J. Electrochem. Sci. Eng. **13(1)** (2023) 173-192; http://dx.doi.org/10.5599/jese.1604

Open Access : : ISSN 1847-9286
www.jESE-online.org

Review paper

Electrochemical behaviour and biocompatibility of claddings developed using microwave route

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Received: November 17, 2022; Accepted: December 12, 2022; Published: December 30, 2022

Abstract

In recent years many different biomedical implants have been created for prolonged usage within the human body. The number of these implants has been steadily expanding. Mechanical characteristics of biomaterials, such as elastic modulus, hardness, tensile strength, and scratch resistance, are essential for implants. Biomechanical incompatibility is associated with implant fracture brought on by mechanical failure. The materials utilized to replace bone must have mechanical qualities comparable to those of bone. Metallic implants deteriorate due to wear, electrochemical breakdown, or a synergistic mix of the two. Biocompatible materials are used to repair or replace joints, fractured, or otherwise damaged bone. Corrosion is the main factor in hip implant failure. These characteristics also contain several other factors, such as solution factors, geometric factors, metallurgical factors, and mechanical factors. The mechanical properties of the implant materials were most important and had a considerable impact on the process of bone restoration. Metals have the highest tensile strength compared to other materials, followed by polymers and ceramics (except for zirconia). There are several issues with the metallic biomaterial that need to be fixed, including the release of harmful substances during metallic corrosion.

Keywords

Metallic implants; mechanical characteristics; corrosion resistance; hydroxyapatite

Introduction

In the modern world, millions of individuals experience a wide range of diseases and injuries, including tissue damage, cancer, and problems with the teeth, hips, and eyes, among others [1]. Additionally, road accidents kill about 1.3 million people annually and damage another 20-50 million people. The majority of traffic accidents worldwide occur in India [2]. In India, road accidents are a major source of mortality and illness. Additionally, there has been an increase in vehicle accidents in India over the past 20 years, with six million non-hospitalized visits and 1.2 million hospital-lizations. In addition, millions of individuals worldwide suffer from musculoskeletal ailments, including osteoarthritis, bone cancer, and spine issues [3].

The Indian Osteoporosis Society predicts that by 2013, there will be 36 million osteoporosis sufferers in India, up from an estimated 26 million in 2003. In India, it is estimated that there are 3 million hip fractures, 20 lacs broken wrists, 5 million spinal fractures, and another 300,000 bone fractures each year. By 2050, it is predicted that osteoporosis will cost the global economy $131 billion [4]. Every year, orthopedic-related issues cost society more than $250 billion. The American Academy of Orthopedic Surgeons (AAOS) classified musculoskeletal conditions as the leading cause of patient visits to the doctor due to their significant impact. Therefore, the years 2000 to 2010 have been designated as the global "decade of the bone and joint" in an effort to raise awareness of musculoskeletal illnesses via the prevention, research, and education. The techniques to address such problems have been a focus of concern for the modern healthcare industry due to the prevalence of musculoskeletal ailments and the lack of a cure. Another potential strategy for treating these problems is the use of artificial biomaterials. Such synthetic biomaterials are implanted clinically to restore often physically injured tissues [5]. The market for biomaterials is thought to be worth more than $300 billion globally and is predicted to grow by 20 % annually. Millions of people receive implants every year from doctors, including hip joints, breast and dental implants, hearing aids, *etc.* Most of these implants are used to replace hips, knees, and spinal discs. By the end of 2030, there will have been an increase in hip joint implantation of 174 %. According to projections, the total number of knee arthroplasties will increase by 67.3 % (348 million operations) from the current pace [6]. By 2020, the market for orthopaedic products will be worth close to $47 billion worldwide. Several variables, including the aging population, changes in technology and lifestyle, enhanced aesthetics, and the desire for better functioning, are projected to cause an increase in the sale of orthopaedic items. The goal of current orthopaedic research is to create multifunctional biomaterials that will perform flawlessly for a lifetime. Accidents are a significant cause of death and morbidity in India hence the development of appropriate biomaterials with high durability and outstanding mechanical, biocompatibility and cytotoxicity qualities is extremely vital. Additionally, there has been an increase in vehicle accidents in India over the past 20 years, with six million non-hospitalized visits and 1.2 million hospitalizations. In addition, millions of individuals worldwide are also affected by musculoskeletal problems, including osteoarthritis, bone cancer, and spine issues. Around 26 million people in India have osteoporosis, according to the Indian Osteoporosis Society [7].

Biomaterials

Following the first biomaterials conference held in 1969 at Clemson University (South Carolina) [8], biomaterials acquired notoriety. Biomaterials are substances, whether natural or artificial, that can come into touch with living tissues and not react negatively. It can be used to take the place of, improve upon, or aid in the operations of a particular organ piece or the complete organ system. As shown in (Figure 1), biomaterials science focuses on the investigation of their physical and biological interactions with the biological environment. [9]. Any biomaterial must meet the essential requirement of being biocompatible with human tissues. Additionally, exceptional mechanical qualities and resistance to corrosion and wear are crucial.

Biocompatibility

A substance's biocompatibility is defined as its capacity to carry out the desired function for a medical therapy without causing any unfavourable local or systemic effects in the recipient while still inducing the necessary beneficial cellular response and enhancing the therapeutic effect in a clinical setting. [10]. Three kinds of biocompatibility are categorized based on tissue response phenomena (Figure 2).

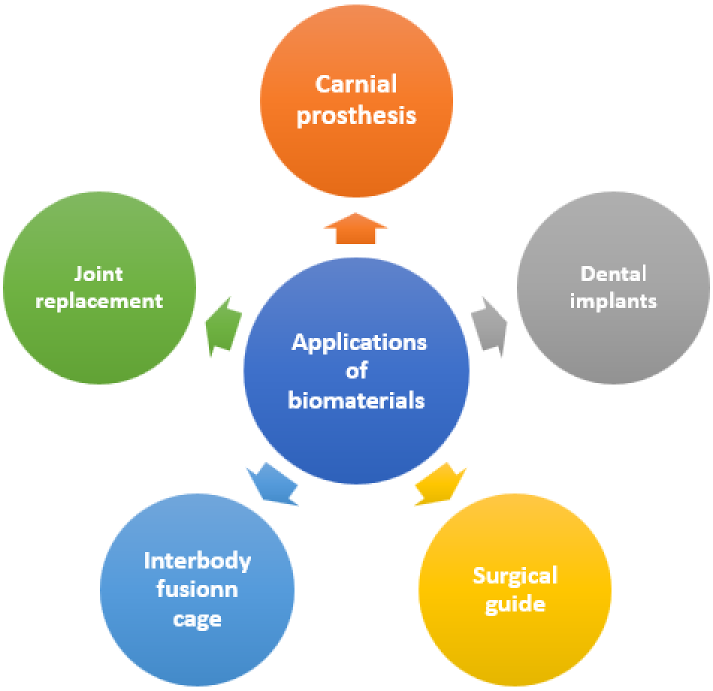

Figure 1. *Application of biomaterials*

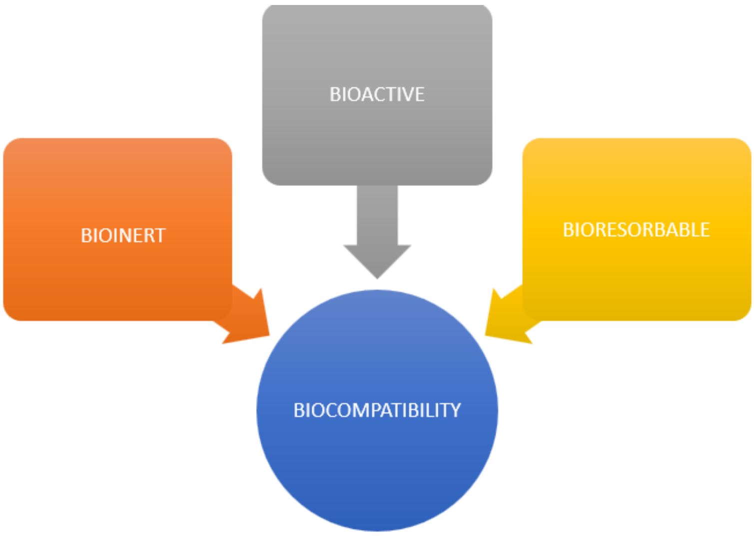

Figure 2. *Classification of biocompatibility as per tissue response phenomena*

Biomaterial mechanical properties

Particularly for load and bearing applications, the mechanical characteristics of biomaterials are crucial. Implants' modulus, elasticity, hardness, scratch resistance, and tensile strength are essential mechanical characteristics. Biomechanical incompatibility is associated with implant fracture brought on by mechanical failure. The materials utilized to replace bone must, for this reason, have mechanical qualities comparable to those of bone. NaOH-treated Ti substrates were calcined by Kim *et al.* [11]. They observed that heat treatment at 600 °C has found maximum mechanical stability from other heat treatments. Piveteau *et al.* [12] looked into the adherence power of TiO_2/hydroxyapatite (HAp) films on Ti substrate utilizing rotating-bending test and tensile bond test techniques. In order to examine the mechanical characteristics and stiffness of cervical plates under static and dynamic loading scenarios, Brodke *et al.* [13] developed a simulated cervical corpectomy model. The flexural strength of hydrothermally produced HAP nanoparticles was reported by Kothapalli *et al.* [14] to be 78 MPa. The effect of the porosity of HAP coatings on the development of osteoblast-like cells was examined by Spriano *et al.* [15]. Microhardness and compressive strength of HAp whiskers were determined by Bose *et al.* [16]. Using nano-indentation and AFM methods, Nathanael *et al.* [17] investigated the surface roughness and strength of sol-gel dip HAP/TiO_2 coatings. Scratch testing was done on calcium phosphate (CaP) coatings that were magnetron sputtered onto stainless steel 316L substrates by Toque *et al.* [18]. Hardness, elastic modulus, surface roughness, and toughness of the sol-gel generated HAP coatings applied to commercially pure Ti (CP-Ti) and Ti6Al4V alloys were determined by Roest *et al.* [19]. Using the indentation technique, Souza *et al.* [20] examined the tribo-mechanical behaviour of films.

Corrosion resistance

The metallic implants must communicate with the human body's aggressive bodily fluid, which contains chlorides and proteins. Therefore, corrosion resistance is essential for implants. During the corrosion cycle, the alloy's metallic components oxidize, turning dissolved oxygen into hydroxide ions. It has certain drawbacks, including (a) the degradation process lowering the structure's quality and (b) the possibility of unfavourable host tissue reactions to the waste products. Metallic implants deteriorate due to wear, electrochemical breakdown, or a synergistic mix of the two [21]. The strength of oxidation/reduction processes and physical barriers that prevent corrosion kinetics control the corrosion of metallic implants. These characteristics also contain a number of other factors, such as solution factors, geometric factors (such as taperness), metallurgical factors (such as microstructure and composition), mechanical factors (such as stress and relative motion), and others (*e.g.,* pH and solution composition). Corrosion, according to Spector [22], was the main factor in hip implant failure. A review of electrochemical corrosion processes, implant responses to host environments, and host tissue responses to implant corrosion products was made by Jacobs *et al.* [23]. The effects of different forms of corrosion on the lifespan and use of orthopaedic devices were reviewed by Kamachimudali *et al.* [24] and discussed. Using a potentiodynamic polarisation test, Kim *et al.* [25] assessed corrosion resistance after HAp/TiO_2 coatings were applied to Ti6Al4V substrates. They came to the conclusion that the TiO_2 coatings' deposition increases the substrate's corrosion resistance. According to Spriano *et al.* [15], the bioactive layer applied to Ti6Al7Nb substrates had outstanding corrosion resistance. According to Balamurugan *et al.* [26], HAp/SS 316L exhibits greater corrosion resistance than bare surfaces. The electrochemical behaviour of HAp/TiO_2 coatings applied to SS 316L by dip coating was assessed by Balamurugan *et al.* [27]. In simulated bodily fluids, electrochemical experiments revealed good corrosion resistance and minimal metal ion leaching (SBF). Many authors claimed that adding double-layer HAp coatings on SS alloy substrates and electrochemically depositing increased corrosion resistance. Tantalum oxide bilayer coatings that had been altered by organophosphate acids were investigated by Arnould *et al.* [28] for their impact on corrosion resistance.

Implant materials

Over the past 20 years, the usage of biocompatible materials in medicine has increased significantly. Gold, silver, and iron are the most often utilized metals in the creation of lengthy bone fracture pins and spinal cables. In orthopaedics, biocompatible materials are mostly used to repair or replace joints, fractured, or otherwise damaged bone [29]. Through examination, these materials are used as long-term implants, screws, and pins through inspection.

Alternative biomaterials are expected to have an impact on key burden areas, such as spinal components, tumor bones, orthopaedic joints, and dental components [30]. Because of this, it is crucial to consider both performance and mechanical strength. Implants can affect cellular responses like differentiation and mineralization because the bone is so sensitive to changes in its mechanical characteristics down to the smallest level. The mechanical characteristics of the implant materials were critical and had a significant influence on the bone repair process. A substance should have both mechanical strength and fracture resistance when used as a transplant material. Metals have the highest tensile strength of all materials, followed by polymers and ceramics. (except for zirconia) [31]. Tensile strength and corrosion resistance are all areas in which metals outperform ceramics. There are several issues with the metallic biomaterial that need to be fixed. The most significant is the release of toxic chemicals from metals during metallic corrosion. Some of the most common examples include cobalt alloys, stainless steel alloys, titanium alloys, and a variety of other forms of metallic biomaterials. Ninety percent of osteosynthesis devices have been constructed of austenitic stainless steel since the 1930s. The material's mechanical strength and biocompatibility were both exceptional qualities. The exceptional strength-to-weight ratio of Co-based alloys makes them a popular choice for anatomical research. The inherent strength and wear resistance of these materials are greatly enhanced by the presence of chromium, which is present in concentrations of more than 18%. When titanium was first used, it was in the aircraft sector in the 1950s. The substance has now been altered to allow its usage in human medical implants [28].

Implant-grade stainless steel was created in 1926 by T. Krupp. It was 18-8 because it had 18 % Cr and 8 % Ni (type 302 in the present-day arrangement). Tools for surgery and dentistry are made of commercial-grade stainless steel in the biomedical industry. Vacuum melting (VM), and other stainless-steel implant-specific manufacturing techniques, such as vacuum arc re-melting (VAR) or electro-slag refining (ESR), provide resistance to pitting and crevice corrosion as well as a reduction in the amount of nonmetallic additives. Austenitic stainless steels are widely utilised due to their flexibility. On the open market, vacuum-melted 316L stainless steel implants with cheap cost, good strength and ductility, machinability, and easily adjustable mechanical qualities are available [32-37]. Since the 1930s, titanium alloy has been used in alloying and thermo-mechanically advantageous to considerably strengthened and low-density (4.5 g/cm^3) titanium alloy implants. The biomedical industries first used titanium implants in the 1930s. A high temperature was required for vacuum processing in order to prevent the creation of an oxygen reaction. The only version of the four grades that is commercially accessible is pure titanium (Cp-Ti).

The existence of porosity on the implant's surface is one of the most critical elements influencing osseointegration [38]. These holes considerably aid in the creation of connective tissues between the implant and the bone, as well as osteoblastic cell penetration. By applying biodegradable polymers to the surfaces of the implants, it may be possible to create in vivo porosities. Mg can be studied as a biodegradable phase, making it one of the most biodegradable materials in use today. A double-layer coating of HAp and Mg was deposited using cold spray (CS), with the PS method applied to the HAp layer in the middle to increase corrosion resistance. The HAp layer was applied using the PS method. Potentiodynamic polarization curves are produced by HAp and HAp-Mg two-layer coatings (Figure 3). The HAp with 10 wt.% of Mg has the best corrosion resistance due to the intermediate dense HAp layer. This is due to the HAp layer's contribution to corrosion prevention. Porosities will form if the Mg phase is dissolved at the surface of coated samples. These porosities provide ideal conditions for bone formation [39]. Because of the porous nature of the HAp-Mg layers on top, it is impossible to increase the substrate's corrosion resistance.

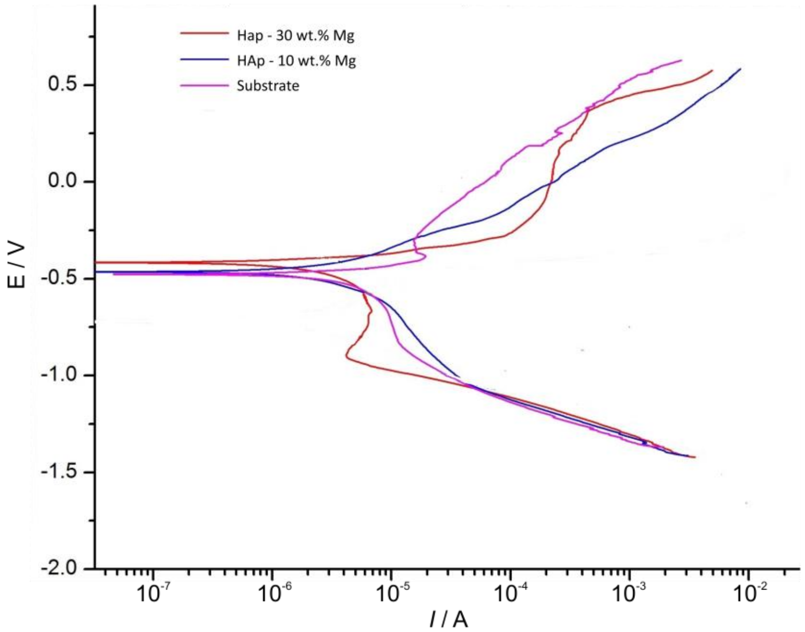

Figure 3. *Potentiodynamic polarization curves*

HAp as a bioactive bone substitute

The main mineral components of teeth and vertebral bone are CaP salts. The biominerals incorporated in the protein matrix that make up bone tissues are regarded as anisotropic composites [40]. The biomineral phase, which is made up of several CaPs types, comprises 65-70 % bone, 5-8 % water, and 5-8 % organic matter [40]. The collagen provides elastic resistance and acts as a mineral deposition and growth [39]. Among the several CaP salts, HAP is a superb biocompatible substance and a crucial biomaterial for applications in the bone and dentistry sectors. About 95 % of HAp is found in the enamel of human teeth. In the human body, CaP exists in a stable crystalline phase known as 92 HAP [39]. Natural bone is composed of nanostructured, nonstoichiometric HAp with dimensions of 20 nm in width and 50 nm in length. There are also a few minor levels of certain substituent ions, such as magnesium, fluoride, and carbonate. The Ca/P molar ratio of natural HAp is often less than 1. Due to synthetic's superior biocompatibility [41] and strong osteogenic potential. For many years, HAP has piqued the attention of the medical community.

Surface modification

Biomaterials and the biological environment communicate at the surface. Therefore, a biomaterial's surface characteristics ultimately determine whether it will be accepted or rejected by the human body. Although the bulk properties of biomaterials regulate their mechanical properties, the surface features govern tissue-biomaterial interactions. Such interactions are projected to occur in a zone no bigger than one nm. [42]. Numerous studies have related surface features such as surface topography and chemical composition to bone implant clinical efficacy. [42]. It is feasible to improve material and biological reactions while maintaining bulk properties constant. Surface modification falls into three broad categories: (a) introducing materials with desirable functionalities to the surface, (b) changing an existing surface to have more desirable compositions and topographies, and (c) removing material from an existing surface to achieve specific topographies. Because just the surface layer must be changed, surface modification is seen as a low-cost strategy from an economic aspect, as illustrated in (Figure 4). Its main goal is to increase biocompatibility, bioadhesion, antimicrobial characteristics, corrosion resistance, and wear resistance. There are several ways to modify surfaces, including the plasma spraying technique is frequently used to cover Ti6Al4V implants with HAP or other CaP minerals in an effort to promote bone bonding.

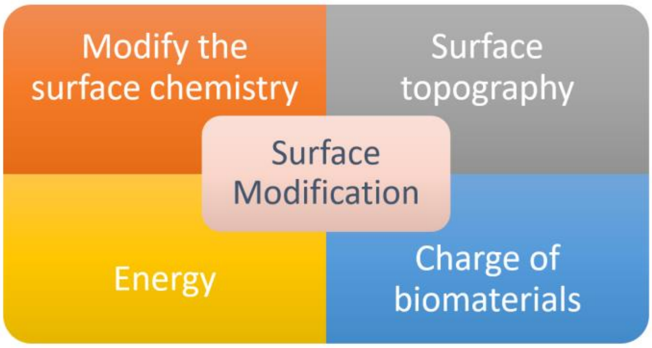

Figure 4. *Through alterations to the chemistry, topography, energy, and charge of the surface of biomaterials*

Lasers and thermal sprays aiding in medicine

Claddings are widely used for the protection of bioimplant materials against corrosion [43-55]. Thin and thick coatings are also applied to bioimplant for protection against corrosion and erosion [56-59]. Corrosion and erosion severely affect the life and durability of the different materials [32,33,60-73]. Coatings techniques such as HVOF/HVAF [32,37,62,68,71-81], plasma spray [82-84], chemical vapor deposition (CVD) [85-88], sputtering [35,58,79,89-95], plasma vacuum deposition, *i.e.* PVD [54,58,96-101], electrochemical vapor deposition (EVD) of yttria-stabilized zirconia films, [102] *etc.* Thin coatings are generally preferred compared to thick coatings in biomedical applications of different alloys [35,95]. Surface modification techniques can be an asset in protecting the alloy bioimplant. Due to its cost-effectiveness, plasma spray has been one of the most popular methods for increasing the bioactivity of substrate material by coating it with HAp. The US Food and Drug Administration has also approved this method. However, this technique creates coatings with a rapid rate of deposition and a uniformly thick layer. However, this procedure needed pricey machinery and a high temperature for the coating to be deposited on the substrate. This approach produces coatings that have poor coating-substrate adhesion. Additionally, the interface between the deposited coatings and the substrate surface is noticeable, which is normally not desired for long-term applications [103,104]. The inherent flaws in the plasma spray approach have been solved using laser energy. In this method, the surface of the implants was modified using a laser by creating a composite layer with HAP that was diffusion-bonded to the surface of the substrate. Balla *et al.* [105] presented the procedure for using laser energy to create a coating of HAp powder onto SS-316L. The surface layer of the bioactive HAP powder on substrate SS-316L may be altered by varying the laser energy input parameters, according to the authors' findings. However, there are significant downsides to the laser processing of materials, such as greater running and installation costs, interface cracking, and thermal distortion [106].

Recent years have seen a significant increase in microwave heating (MH) as a cutting-edge method for treating a wide range of materials, including metals, ceramics, and composites [107]. Depending on the dielectric characteristics of the target material, electromagnetic energy is transformed into heat energy in MH. Depending on the dielectric and magnetic characteristics of the target material, both fields (electric and magnetic) interact with the substance during microwave irradiation. Both electric and magnetic field components interact with the substance during microwave irradiation, depending on the magnetic and dielectric properties of the target material. The target material is heated uniformly by absorbing electromagnetic energy at the atomic or molecule level via polarisation by dipole rotation and ionic conduction. Because heat transfer in traditional heating depends on the target material's thermal conductivity, MH differs substantially from traditional material heating. Due to prolonged heat exposure, the component exterior may be injured during conventional heating of thick materials with restricted thermal conductivity. Furthermore, there is a considerable temperature difference between the core and surface of these materials during conventional heating, which affects the mechanical properties of treated materials. During microwave irradiation, on the other hand, the target material's core rapidly warms up as a result of heat created within the material itself, and the heat is subsequently transmitted to the surface. As a result, the heating profile is reversed, as opposed to typical heating, which is "outside-in." This one-of-a-kind heating phenomenon associated with MH has advantages in terms of an improved diffusion process, low thermal gradient, reduced residual stresses due to less variation in a temperature gradient, and significantly shorter processing time due to the volumetric and uniform heating caused by microwave interaction with the target material [107]. According to Agarwal [108], microwave energy has far greater potential for material processing than conventional methods due to benefits such as shorter processing times, which result in significant energy and cost savings, and finer microstructures, which improve the mechanical properties of the target material.

Applications of microwave radiation as a surface engineering tool in biomaterials

Because they are good microwave absorbers at room temperature, ceramics, polymers, composites, and semiconductors were the main materials that could be processed using microwaves up until 1999. However, today, all powdered metals can be processed using microwave energy just as effectively and efficiently as ceramics. Microwave (MHH) is also used to treat bulk metallic materials utilizing microwave radiation [109]. This has created a new area for research into the challenging and expanding demand of many metallurgical applications as well as to take full advantage of the benefits of microwave processing of materials. Cladding, coating, and glazing are the three categories under which microwave technology is most commonly used in surface engineering.

Microwave cladding

Depending on the demands, an overlay of acceptable materials on a substrate can be created using the surface engineering approach known as microwave cladding. In general, cladding is used to provide the surface of a substrate with certain qualities, increase wear resistance, or both. It entails total melting of the clad material and partial melting of the substrate material in terms of the top layer. There are several different cladding methods available today. Better mechanical characteristics can be found in the fine microstructure. Despite these benefits, laser cladding has certain inherent drawbacks, such as the creation of high thermal stress that might result in clad layer breaking and thermal distortion. Furthermore, when covering huge regions, laser cladding is not a particularly cost-effective technology. Due to its unique benefits, microwave processing of materials has recently become more common. Gupta and Sharma were the first to disclose the use of microwave technology for altering a material's surface properties in the form of cladding in a patent [110]. With the aid of MHH, the authors have effectively established the cladding of metallic and composite powders on metallic substrates. Figure 5 schematically depicts the irradiation process used for cladding. The scientists employed charcoal powder as a susceptor in this approach, which easily absorbs microwave radiation and heats to a high temperature. Later, the heated system uses a standard way of heat transmission to warm the coated powder particles. As a result, the metallic powder particles begin to absorb the microwaves at greater temperatures. To prevent any contamination of the powder with charcoal powder, a solid layer of graphite was utilized as a separator. According to the authors, chromium carbides and other intermetallic compounds have formed, increasing the hardness of the clad area. Therefore, this changed surface could be an excellent defense against functional surface wear.

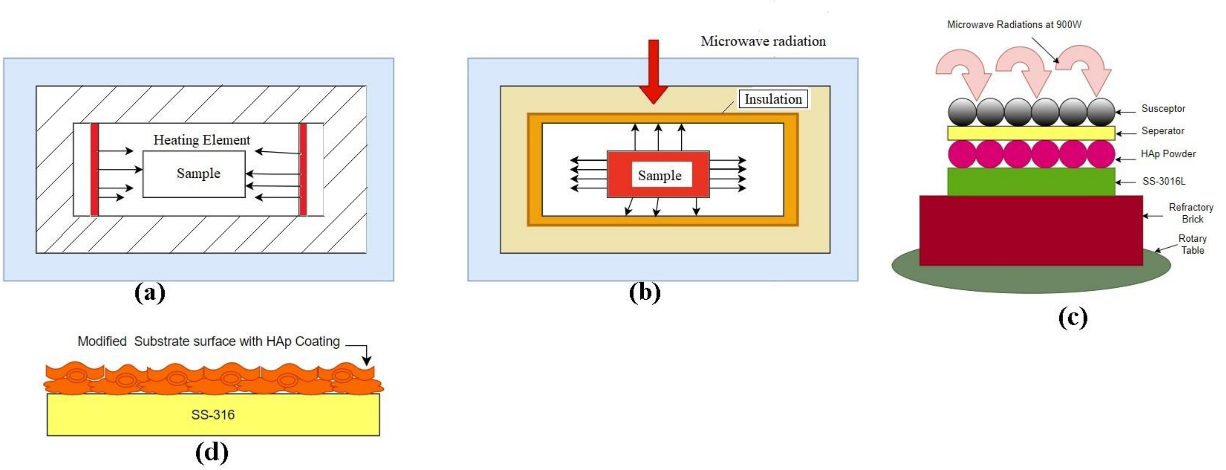

Figure 5. *(a,b) Surface modifications using hybrid MH were performed on SS-316L alloy, and this schematic depicts the experimental setup used. (c) Initially, the susceptor heats up because of an increase in microwave energy (d) As the HAp powder melts, it mixes with the surface layer of the substrate to form a composite, modified layer*

An appropriate hard material particle (such as SiC, Al2O3, or diamond) is added to the metal-based matrix used in composite cladding to improve wear and abrasion resistance. The additive reinforcement phase may additionally contain a dry lubricant (such as MoS2, PTFE, and graphite) to impart lubricating qualities and decrease the coefficient of friction [111]. It has been discovered that adding extra hard phase(s) to composite cladding has a significant influence on the mechanical characteristics of the material, such as hardness, strength, and elastic modulus. Cladding hard particles with a metallic binder in the form of a metal-matrix composite improve the surface's wear properties, according to Huang *et al.* [112]. Thermal spraying, arc welding, and laser cladding are all methods for generating composite cladding on metallic substrates. Heat spraying, while effective, cannot produce diffusion bonding between the hard phase particle and the substrate. Flexural strength is minimal because of the produced clad's poor bonding with the substrate. Due to the tremendous heat created during arc welding, a melt pool develops on the surface of the substrate, resulting in strong metallurgical bonding. The substrate may become severely distorted due to the high dilution of the composite coating (due to the high energy input). Furthermore, arc welding and laser cladding commonly exhibit the creation of heat-affected zones (HAZ). On the other hand, microwave cladding only causes the surface layer of the substrate - between 10 and 15 m - to melt. According to Gupta and Sharma [113], the overlaid particle to completely melt. Because the clad powder particles effectively pair with the microwave, this is feasible (due to better skin depth).

While the molten powder layer's traditional conductive heating causes the substrate to melt to a few microns (just the skin depth, 10-15 m) thicknesses. The metal-based substrate's main body reflects the microwave. Minimizing the heating caused by microwave contact as a result. This cladding has the benefit that the metallic reinforcement produces improved hardness and hence greater wear resistance, while the metallic matrix demonstrates the superior toughness. Microwave cladding has been developed and in-depth characterized by Sharma and Gupta [114]. In cermet cladding, scientists have reported a skeleton-like reinforcing (hard carbide) structure in the Ni-based matrix. Additionally, they show how evenly dispersed the hard particles are in the soft matrix; the clad shows a markedly superior wear resistance [115]. Table 1 is an overview of the outcomes from the microwave clads as reported by different sources. It is important to highlight that microwave cladding significantly reduces porosity as a result of the uniform heating brought on by MHH [116]. Furthermore, it was noted that the microwave clads did not exhibit the stress corrosion cracking often seen in laser clads.

Table 1. *Results after cladding of metallic powder on metallic substrate developed through MHH heating mechanism in multimode microwave oven*

Sr.	Author	Substrate	Coating	Cladding results
1	Gupta and Sharma [113]	ASS-316	WC10Co2Ni powder	Average microhardness: 1064 ± 99 HV Porosity: 0.89 %.
2	Gupta and Sharma [114]	ASS-316	EWAC + 20 % composite powder	Flexural strength: 629 ± 8 N Average microhardness: 416 ± 20 HV
3	Gupta and Sharma [117]	ASS-316	Cu powder of size 5 µm and a purity of 99.5 %	Mean hardness: 270 ± 30 HV
4	Gupta and Sharma [118]	ASS-316	EWAC powder (Ni-based having a size of 40 µm)	Average microhardness: 304 ± 48 HV.
5	Gupta and Sharma [115]	ASS-316	WC10Co2Ni powder	Average microhardness: 1064 ± 99 HV
6	Gupta and Sharma [116]	ASS-316	EWAC (Ni-based + 20 % $Cr_{23}C_6$ powder)	Phases: $FeNi_3$, NiSi, and $Cr_{23}C_6$ phases average microhardness: 425 ± 140 HV
7	Kaushal *et al.* [119]	ASS-316 L	Ni-WC-Cr3C2 powder	Phases: Cr_7Ni_3, NiC, Fe_6W_6C, $Co_3W_3C_4$, $FeNi_3$, and NiW; average Vicker's microhardness: 503 ± 34 HV (1.6 times that of substrate's HV)
8	Singh *et al.* [120]	Mild Steel	Inconel-625	Average microhardness: 550 ± 15 HV
9	Singh *et al.* [121]	SS-316	Inconel-625	Microhardness increases at 980 STA

Microwave cladding of biomaterials

Research describes employing the hybrid microwave heating (MH) procedure to reinforce hydroxyapatite (HAp) powder on the surface layer of stainless steel (SS-316L) to boost the metal's bioactivity [110]. The modified substrates were post-heated for one hour at 400 and 700 degrees Celsius. The microstructural study of the layer revealed the presence of HAp particles as well as certain reaction-induced products in the iron-based austenite dendritic matrix of the modified composite layer. Heat-treated substrates showed a higher microhardness value than freshly deposited substrates due to densification of the modified layer after heat treatment. Porosity, surface imperfections, and faults were reduced after heat treatment. SEM images of the modified and unmodified SS-316L after immersion testing in simulated body fluid demonstrated quick apatite production capabilities on the modified substrates. The amorphous phase and porosity contents of the modified substrates reduced after heat treatment at 700 °C, lowering their ability to form apatite. A superalloy substrate with strong glass-ceramic coatings that are nickel-based has been reported. The MgO-Al_2O_3-TiO_2 system, which is based on glass, was used to create the ceramic coatings. The authors have compared the coating produced by both traditional and microwave techniques. The findings demonstrate that coatings produced by microwave processing have finer microstructure than crystallites made using more traditional methods. When compared to traditionally treated coatings, the hardness of the coating produced by microwave processing was much greater (6 GPa), as determined by the depth-sensitive indentation method. It has been proven that coatings produced using microwave technology exhibit lower surface roughness (Ra) values than coatings prepared using more traditional methods.

In the work, authors applied microwave radiation to alumina-reinforced hydroxyapatite clad UNS S31254 stainless steel to change the surface layer to enhance its bioactivity [122]. An industrial microwave oven that operated at 1.1 kW and 2.45 GHz and was accompanied by an infrared pyrometer was used for the microwave surface modification procedure. Additionally, the surface-modified samples underwent a 1-hour thermal heat treatment at 400, 600 and 800 °C in a muffle furnace. SEM, energy-dispersive spectroscopy, an X-ray diffractometer, and simulated physiological fluid tests were used to investigate the metallographic, compositional, phase analysis, and bioactivity of microwave surface-modified materials. The samples with microwave surface modifications included alumina, according to the X-ray diffractometer analysis. The microwave-assisted surface modification layer is largely composed of iron (Ni-Fe)-based austenite dendrites, together with hydroxyapatite and specific reaction products, notably in the interdendritic areas, according to the microstructural investigation. The results reveal that heat-treated samples exhibit lower porosity and higher hardness when compared to samples with surface changes made when they were deposited. Furthermore, the 800 °C heat-treated samples showed the lowest porosity (about 56 % less than the as-deposited sample) and maximum hardness (roughly 23.5 % greater than the as-deposited sample). The loss of pores and amorphous phase as a result of heat treatment reduced the surface-modified materials' ability to bind bone. In another work, the authors employed microwave hybrid heating (MHH) to enhance the biocompatibility of implants composed of austenitic stainless steel (SS-316L) by changing the surface chemistry and shape [123]. Bioactive hydroxyapatite (HAP) powder was added to the polished SS-316L alloy surface layer using an industrial microwave oven operating at 2.45 GHz and 1.1 kW. The altered layer of SS-316L alloy was characterised using phase analysis, microstructure inspection, porosity analysis, and the ability to generate apatite. According to the microstructural examination, the inter-dendritic parts of the modified layer of SS-316L alloy contained austenite dendrites with HAP as well as numerous reaction-induced phases such as Fe3P, FeP, and Fe2P. The modified layer of the SS-316L alloy outperforms the unmodified layer in terms of microhardness. The altered layer has a porosity of 2 % to 3 %. In an in-vitro study using simulated bodily fluid (SBF), the microwave-assisted surface-modified layer of SS-316L alloy demonstrated improved bioactivity. The SEM images demonstrate that an appetite layer has developed in the modified layer of the SS-316L alloy. The observed biological improvements in the microwave-assisted modified layer were caused by changes in the surface morphology and chemistry of the SS-316L substrate material caused by cladding.

Conclusions

The primary objective of this review is to compile a substantial body of knowledge regarding HAp coatings and coating methods for the purpose of the production of biomedical implants. It has been reported that a superalloy substrate has nickel-based coatings that are robust glass-ceramic coatings. Ceramic coatings were manufactured by employing the MgO-Al_2O_3-TiO_2 system, founded on the glass as its primary component. When compared to coatings that had been subjected to conventional treatment, the coating that had been produced by microwave processing had a significantly greater degree of hardness (6 GPa). A higher microhardness rating can be found in the modified layer of the SS-316L alloy in comparison to the layer that has not been modified. The altered layer had a porosity of between 2 and 3 % throughout its entirety. According to an in-vitro investigation that was carried out in a solution that was meant to mimic bodily fluid, the modified layer demonstrated an increased level of bioactivity.

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