promote bone growth. Prabhakaran et al. [217] fabricated composite material of PLLA/collagen and HA and reported improved bone growth compared to PLLA. Further, PLLA/chitosan composite material is used in tissue engineering as chitosan material significantly neutralizes acidity caused by PLLA degradation [218].

6.3.2 Natural Polymers

Natural polymers have gained a lot of research attraction recently compared to synthetic polymers because of their biocompatible and biodegradable behavior. They consist of highly organized structure and provide biological signaling, thus promote cell adhesion and remodeling [219]. Low toxicity and degradable nature of natural polymer further helps them to find application in tissue engineering [220].

However, inadequate physical properties and loss of biological properties during formulations restrict their use in the field of medical implant [221]. Some significant natural polymers are listed as follows.

Cellulose offers excellent biocompatibility and high tensile strength. It has excellent mechanical properties despite high water content. It finds application in bone and cartilage tissue engineering as it promote cell adhesion [222]. However, small pore size of cellulose limits the cell infiltration which can be improved by addition of paraffin microspheres as it results in better pore connectivity and also increases the pore size [223]. Daugela et al. [224] prepared cellulose/HA scaffolds which promote bone formation as highly interconnected porous structure is formed. Eftekhari et al. [225] fabricated a composite material consisting of cellulose/PLLA/HA in which reinforcing agents are homogeneously dispersed in PLLA matrix, thus improving mechanical strength of composite material.

Collagen is biocompatible, hydrophilic and has good cell binding properties. It is also used in bone healing and provides structural and mechanical support [226]. It can be treated as a composite material with hydroxyapatite and thus promotes cell growth. However, due to high ionic forces of hydroxyapatite there is slight decrease in piezoelectricity of collagen-HA composite [227]. Yilmaz et al. [228] studied HA/graphene oxide/collagen composite mixture and reported increase in mechanical strength. It has also been reported that penetration of tissue increases with increase in pore size. But rapid degradation and low mechanical stiffness of collagen limits its application in tissue engineering.

Chitin is a natural polysaccharide exhibiting biocompatible and hydrophilic nature. It promotes cell adhesion. Chitin as a composite material with other suitable biomaterial find application in bone and cartilage regeneration [220]. Chang et al. [229] reported that mechanical properties and biocompatibility improved when HA was incorporated with chitin hydrogel network. Chakravarty et al. [230] fabricated a composite mixture of chitin, polylactic acid and hydroxyapatite and found that it acts as excellent material for bone regeneration. However, it is not suitable for hard tissue application as it fails to maintain predefined shape and suffers from poor mechanical properties [231].

6.3.3 Piezoceramics

Piezoceramics such as boron nitride, zinc oxide, and potassium sodium niobate have very high piezoelectric coefficient. Ciofani et al. [232] reported that piezoelectric property of boron nitride nanotube is superior compared to piezoelectric polymers. In addition, it provides excellent mechanical properties and suitable for orthopedic applications as it increases the adhesion of osteoblast cells [233]. Tayel et al. [234] evaluated the antibacterial property of ZnO

nanoparticles and reported that it enhances the antibacterial property of implant material. Ito et al. [235] reported that composite mixture of HA–ZnO enhances bone growth. It has also been reported that the doping of ZnO in composites can increase their flexural strength, decrease the shear bond strength, and decrease the compressive strength [236].

However, piezoceramic material suffers from toxicity, which has limited their use in tissue engineering [237]. Mechanical properties (young modulus (Y) and tensile strength (σ)) and electrical properties (dielectric constant (ϵ') and AC conductivity (σ_{ac})) of some of the engineering materials which are used as implant material are shown in Table 4.

Table 4 Mechanical and electrical properties of different implant material

Materials	Mechanical properties	Electrical properties	References
Hydroxyapatite	• $Y = 100$ • $\sigma = 40$	• $\epsilon' = 15$ • $\sigma_{ac} = 10^7$	[151, 238]
Gallium	• $Y = NA$ • $\sigma = 150$	• $\epsilon' = 20$ • $\sigma_{ac} = 10^7$	[151, 239]
Yttrium	• $Y = 50-60$ • $\sigma = NA$	• $\epsilon' = 10$ • $\sigma_{ac} = 10^{-3}$	[155, 240]
Strontium	• $Y = 17.2$ • $\sigma = 77$	• $\epsilon' = 16$ • $\sigma_{ac} = 3 * 10^{-4}$	[153, 241]
Iron	• $Y = 210$ • $\sigma = 320$	• $\epsilon' = 10.5$ • $\sigma_{ac} = 10^{-3}$	[187, 242]
Bismuth	• $Y = 31.7$ • $\sigma = NA$	• $\epsilon' = 9$ • $\sigma_{ac} = 0.5 * 10^{-3}$	[170, 239]
Barium titanate	• $Y = 99$ • $\sigma = NA$	• $\epsilon' = 700$ • $\sigma_{ac} = 10^{-4}$	[178, 243]
Chitosan	• $Y = 1.08$ • $\sigma = 37.7$	• $\epsilon' = 2.3$ • $\sigma_{ac} = NA$	[203, 244]
Collagen	• $Y = 0.1-0.36$ • $\sigma = NA$	• $\epsilon' = 31.62$ • $\sigma_{ac} = 0.03-0.08$	[245, 246]
Forsterite	• $Y = 77$ • $\sigma = 152-253$	• $\epsilon' = 8$ • $\sigma_{ac} = NA$	[247, 248]
Alumina	• $Y = 420$ • $\sigma = 282-551$	• $\epsilon' = 11$ • $\sigma_{ac} = 10^{-6}$	[249, 250]
Zirconium	• $Y = 92$ • $\sigma = 388$	• $\epsilon' = 24$ • $\sigma_{ac} = NA$	[251–253]
Stainless steel	• $Y = 200$ • $\sigma = 480-1350$	• $\epsilon' = 1NA$ • $\sigma_{ac} = NA$	[254–256]
CoCrMo alloys	• $Y = 210-240$ • $\sigma = 800-1800$	• $\epsilon' = NA$ • $\sigma_{ac} = NA$	[254–256]
Ti alloys (Ti_6Al_4V)	• $Y = 110$ • $\sigma = 1050-1100$	• $\epsilon' = NA$ • $\sigma_{ac} = NA$	[254–256]
PMMA	• $Y = 1.8-3.1$ • $\sigma = 30-48$	• $\epsilon' = 3$ • $\sigma_{ac} = 10^{-6}$	[257, 258]
PVDF	• $Y = 2.1-2.9$ • $\sigma = 50-57$	• $\epsilon' = 10$ • $\sigma_{ac} = 10^{-6}$	[257, 259]
PTFE	• $Y = 0.4-0.75$ • $\sigma = 25-36$	• $\epsilon' = 2.1$ • $\sigma_{ac} = NA$	[257, 260]
PDMS	• $Y = 0.3-0.87$ • $\sigma = 2.24$	• $\epsilon' = 3-4$ • $\sigma_{ac} = 10^2$	[257, 261]
PLLA	• $Y = 1.1$ • $\sigma = 28-50$	• $\epsilon' = 2$ • $\sigma_{ac} = NA$	[262–264]

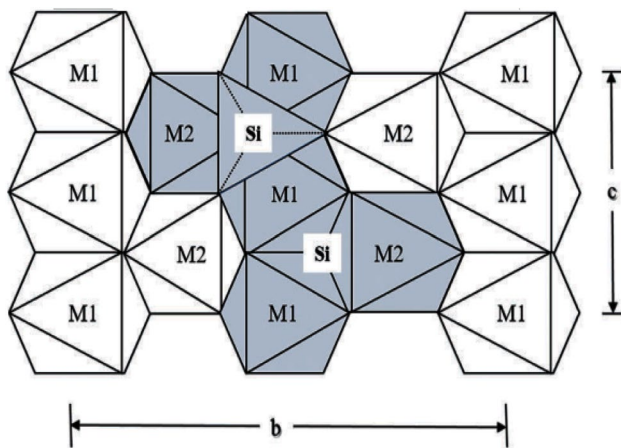


Fig. 9 Forsterite crystal structure [269]

7 Potential Materials for Future Studies

As hydroxyapatite chemical structure and composition is similar to bone, most of the research regarding biomedical implant revolves around it. It can also be used as a composite material. But, hydroxyapatite suffers from basic problem such as it is not suitable for load-bearing application due to poor mechanical properties. It also suffers from high dissolution in a physiological environment, thus affecting long-term stability which can cause implant failure [265, 266]. Forsterite (Mg₂SiO₄), which has an orthorhombic-dipyramidal crystal structure can be used as an alternative to hydroxyapatite since it provides good adhesion strength and osteointegration [267, 268]. Forsterite crystal structure [269] is shown in Fig. 9.

Previous studies showed that magnesium keeps bone strong and is one of the essential components in the human body. It helps in maintaining blood glucose level and also assist in production of protein [270]. Silicon also plays an important role in bone development and maintenance. It has also been reported that silicon is mainly found in areas where new bone formation takes place [271]. Forsterite attracted a lot of interest due to better biocompatibility and good mechanical properties compared to HA [272]. Kichi et al. [273] prepared polycaprolactone/gelatin/forsterite nanocomposite. The Non-toxic and biodegradable properties of polycaprolactone allow them to show resistivity in bio fluids and it is also suitable for load-bearing applications. The hydrophobic nature results in reduced corrosion rate of implant materials [274, 275]. Gelatin, a natural polymer is obtained from hydrolysis of collagen, which in turn is extracted from skin and bone of animals. Moreover, Gelatin can be used as a composite material as it contains functional groups such as COOH which can form crosslinking with other biomaterial. It also makes the material more stable with good mechanical

properties when mixed with other polymers [276]. It has been reported that as percentage of polycaprolactone in the coating increases, the corrosion resistance increases. In addition, cell growth of nanocomposite structure increases by decreasing the amount of gelatin [277]. Jeffery et al. [278] synthesized manganese oxide-doped forsterite powder. It has been reported that the hardness and fracture toughness increases up to certain temperature and then it starts decreasing. Naga et al. [279] studied the effect of forsterite addition in hydroxyapatite and concluded that in order to improve the physical and mechanical properties of the produced composite, maximum 20% of forsterite can be added.

Steatite can be used as another alternative to hydroxyapatite. Steatite (MgSiO₃) is a magnesium silicate composite material. It offers good electrical properties and high mechanical strength [280]. Due to its good biocompatibility, it is used as a implant material in artificial bone. Further, it can also be used as dental implant due to high hardness and bending strength [281].

8 Conclusions

In this review paper, we discussed about the role played by implant materials in improving the life standard and life expectancy of people. Implants can be classified based on material and area of application. Different coating technique was discussed in detail, and it was shown that surface modification of biomaterials through coating improves corrosion and other (mechanical, electrical, biological) properties of implant materials. This review attempts to show that HA properties can be improved by dopant ions such as Zn²⁺, Sr²⁺, and Y³⁺ and consequently bone regeneration time was reduced under the influence of electrical polarization and negative surface charge. Information is also provided on forsterite, steatite which could be used as an appropriate alternative to HA because of their better mechanical properties. However, electrical properties of forsterite and how it affect bone regeneration are yet to be studied in detail.

Funding The authors did not receive support from any organization for the submitted work.

Availability of Data and Materials Data sharing not applicable to this article as no datasets were generated or analyzed during the current study.

Declarations

Conflict of Interest On behalf of all the authors, the corresponding author states that there is no conflict of interest.

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