

## Review Article

# A Review of Additive Mixed-Electric Discharge Machining: Current Status and Future Perspectives for Surface Modification of Biomedical Implants

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Surface treatment remained a key solution to numerous problems of synthetic hard tissues. The basic methods of implant surface modification include various physical and chemical deposition techniques. However, most of these techniques have several drawbacks such as excessive cost and surface cracks and require very high sintering temperature. Additive mixed-electric discharge machining (AM-EDM) is an emerging technology which simultaneously acts as a machining and surface modification technique. Aside from the mere molds, dies, and tool fabrication, AM-EDM is materializing to finishing of automobiles and aerospace, nuclear, and biomedical components, through the concept of material migrations. The mechanism of material transfer by AM-EDM resembles electrophoretic deposition, whereby the additives in the AM-EDM dielectric fluids are melted and migrate to the machined surface, forming a mirror-like finishing characterized by extremely hard, nanostructured, and nanoporous layers. These layers promote the bone in-growth and strengthen the cell adhesion. Implant shaping and surface treatment through AM-EDM are becoming a key research focus in recent years. This paper reports and summarizes the current advancement of AM-EDM as a potential tool for orthopedic and dental implant fabrication. Towards the end of this paper, the current challenges and future research trends are highlighted.

## 1. Introduction

Electrical discharge machining (EDM) is an advanced machining process used to shape any electrically conductive material. It serves as an alternative to traditional processes such as milling, drilling, and polishing [1]. EDM has proven to have an outstanding performance in machining advanced materials and in producing highly complex shapes [2–6]. Thus, it recently draws the attention of quite several researchers and industries. The use of EDM in mold, tools, automobiles, and aerospace industries has been largely documented [7, 8]. However, Volosova et al. [9] report the usage

of EDM to produce thin-walled separators in the nuclear industry. A new technological innovation which involves mixing the dielectric fluids with certain additives such as metal powders, surfactant, nitrogen, and oxygen gasses has broadened the EDM applications. Thus, when the powder particles are added to the dielectric fluid, the machining technique is called powder mixed-EDM (PM-EDM) or is generally referred to as additive mixed-EDM (AM-EDM). Recently, the application of AM-EDM in the biomedical field has drawn a commendable attention [10–12]. Apart from enhancing the corrosion and wear resistance, AM-EDM has immensely contributed to improving the fatigue life and

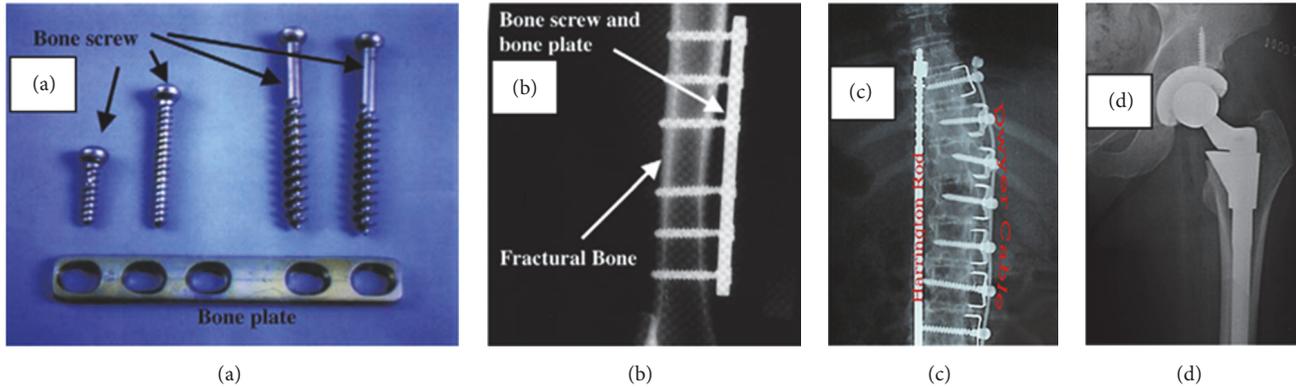


FIGURE 1: (a) Bone plate screws. (b) Implanted bone plate and screws. (c) The Harrington rod. (d) The stem of the total hip replacement (SS, Co-based, or Ti-based alloys) [41, 42].

mechanical properties of the orthopedic implants [13–18]. However, the use of AM-EDM to deposit a nanoporous and biocompatible layer on the machined implant surface has been lately reported [19, 20]. This deposited layer provides a strong implant-bone bonding.

The routinely used orthopedic and dental materials are stainless steel, Co-Cr alloys, Mg alloys, and Ti alloys. These materials are normally shaped in the form of wires, rods, pins, screws, and plates (Figure 1) depending on the application site. A good biomaterial should have enough biocompatibility, sufficient strength, good wear, and corrosion resistance as well as bone-matched modulus of elasticity [21]. Stainless steel is easily affordable and has appreciable manufacturability, but its low strength, high modulus of elasticity, and release of toxic elements after implantation limit its application. Despite the fact that Co-based alloys have proved to be superior to stainless steel in terms of strength and corrosion resistance, its manufacturing cost is too high. Ti-based alloys are reported to be the most suitable in the field of biomedical research when compared with stainless steel and Co-based alloys, owing to their reasonable biocompatibility, good corrosion resistance, lower modulus of elasticity, and high specific strength [22]. The main drawbacks of Ti-based alloys include their high cost, severe wear, and insufficient osteointegration [23]. Among biodegradable alloys, Mg alloys were reported to have higher biocompatibility and bone-suiting mechanical properties. However, their faster degradation before the maturity of the repaired bone crippled their biomedical applications [24–27]. Several efforts to produce safer and higher quality biomaterials that closely match the bone properties have been introduced. Thus, the discovery of bioglass emerged around the 1970s. Unfortunately, this material developed unsatisfied mechanical properties, especially when applied in load-bearing areas [28]. In the 1960s, metallic glasses (MGs) were discovered and revealed to possess sufficient strength and superior corrosion resistance when compared to Ti alloys [29]. However, the drawback of MGs is their limited specimen size (usually in micron) which retards their application in research and biomedical field [30]. In recent years, higher glass formers elements such as Pd, Ni, Be, and Al were discovered. This led to the

development and production of MGs into larger size (up to 10 mm diameter and 100 mm long) called Bulk Metallic Glasses (BMGs) [31]. It is believed that BMGs maintained the corrosion and wear behavior of the MGs as revealed by Ge et al. [32]. Biodegradable and nonbiodegradable BMGs can be fabricated depending on the alloy combination. The fabrication method of BMGs is like that of plastics. They are commercially processed mainly by injection molding or die casting.

Despite the continuous emergence of new biomedical materials, there are still unsolved issues such as poor bioactivity, high elastic modulus, and release of toxic elements after implantation. To solve these problems, several authors considered depositing a biocompatible, bioactive, and non-porous layer on the implant surface through several kind methods [33]. Calcium phosphate (CaP) compounds, such as hydroxyapatite (HA) which contains the main inorganic component of the bone, were adopted as coating material by many researchers. It was revealed that CaP/HA surface coating not only increases the biocompatibility but also provides a very tight implant-tissue bonding [34, 35]. In addition, porous material is believed to have a reasonable and an adjustable modulus of elasticity [36].

The outstanding performance of AM-EDM, especially with regard to material migration and deposition, has recently drawn the attention of many researchers. Several authors have attempted to investigate the potential of AM-EDM to machine biomedical implants and devices. Klocke et al. [37] reported that EDM can efficiently shape Mg alloys. However, a good finishing without compromise of the material's crystal arrangement or initial performance was achieved. Addition of graphite powder and the surfactant to the dielectric fluid greatly reduced the thickness of the formed recast layer, thereby enhancing the biomedical performance of Ti-6Al-4V [38]. A reasonable transfer of material between the dielectric additives, electrodes materials, was observed during AM-EDM of die steels [39, 40].

This paper critically reviewed the technological advancements of EDM and the current perspective of AM-EDM as a surface modification technique for orthopedic and dental applications. The paper covers the structure and properties

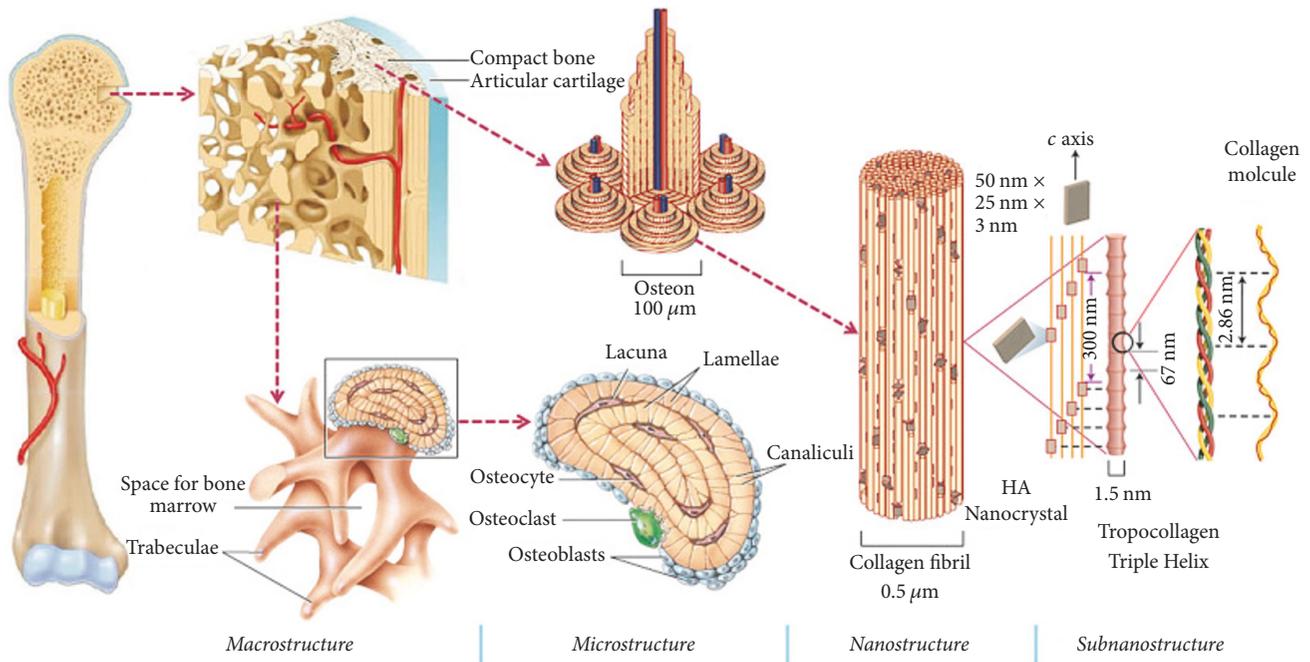


FIGURE 2: Ordered arrangement of bone structure. Copyright @Elsevier, reproduced with permission from Elsevier [43].

of natural bone, the research trends of the available bio-materials, the surface modification of biomedical implants, the implant surface coating techniques, and the potentials of EDM tool in surface treatment of biomedical implants. The mechanism of material migration and deposition during AM-EDM process was also elaborated. The advancement of EDM process from material removal to deposition process is largely contributed by the dielectric additives. However, the authors of this paper classified the dielectric fluids additives into three main classes: solid, liquid, and gasses. The role of these additive types on the surface quality of the biomedical implant was outlined. The last section of this paper stated some of the current challenges and future research directions.

## 2. Natural Bones: Structure and Properties

The larger portion of the human body is occupied by water (about 60–90 wt%). Thus, oxygen, carbon, nitrogen, and hydrogen contain about 96% of the body weight, while the bone and teeth, which mainly contain Ca, Mg, and P together with electrolytes (Na, K, and Cl) in blood, occupy the remaining 4% [42]. The structure of the bone consists of organic (mainly collagen) and inorganic components (mainly CaP-based compound such as HA) as depicted in Figure 2. Collagen and HA function like a steel rod and cement in building reinforced with cement concrete, thus providing the shape, strength, toughness, and flexibility to the body [43]. Cortical bone, which bears the major weight of the body, is considered the highest load-bearing tissue in the lower limb [44]. Despite the high strength, light weight, and extreme toughness of the cortical bone, it has the highest fracture risk

[45]. In most cases, aging, accident, or diseases are the main causes of the bone failure. Although there are several attempts to study the mechanical behaviors as well as the macro-, micro-, and nanobone structures [44–46], reports revealed that the rate of bone fracture is rising every year. Thus, there is still need for thorough investigation on the biomaterials, implants design, and manufacturing.

## 3. Biomaterials and Implants

The major differences between biomaterials and other materials are their capability to associate with the human tissues without eliciting any damage or harm to the body. Biomaterials refer to materials (synthetic or natural) which are used in the form of implant/medical device to replace or repair a damaged/diseased tissue. Based on material properties, material scientists classify biomaterials into three main classes: first are the *metals and alloys* (Ti and Ti alloys, Cobalt-based alloys, stainless steel, magnesium alloys, and amorphous alloys) which are mainly based on metallic bonding. The second class are *ceramics and carbon* (Aluminum oxide ( $\text{Al}_2\text{O}_3$ ), HA, and tricalcium phosphate (TCP), carbon (C), and C-Si) which is built upon ionic bonding, while the last class are *composite* which include both the ceramics and polymers (polymethylmethacrylate (PMMA), polytetrafluoroethylene (PTFE), polysulfone (PSF), and polyethylene (PE)) [47, 48]. Medical implants may be permanent or temporary. Permanent implants, such as hip and stent, are usually implanted in a human body to serve for over one year. On the other hand, temporary implants like bone screws, plates, and suture threads are mostly made of biodegradable polymers. For long-term application, such as fracture

TABLE 1: Routinely used metallic biomaterials and their primary applications [23, 41, 42].

Routinely used biomaterials	Applications
Stainless steel	Orthopedic: for total hip replacement and production of temporary devices such as plates, screws, pins, and nails
Co-based	Orthopedic: for total joint replacement Dentistry castings
Ti-based	Orthopedic: for stem and cup total hip replacement, production of permanent devices such as nails and face makers Dentistry: dental screws (permanent implant)

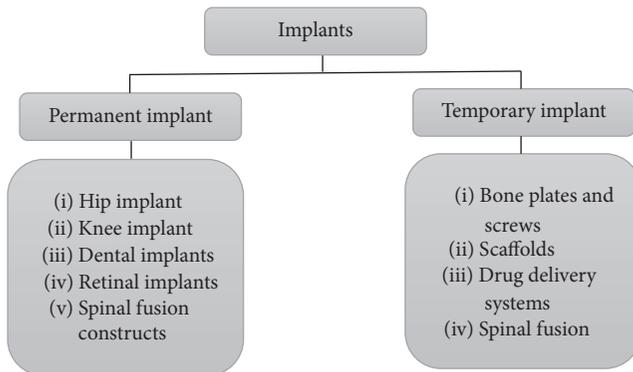


FIGURE 3: Classification of biomedical implants.

fixation or bone replacement, metallic implants are mostly employed. Figure 3 shows the classification of biomedical implants.

The application of metallic biomaterials in orthopedic can be traced back to 1895 when Lane developed the metallic plate for bone fracture fixation [21]. In recent years, the increase in population, aging, accident, and diseases has resulted in a rapid rise in bone fracture, especially knee and hip [49]. Figures 4 and 5 depict different types of total knee and hip replacement. Those with the age of 65 years and above are more prone to bone failure. The reported rate of knee replacement for ages 45–64 in United States rose from 13.1 to 27.3/10,000 population in 2000 and 2006, respectively. However, for those aged 65 and above, the rate increased from 60.1 in 2000 to 88.0/10,000 population in 2006 [50]. Figure 6 depicts the rate of knee replacements in the United States for hospital inpatients aged 45 and above years (2000–2006).

Based on the clinical requirements, metallic biomaterials are generally classified into three distinct generations, namely, first generation, for example, Co alloys, second generation, for example, Ti alloys, and third generation, for example, Mg alloys. The first generation aimed at developing metallic implant with high strength and nontoxic (bioinert) to the surrounding tissues. Second generation metallic biomaterials were developed to provide tight bonding between tissues and the implanted bone, thereby bridging the loosening effect of the first generations. However, it was revealed that the first and second generation biomaterials do not last long after implantation. Thus, to avoid the risk of revision surgery, third generations were developed which are mainly

biodegradable [42]. A good metallic implant is expected to have the following characteristics:

- (i) An outstanding biocompatibility
- (ii) Good corrosion and wear resistance
- (iii) Bone-matched and well-suited mechanical properties
- (iv) A reasonable porosity of 20–50%
- (v) Good bioactivity for sufficient osteointegration [42, 51, 52]

Although new biomaterials are continuously evolving, to date, there is no developed metal or alloy that is completely bioinert or possesses all the required characteristics. Thus, there is still need to synthesize or modify the surface of new biomaterials [53]. However, among various developed biomaterials, Ti alloys are considered the most suited and possess superior characteristics compared to stainless steel and Co-Cr alloys [23]. Currently, metallic biomaterials belonging to either stainless steel, Co-based alloys, or Ti-based alloys were the only materials routinely used for orthopedic and dental applications as summarized in Table 1. Thus, implants belonging to these three classes of biomaterials are already approved by the United State Food and Drugs Administration (FDA) [23].

On the other hand, new generations of metallic biomaterials are currently emerging by altering the alloying composition, the microstructure, and the processing technology of the previous biomaterials. For instance, Ni in stainless steel was replaced with other nontoxic elements to produce Ni-free systems such as FeCrN, FeCrMoN, and FeCrMnMo. The wear resistance of Co-Cr alloy was improved by alloying it with Zr and N and by increasing the carbon content. The high elastic modulus in Ti-alloy was greatly reduced to produce  $\alpha$ -Ti alloys with the lowest modulus of elasticity [54, 55]. However, porosity was intentionally introduced to some metallic biomaterials which significantly reduced the elastic modulus of the alloys [51]. In addition, novel systems of biomaterials called BMGs were recently introduced in biomedical field [28]. Nowadays, several authors are working on BMGs potentialities especially for industrial and biomedical use [56]. Table 2 compares the characteristics of stainless steel, Co-based alloys, Ti-based alloys, and BMGs in biomedical application. The mechanical properties of these alloys are compared to that of cortical bone in Figure 7.

TABLE 2: Characteristics of nonbiodegradable metallic biomaterials (grades ranged from 5, excellent score, to 1, poor score) [22, 23, 28, 57].

Characteristics	Stainless steel	Co-based alloys	Ti-based alloys	BMGs
Machinability				
Conventional	5	1	1	2
Advanced	5	3	3	4
Corrosion resistance	2	3	4	5
Wear resistance	2	2	1	5
Biocompatibility	1	2	3	2
Bioactivity	1	1	1	1

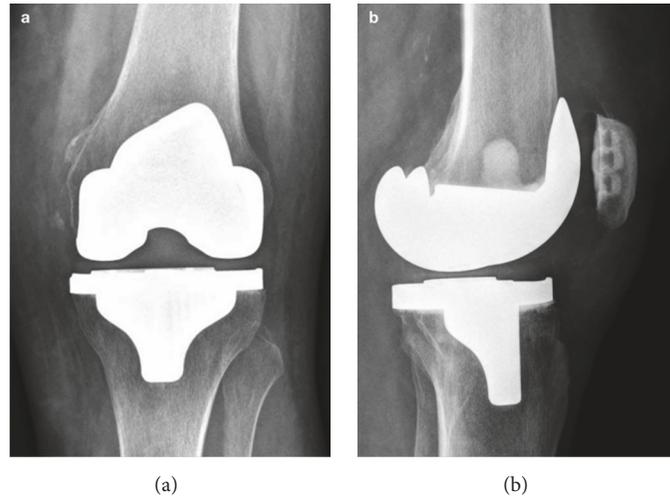


FIGURE 4: Anteroposterior (a) and lateral (b) radiographs demonstrating a total knee replacement [49].

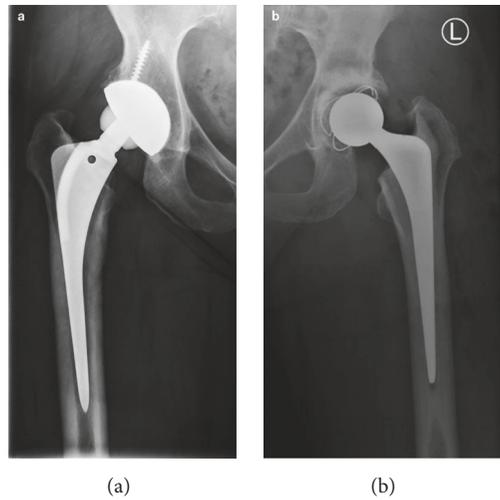


FIGURE 5: (a) Uncemented total hip replacement (THR) with a ceramic on ceramic bearing. (b) Cemented THR using the Exeter system with a metal on plastic bearing [49].

3.1. *Implant Failure.* Even though most of the synthesized implants are much stronger than the human bone, a high number of femoral and dental implants were reported being failed at an early stage of implantation. However, the most successful orthopedic implants last for about 15 years, which is comparatively shorter than the lifespan of human (even the elderly person with a life expectancy of 17.9 years) [42]. Thus, the patient mostly undergoes a revision surgery, which

is not normally recommended. The possible causes of this implant premature failure are due to several reasons such as incompatible mechanical properties of the implant or fixation device, the development of undesired osteointegration between living bone and the synthetic surface, and poor implant surface finishing as well as poor corrosion and wear resistance. To solve this problem, several researchers attempt to modify the implant surface through various methods.

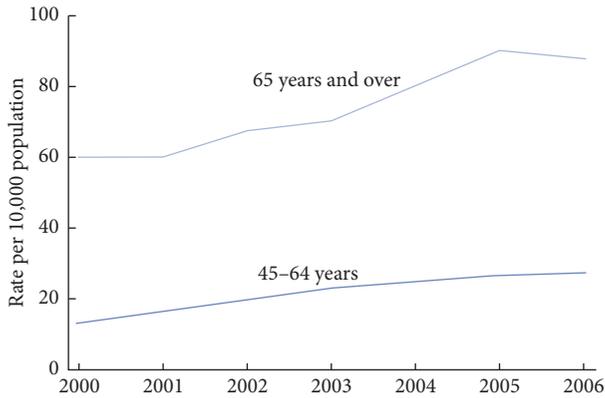


FIGURE 6: Rate of knee replacements in United States for hospital inpatients aged 45 and above years (2000–2006) [50]. Note: rates for 2000–2006 were calculated using US Census Bureau 2000-based postcensal population estimates. Source: CDC/NCHS, National Hospital Discharge Survey.

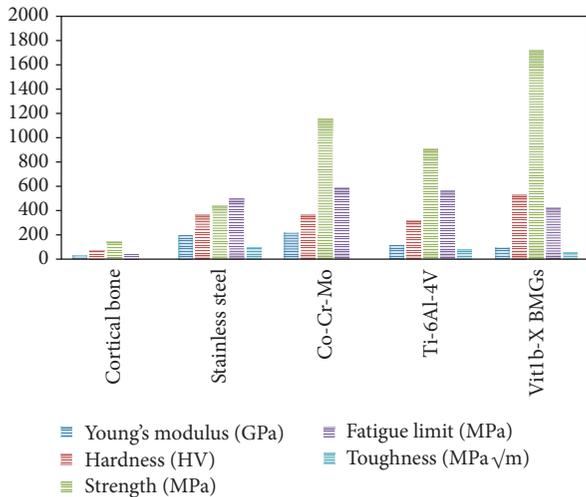


FIGURE 7: Comparison of mechanical properties of the cortical bone and the most commonly used metallic biomaterials [30, 58, 59].

Table 3 critically analyzes some reports on failed orthopedic (femoral) and dental implants. Figures 8 and 9 show the failed femoral plate and dental (inner screw) implants, respectively.

**3.2. Surface Treatment of Metallic Implant.** Despite various attempts to select a material with alloying elements that suit the essential elemental constituents in the human body, there are still some problems of cell adhesion and tissue inflammation which normally developed in the bone-implant interface. This is because the surface of a biomedical implant is always exposed to the dynamic bioenvironment. However, so many reactions take place between the implant surface and the surrounding tissues due to mismatch of the implant surface topography, surface contamination, and chemical and mechanical properties with the host tissues [93]. Although most of the developed implant surface coating techniques are still in the experimental stage, surgeons, bioengineers, and

material scientists believed that surface treatment (modification) remained the key solution to these issues [41, 94]. The surface modification may be morphological (roughness, porosity, texture, and topography) or chemical as elaborated in Figure 10. Morphological surface treatment is considered as the most crucial in the field of biomedical. Through surface modification, the wear and corrosion rate of the implant can be greatly improved. However, modified implant can interact and nicely adhere to the surrounding tissues with appreciable biocompatibility and osteointegration [42].

**3.2.1. Deposition of HA/CaP-Based Compounds.** Calcium phosphate constitutes the major structure of the bone and teeth. Since 1920s, there are several attempts to repair fractured/diseased hard tissues using CaP-based compounds. However, 40 years ago, first CaP-based compound was commercially launched and applied clinically [95]. Since then, the surface treatment of orthopedic and dental implants mainly focused on inorganic HA and/or CaP compounds. Depending on the preparation method, CaP-based implant coating was revealed to improve the mechanical characteristics, biocompatibility, and osteointegration of the implanted bone/tooth [73, 85, 96] Table 4 displays the requirement of HA coatings. Although several methods of depositing HA and CaP have been reported [68, 74, 80, 83], new techniques of preparing and applying a coating on the implant surface are contentiously evolving. These processes are mainly classified into physical and chemical deposition methods. Physical deposition techniques include the following: thermal spray techniques (atmospheric plasma spray (APS), suspension plasma spray, vacuum plasma spray, etc.), pulse laser deposition (PLD) (matrix assisted and ion beam assisted), and sputtering process (radio frequency and direct frequency), while chemical deposition techniques covered sol-gel, dip coating, hot (isostatic) pressing, microarc oxidation (MAO) (anodic spark oxidation and plasma electrolytic oxidation), electrostatic spray deposition (ESD), and electrophoretic coating [69, 77, 84, 97]. The downside of physical methods is their inability to deposit organic compound on the substrate. On the other hand, chemical deposition methods which developed recently are gaining more attention. They serve as hybrid methods for applying both organic (natural bone components) and inorganic compounds on the implant surface [98]. Selection of proper coating method determines the long-term performance and stability of the implanted tissue/bone. Plasma spray which involves spraying of molten (extremely heated) CaP/HA on the substrate is the most commonly used and commercially available technique. Besides the high deposition rates of this method, issues like the need for very high temperature (>10,000°C) and formation of the amorphous coating limit its application [70, 75]. The benefits and drawbacks of different CaP/HA coating methods are summarized in Table 5.

## 4. Overview of Electrical Discharge Machining

The concept of electrical discharge machining (EDM) can be traced back to 1751 when Benjamin Priestly observed a

TABLE 3: Critical review on some prematurely failed orthopedic and dental implants.

Author, year	Material (implant type)	Expected causes/effect of implant failure	Implant duration (months)
Stronach et al. [60], 2016	No evidence of material (Hip stem)	(i) Fretting corrosion (ii) Crevice corrosion (iii) Fatigue failure	72
Guerra-Fuentes et al. [61], 2015	316L stainless steel (Femoral bone)	(i) Bone callus formation (ii) Fretting wear of fixation screws (iii) Fatigue failure (iv) Plate implant fractured	4
Hernandez-Rodriguez et al. [62], 2015	Ti-6Al-4V (dental)	(i) Bone resorption (ii) Rough surface finish in the screw (iii) Crack and fractured	6
Gervais et al. [63], 2016	316L stainless steel (femoral bone)	(i) High-fatigue cycles ( $10^6$ cycles) (ii) Unexpected fall of the patient	<24
Kanchanomai et al. [64], 2008	No evidence of material (femoral bone)	(i) Fatigue crack in the compression hole (ii) Fatigue crack in the locking hole	0.27
Amel-Farzad et al. [65], 2007	Stainless steel (femoral bone)	(i) Corrosion fatigue (ii) Crevice corrosion pitting (iii) Nonstandard material (Supplied in the alloy)	24
Marcomini et al. [66], 2014	Stainless steel (femoral)	(i) Poor material quality (ii) High phosphorous content (nonstandard) (iii) Segregation at grain boundaries (iv) Crack propagation (cyclic loading) (v) Loss of ductility due to cold work	1st failed at 4 sent back to the manufacturer, and 2nd failed at 6

TABLE 4: FDA HA coating requirement [67].

Crystallinity	Phase purity	CaP ratio	Density	Heavy metals	Tensile strength	Shear strength
Min. 62%	Min. 95%	1.67–1.76	2.98 g/cm <sup>3</sup>	<50 ppm	>50.8 MPa	>22

TABLE 5: Benefits and limitations of CaP-based coating methods.

Coating methods	Thickness	Benefits	Drawbacks	Refs.
Sol-gel	<1 $\mu\text{m}$	Coat complex shapes, low cost (thin), low processing temperature, high purity	Controlled atmosphere required, costly raw materials	[34, 68, 69]
Sputter processes	0.5–3 $\mu\text{m}$	Thickness uniformity on Coat flat substrate, good adhesion, dense coating	High cost, time consuming, produces amorphous coatings	[67]
Plasma spray	<20 $\mu\text{m}$	High deposition efficiency, low cost, reduced risk of coat degradation	High temperature led to grain growth, Amorphous coatings, poor adhesion	[70–72]
PLD	0.05–5 $\mu\text{m}$	Crystalline and amorphous coating, dense and porous coating	High cost, time, no uniformity in thickness consuming, surface need to be pretreated	[73–76]
ESD	1–10 $\mu\text{m}$	Low cost, uniformity in coating flat substrate	Coats only visible area, it is fragile	[77–79]
Electrophoretic coating	0.1–2.0 mm	Uniformity in coating thickness, coat complex substrate, rapid deposition rate	Produces cracks coated surface, high sintering temperature required	[80–82]
Hot (isostatic) pressing	0.2–2.0 mm	Dense coating is achieved, coat any dimension	Requires high temperature, complex shapes cannot be coat, high cost, mismatch thermal expansion, differences in elastic property	[83, 84]
Dip coating		Complex 3D and porous substrate can be coat, low cost, faster application	Requires high temperature for sintering, fragile due to thickness, amorphous due to thermal expansion	[85]

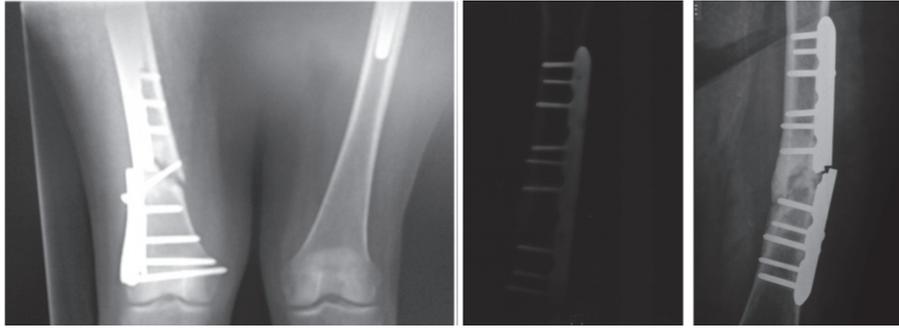


FIGURE 8: Bone and the failed supracondylar plates before and after being fractured. Copyright @Elsevier, reproduced with permission from Elsevier [61, 66].

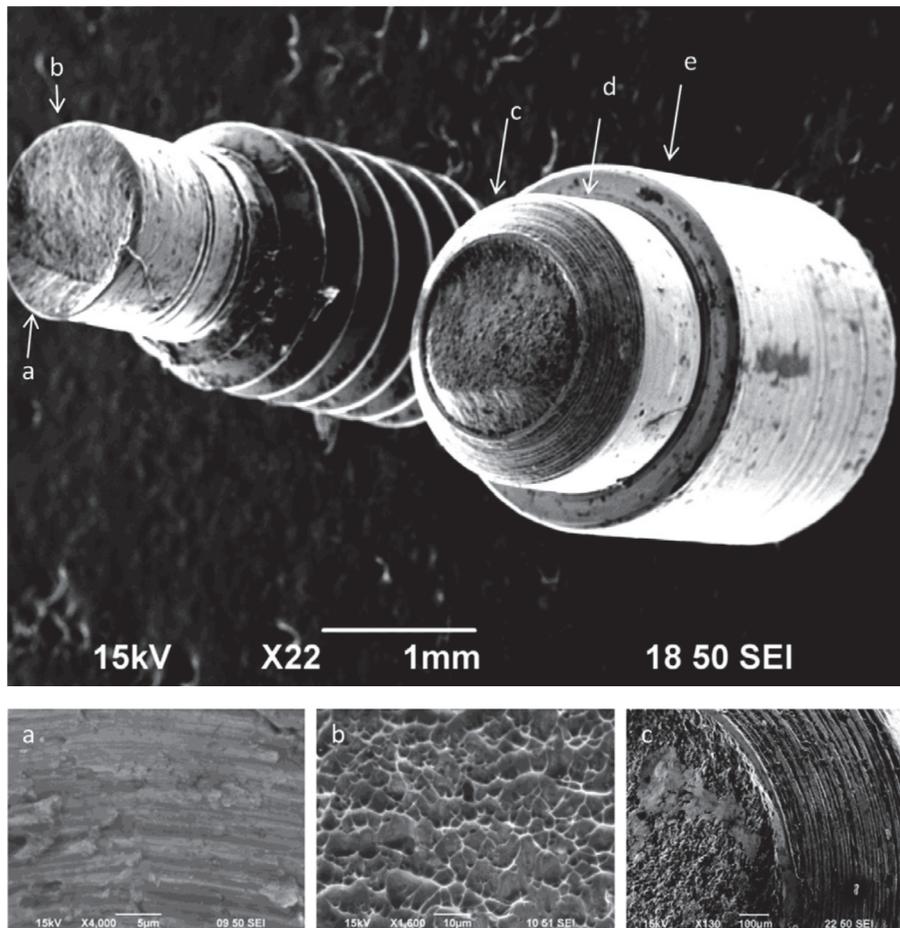


FIGURE 9: SEM images depicting different regions of the fractured dental implant surface. Copyright @Elsevier, reproduced with permission from Elsevier [62].

material removal between two electrodes through electrical discharging. However, material removal through EDM process was developed during the era of World War II. Since then, there has been an upsurge in developing the EDM process, including the machining parameters' setting and the performance measures. One of the novelties of the EDM process over traditional machining techniques is its ability to shape any electrically conductive material. In addition, certain amount of material can also be deposited

on the machined surface, thereby enhancing its surface characteristics. EDM is demonstrated to be a hitch-free method that can automatically be used to produce a three-dimensional deep and complex cavity [99, 100]. The working principle of EDM is a thermo-electrical phenomenon which uses electricity and thermal energy to generate sparks and erodes both the electrodes' surfaces. A series of repetitive sparks are produced through a high supply voltage at a narrow gap between the electrodes. Thus, at each spark, a minute

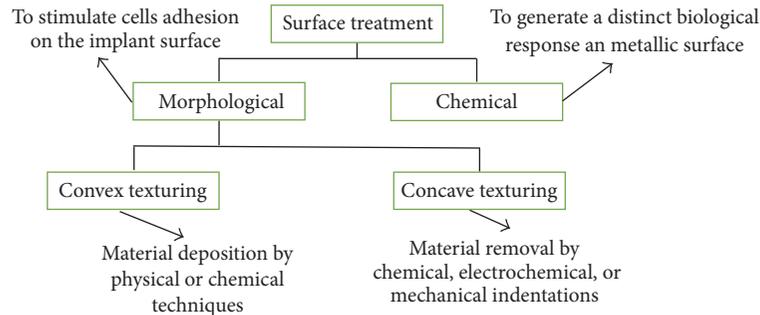


FIGURE 10: Classification of surface modification [21, 52].

particle of material is removed from the electrodes' material in the form of debris [101]. In most cases, the electrodes are immersed completely in the dielectric fluid, which is normally in the form of mineral oil or deionized water [102, 103]. The EDM performance depends strongly on the proper setting of both electrical (discharge current, discharge time, OFF-time, supply voltage, and tool-electrode polarity) and nonelectrical (flushing type, machining time, machining depth, dielectric medium, and electrode speed) machining parameters. Several authors have attempted to outline the EDM parameters and their influence on the machining performance. Pulse-on time, supply current, gap voltage, and powder concentration have been identified to have a major influence on the machining characteristics like material removal rate (MRR), tool electrode wear rate (TWR), and surface roughness (SR). However, a lot of studies have been made to optimize the performance of these parameters on different materials [104–111].

**4.1. EDM Dielectric Fluids.** The dielectric fluid plays a significant role in insulating the electrodes, clearing the debris from the machining gap, and cooling a very high temperature (800–1200°C) generated in the surface. Some EDM machines are customized with certain types of working fluids. However, in several situations, there is a need to select proper dielectric fluid. Generally, sinking EDM uses mineral oils while micro- and wire-cut EDM deal with deionized water [112]. These fluids have some benefits as well as limitations in different applications. For instance, in biomedical applications, deionized water stimulates the corrosion rate of EDMed implant surface, while oils based dielectric fluids burn and release carbons, which react to the alloying elements and form a very hard carbide surface. Leão and Pashby [113] reported the current trends on the use of environmentally friendly dielectric fluids and confirmed that water-based dielectric fluids will overrule the use of oil-based fluids. On the other hand, to mitigate the effect of dielectric fluid on health and environment, dry or semidry EDM was employed, whereby gasses such as oxygen and nitrogen are used as EDM working fluid [114–120].

**4.2. Evolution of EDM Applications.** In the past decades, EDM process is commonly used in the manufacturing of cutting tools, molds, and dies [110, 121, 122]. To reduce the cost

of oil dielectric fluid and the environmental pollution, dry (air) EDM was introduced. Kunieda et al. [123] established that EDM can be achieved in gas especially in machining 3D shape. Through dry EDM, small or zero tool-electrode wear with improved surface quality can be achieved [114]. Aside from material removal capability, EDM process is also considered as surface modification/alloying technique, due to the evidence of material transfer which occurs between the electrodes and the suspended dielectric fluid additives during the process [124–126]. The development of smaller (micro-size) products widens the applications of EDM to produce microcavities with high-aspect ratio [127], micro molds and dies [128], and micro rotational components [129]. The EDM productivity and the surface quality of the machined surface were greatly enhanced when dry EDM was augmented with the magnetic field [130, 131]. Thus, a wide application of EDM in the manufacturing of automobiles, aerospace, and nuclear components with a polished surface finishing which requires no secondary process has been recorded [9, 132–135]. Recently, EDM process evolved in the biomedical field owing to its ability to generate a mirror-like, extremely hard, and a biocompatible nanoporous surface through the phenomenon of material migration. A work presented by Peng et al. [19] shows a generation of nanoporous biocompatible layer on the EDMed Ti surface. Sales et al. [136] and Shabgard and Khosrozadeh [137] modify the surface of Ti-6Al-4V by adding calcium and carbon nanotubes in the dielectric fluids, respectively. The surface of the biodegradable magnesium alloy was improved when PM-EDM uses zinc powder [17]. The advancement of EDM process based on applications is summarized in Figure 11.

**4.3. Additive Mixed-Electrical Discharge Machining.** Using oil-based or deionized water as the EDM dielectric fluid is not enough to provide the satisfactory surface quality, especially in automobiles, aeronautics, and medical field, where a special surface tolerance is required. The substance added during AM-EDM may be grouped into three main classes. These are solid additives such as metallic powders, liquid additives like surfactant, urea solution, and calcium aqueous solution, or gaseous additives like nitrogen and oxygen gasses. The additives mixed with the dielectric fluid not only stabilize the EDM process but also improve the EDM efficiency and quality of the machined surface [138, 139]. The EDM performance, tool-electrode life, and quality of the

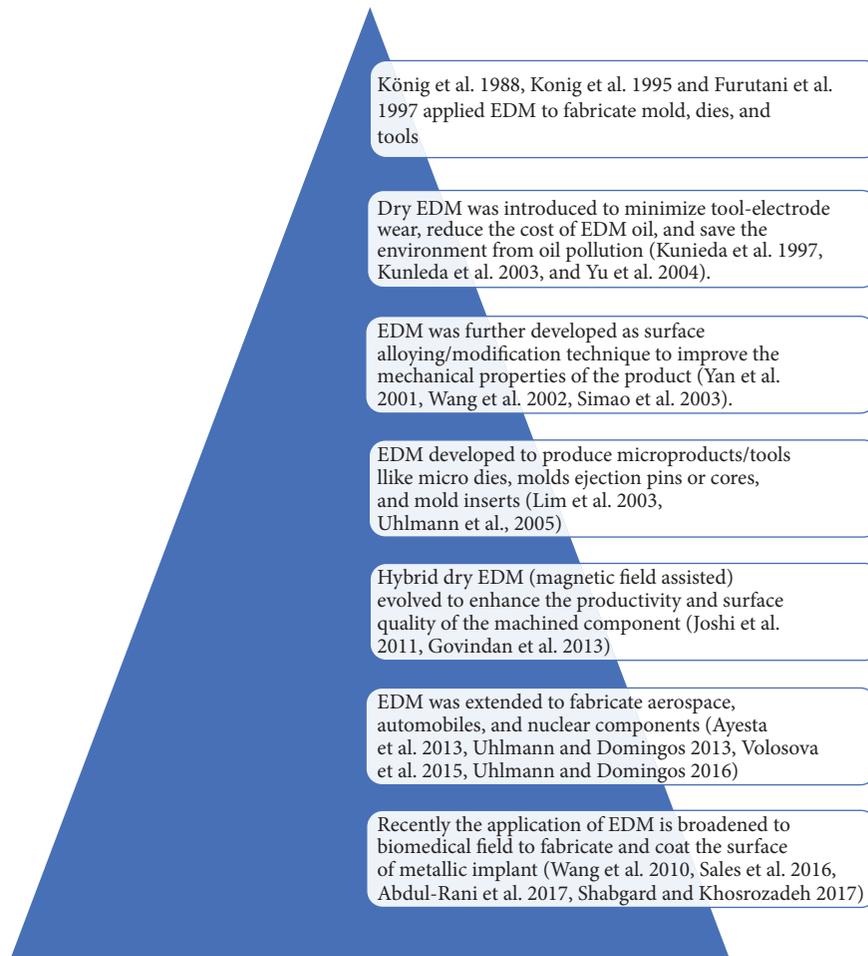


FIGURE 11: EDM process advancement based on applications.

machined component depend greatly on the additive type, size, concentration, and properties [140, 141]. Figure 12 shows a complete schematic diagram of AM-EDM.

**4.3.1. Solid Additives (PM-EDM).** The addition of solid substances to the dielectric fluid in the form of powder particles (PM-EDM) stabilizes the EDM process and improves its machining efficiency by decreasing the insulating strength of the suspended dielectric fluid and increasing the discharge gap. Several reports on adding different kinds of metallic powder have been documented. Al-Khazraji et al. [104] reported an improvement in the fatigue life and the recast layer thickness by adding SiC microsize powder to the dielectric fluid. The surface morphology and surface roughness of PM-EDMed AISI D2 steel were greatly enhanced when Ti nanopowder was added to the dielectric fluids [88]. Addition of chromium powder into the dielectric fluid notably improved the surface hardness and the surface finish of the machined tool steel during PM-EDM [142]. Singh et al. [143] studied the effect of EDM machining parameters without powder and with tungsten powder mixed dielectric fluid. Reduced recast layer thickness and enhanced surface finish were achieved after the PM-EDM of SiC. Bajaj et al. [144]

present a laudable review on the use of micro- and nanosize powder particles and revealed that the powder particles in nanoform are superior to microsize particles, in terms of surface quality and MRR improvement. An improved surface quality, MRR, and TWR were significantly observed when carbon nanotube was added to the dielectric fluid during the EDM of Ti-6Al-4V [137]. Figure 13 depicts SEM micrographs of surfaces machined with and without added powder. It can be observed that EDMed surface (Figure 13(a)) has thicker and irregular white layers when compared with PMEDMed surface (Figure 13(b)).

**4.3.2. Liquid Additives.** Liquid substances may be used purely or augment with the powder particles mixed with the dielectric fluid to enhance the efficiency of the EDM machine and improve the properties of the machined surface. Addition of surfactant to the dielectric fluid improved the overall EDM performance by increasing the conductivity of the suspended powder particles and reducing its agglomeration. This results in reduced SR and recast layer thickness of the modified titanium alloy as shown in Figure 14 [145]. Yan et al. [146] studied the influence of urea solution added to the water dielectric fluid during EDM of pure titanium. Urea mixed

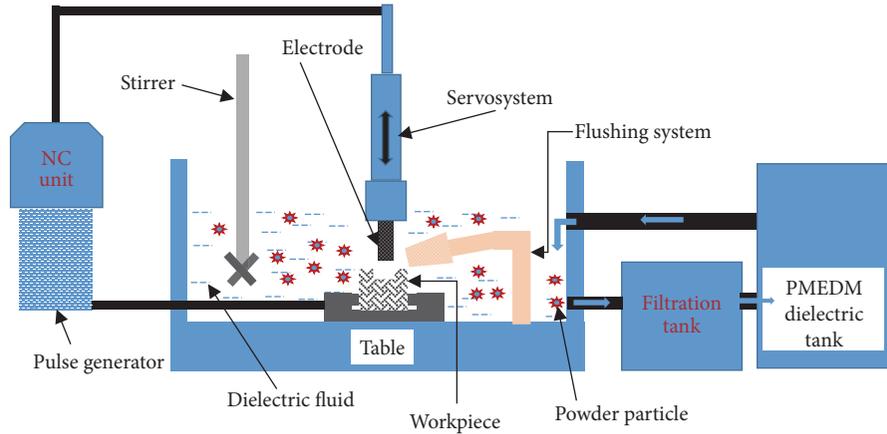


FIGURE 12: Schematic diagram of complete AM-EDM experimental setup.

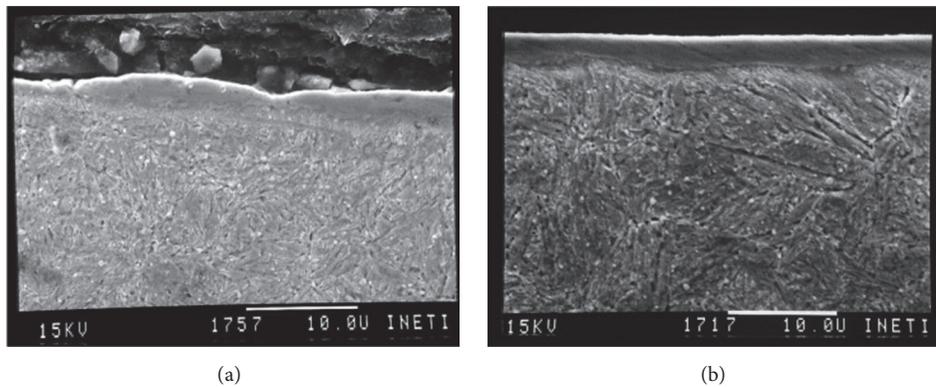


FIGURE 13: SEM micrograph of (a) EDMed (b) PMEDMed surfaces. Copyright @Elsevier, reproduced with permission from Elsevier [138].

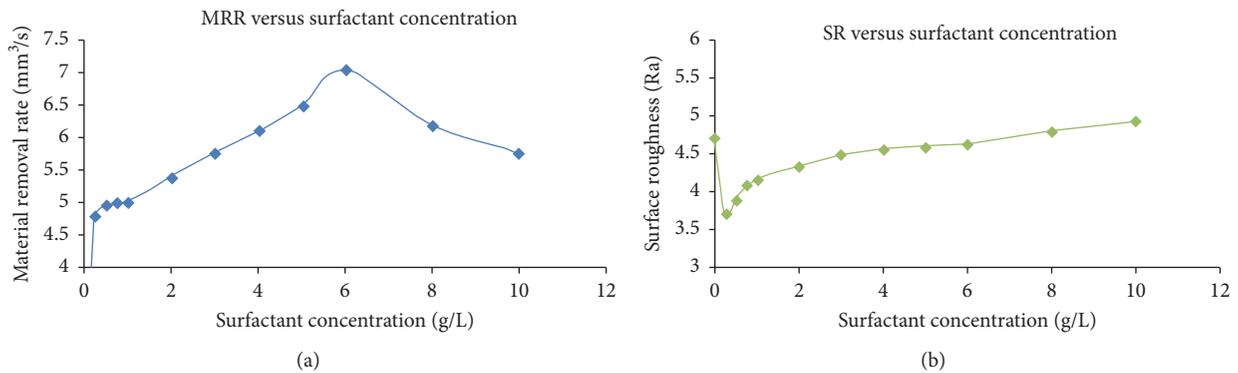


FIGURE 14: The effect of added surfactant on MRR and SR [38].

dielectric fluid decomposed and liberated nitrogen, which melted and migrated to the workpiece-electrode surface.

4.3.3. *Gaseous Additives (Dry EDM)*. To reduce cost of liquid (oil-based) dielectric fluid and overcome its environmental pollution and health hazard effect, dry EDM was introduced by using a gaseous substance like nitrogen, argon, or oxygen gas as a working fluid. Figure 15 shows a schematic diagram of oxygen mixed-EDM. The quality of the machined surface

and the rate of material removal depend on the type of the gas used [147]. For instance, machining with oxygen mixed dielectric fluids produces deeper and wider craters as depicted in Figure 16. During this process, a high-speed air is usually blown into the discharge gap which removes away the debris from the cutting zone. The greatest improvement in surface quality with discontinuous recast layer as well as a thin heat affected zone is recorded [120]. Yu et al. [114] used oxygen as dielectric fluids to assist in the machining

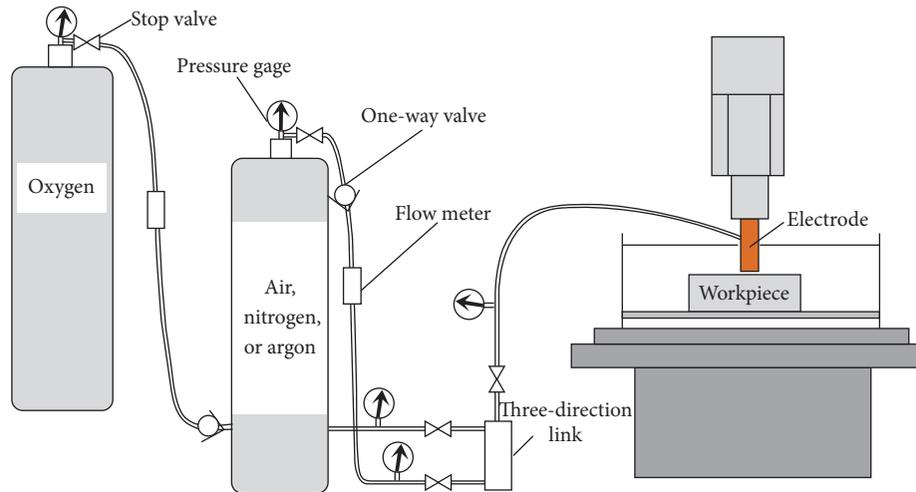


FIGURE 15: Schematic diagram of oxygen mixed-EDM. Copyright @Elsevier, reproduced with permission from Elsevier [147].

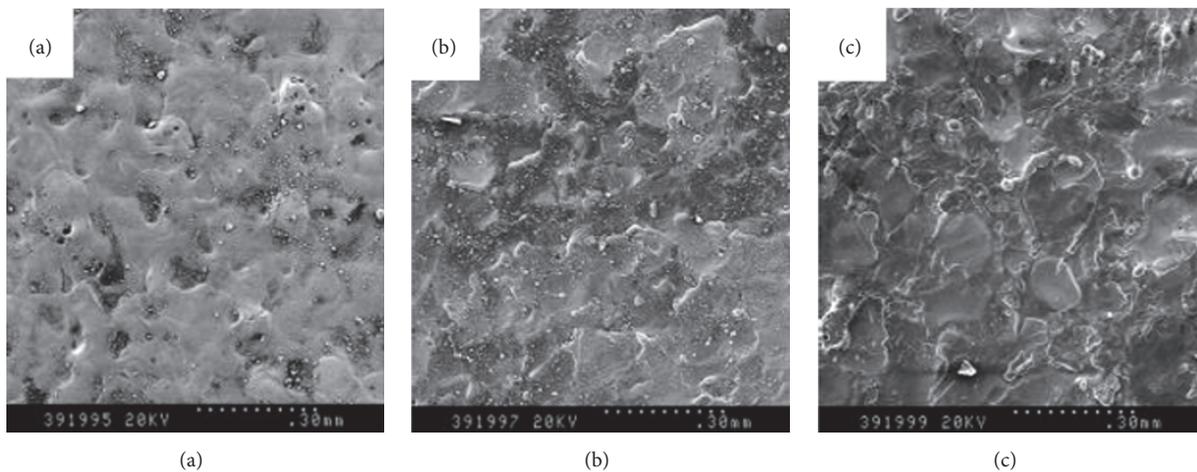


FIGURE 16: Surface topography of the machined samples in (a) argon, (b) nitrogen, and (c) oxygen. Copyright @Elsevier, reproduced with permission from Elsevier [147].

of cemented carbide. It was found that dry machining is more efficient, especially when cost and machining time are considered. In a study by Dhakar and Dvivedi [148], near-dry EDM (liquid and air) of high-speed steel produces a finer surface, negligible tool-electrode wear, and thinnest recast layer when compared to conventional EDM. During dry EDM, the debris in the machining gap can be easily removed, thereby overcoming the problem of its reattachment to the electrodes surfaces. Huang et al. [149] reported the use of nitrogen gas as the dielectric fluids. A nanostructured surface in the form of TiN and CrN with extreme hardness and excellent adhesion was found.

### 5. AM-EDM as Potential Surface Treatment Technique for Biomedical Implants

The applications of EDM are broadening from mere tools and dies, to fabrication and shaping of automobiles, aerospace,

nuclear components, and, recently, to fabrication and surface treatment of biomedical implant. Despite the complexity of the EDM process, several studies have revealed the wide potentialities of this process in the biomedical field, especially with regard to the fabrication and surface modification of metallic implant. This is evident when a significant material transfer from the tool electrode and the suspended dielectric additives to the workpiece-electrode surface was observed during the EDM process. This serves as a strong route of providing a well finished and qualitative machined surface. In addition to generation of extremely hard, nanofinished, and nonporous surface, EDM produces a bioactive and a biocompatible layer on the implant surface through the concept of material migration.

*5.1. Mechanism of Material Migration.* During the EDM process, materials from the tool electrode and the additives in the dielectric fluids are transferred and deposited on the

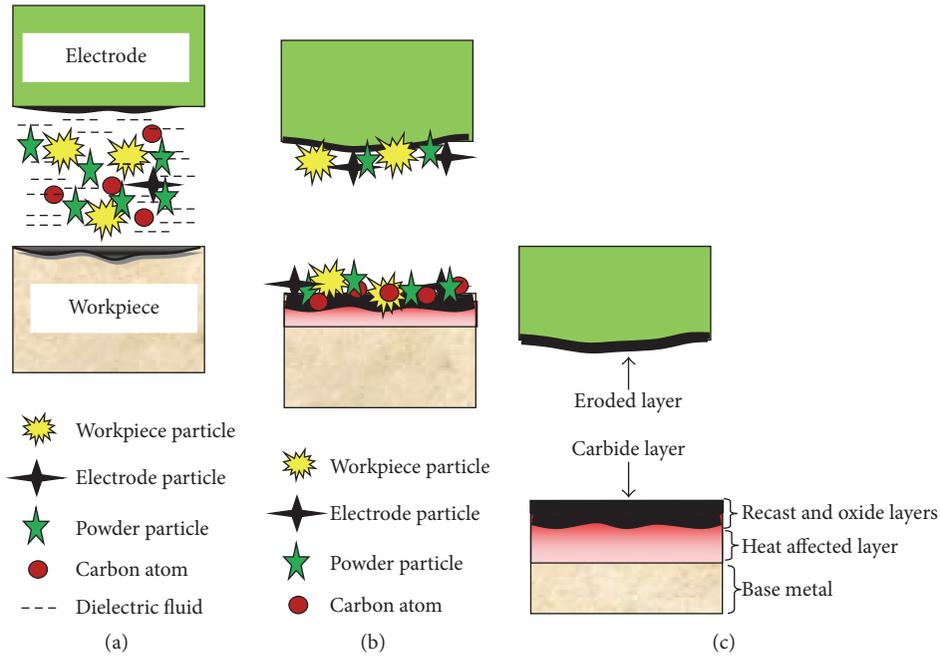


FIGURE 17: Mechanism of material migration and deposition. (a) Initiation of spark. (b) Carbon atoms, powder, and tool-electrode particles adhered on the electrodes surfaces. (c) Formation of three distinct layers.

workpiece-electrode surface. Batish et al. [39] investigated the mechanism of material transfer during EDM of three different die steel samples, thus confirming a reasonable deposition of tool-electrode material and powder particles on the EDMed surface. Gülcan et al. [150] reported a substantial deposition of tool-electrode materials on the workpiece-electrode surface after AM-EDM. Thus, a very hard surface with strong abrasion and corrosion resistance surface was achieved. It is believed that material transfer by AM-EDM altered the chemical content, mechanical properties, and the metallurgical structure of the machined surface, thereby forming different layers on the AM-EDMed surface [40]. The topmost layer is the extremely hard carbide layer, which is formed due to the reaction of carbon in the dielectric hydrocarbon fluid and the workpiece-electrode alloying elements. The recast and oxide film constituted the middle layers. The recast layer is produced because of resolidification of unflushed metallic debris in the craters. On the other hand, biocompatible, nanostructured, and nanoporous oxide layer is developed due to fusion of oxygen (foreign element) and the elemental constituents in the work material. The last layer is the heat affected zone (heated but nonmelted layer). This layer is formed due to heating and subsequent quenching during the EDM process. The characteristics of these layers mainly depend on the type of electrodes material, powder particles, dielectric fluid, and the setting of the machining parameters [151]. The schematic illustration of material migration mechanism and material deposition during AM-EDM is displayed in Figure 17. In the first stage, a spark is generated through a powder suspended dielectric fluid. In this phase, the electrodes materials are eroded, forming a large crater. The detached electrodes particles, powder

particles, and the carbon atoms migrated to the electrodes surfaces as shown in Figure 17(b). The final stage is shown in Figure 17(c); whereby all layers are completely formed, with no or shallow craters on the AM-EDMed surface. However, the tool-electrode size reduced due to erosion of the contact surface.

**5.2. Formation and Influence of Recast Layer on EDMed Implant.** Although a hard and adhesive recast layer is expected to enhance the corrosion and wear resistance of the metallic implant, a thick recast layer should be avoided or polished, to reduce the risk of the implant mechanical failure [149]. The recast layer thickness largely depends on the dielectric fluid type, the pulse-on time, and the powder concentration [11, 38]. EDM with kerosene as dielectric fluid produces thinner and porous recast layer as well as finer and harder surface when compared to water emulsion. It was also reported that the white layer thickness decreases with a decrease in pulse duration (electrical energy) [152]. In addition, an EDM automatic finishing setting and the addition of powder to the dielectric fluid reduced the formation of a recast layer [11, 153, 154] as depicted in Figure 18.

**5.3. Biocompatible and Bioactive Oxide Layer Formation.** For long-term implantation of the biomedical implant, biocompatibility and bioactivity of such material are the key concern. A mechanically strong bonding between the implant and the body tissues which occurs due to biophysical and biochemical reactions is referred to as bioactive. Various biomaterials have different interfacial bonding time, strength, and bonding thickness [155]. Bioactive glass, glass ceramics, and hydroxyapatite are the most common and clinically

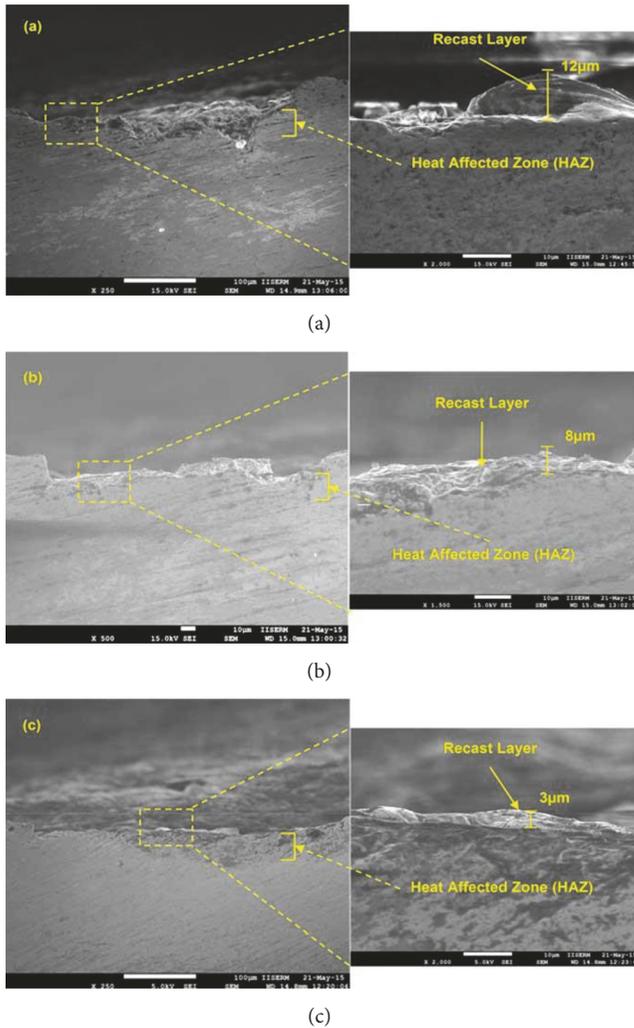


FIGURE 18: It shows a microscopic cross section of AM-EDMed  $\beta$ -phase Ti-alloy at Ip (5 A), Ton (10  $\mu$ s), and Si-powder concentration of (a) 0 g/L, (b) 2 g/L, and (c) 4 g/L [11].

approved bioactive materials [156]. A good bioactive layer is expected to generate a bone-like structure and bone-like mechanical properties which will allow for tissue growth and attachment. Implant biocompatibility and bioactivity do depend on not only the material properties but also the implant manufacturing and coating techniques which the implant undergoes. AM-EDM is a novel technique which can be used to deposit a biocompatible and bioactive layer on the implant surface. One of the approaches of depositing these layers is by mixing the EDM dielectric fluid with a bioactive powder. Another way is to use a bioactive tool electrode. The last approach is to employ both powder and the tool electrode. A nanoporous and calcium-rich oxide layer is produced on Ti-6Al-4V surface when 0.1 Mol/L of calcium chloride aqueous solution was added to deionized water during sinking EDM [136]. This oxide surface was proved to have high hardness and reasonable biocompatibility, enough for biomedical applications, especially for bone and teeth. A SEM micrograph of EDMed Ti-6Al-4V surface with deposited

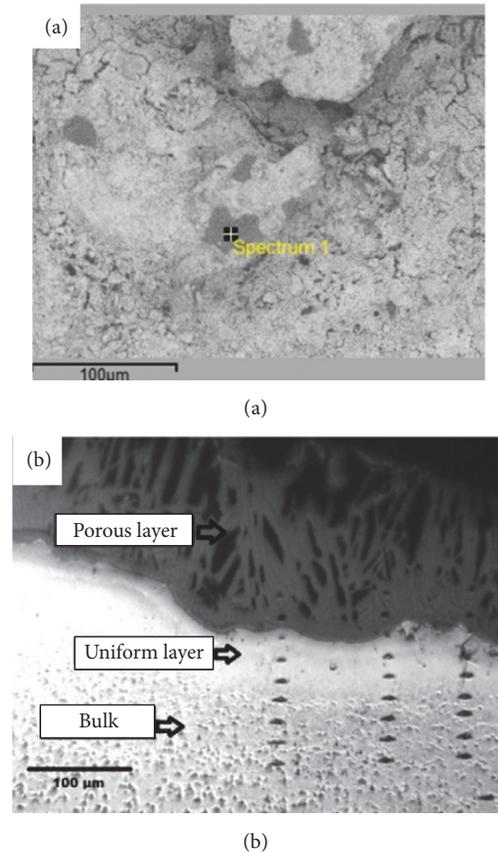


FIGURE 19: SEM micrograph: (a) Ti-6Al-4V after calcium deposition by AM-EDM; (b) uniformly distributed porous layers. Copyright @Elsevier, reproduced with permission from Elsevier [136].

calcium is shown in Figure 19(a) and a porous surface in Figure 19(b). A surface of Ti-6Al-4V treated via an EDM process by Strasky et al. [157] shows a substantially higher number of cells when compared with a commercial plasma spray coating technique as displayed in Figure 20. Thus, it can be stated that the substrate modified through AM-EDM produces exceptional substrate for the adhesion and growth of human bone compared to the alloy plasma sprayed. A similar research conducted by Yang et al. [158] revealed a high cells attachment and proliferation on EDMed samples when compared with the untreated samples. The in vitro cell response of the treated and untreated substrates is depicted in Figure 21. Another study by Otsuka et al. [159] revealed a great rise of the cell attachment on the EDMed surface while the cell proliferation remained restrained. However, the number of cells was found to be increased. Chen et al. [91] studied the surface modification of Fe-Al-Mn alloy through AM-EDM. The experimental results presented a nanostructured recast and oxide layers which greatly enhanced the biocompatibility and osteointegration of the machined alloy. A nanoporous TiO<sub>2</sub> formed during EDM of titanium is believed to be responsible for the generation of bioactive and biocompatible layer on the machined sample [19]. A study on the EDM of  $\beta$ -Ti-alloy presented by Prakash et al. [160] revealed a

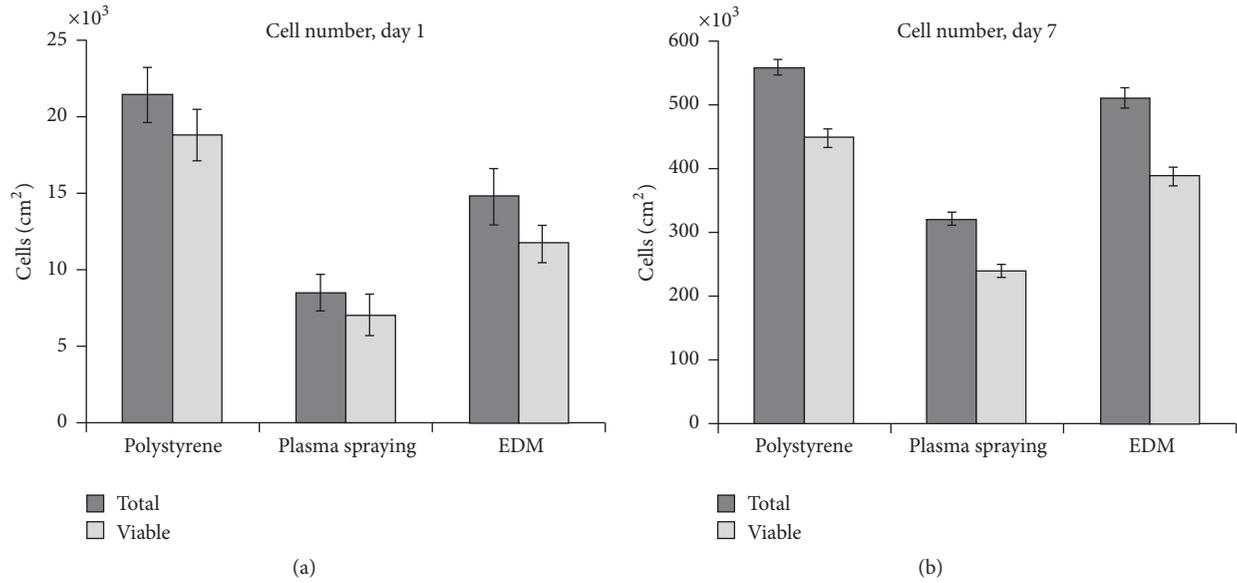


FIGURE 20: A quantity of human osteoblast-like cells for a Ti-6Al-4V alloy treated by plasma-spraying with TiO<sub>2</sub> and EDM seeded on standardized cell culture polystyrene dishes for day 1 and day 7 [157].

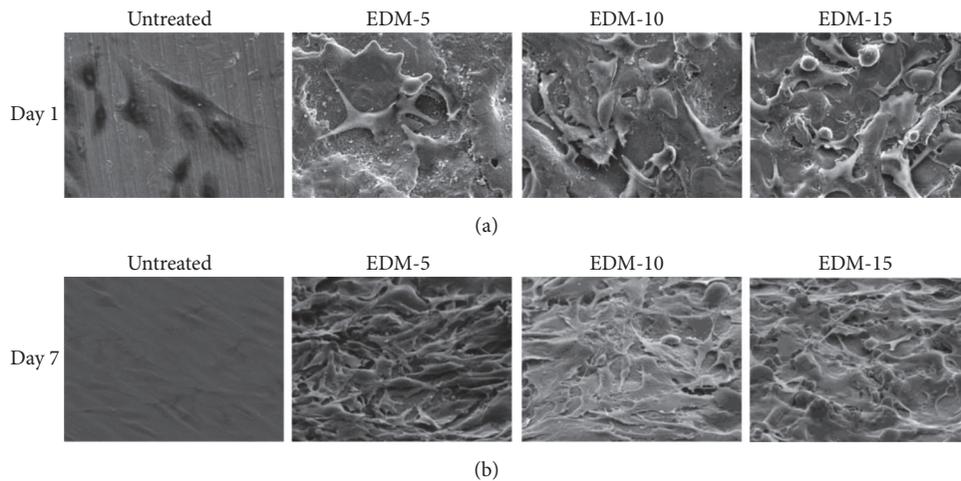


FIGURE 21: SEM micrograph of the in vitro MG-63 cells attachment and proliferation on the treated and untreated samples for (a) day 1 and (b) day 7 [158].

higher cell growth and adhesion on the machined surface. In similar studies, carbide and oxide layers which enhance the biocompatibility of  $\beta$ -Ti-alloy were achieved when Si-powder was added to the EDM dielectric fluid [11]. It was also proven that the nanoporosities fabricated by PM-EDM process promote the cell growth and adhesion [161, 162]. A nanofinished, bioactive, and biocompatible surface was achieved during PM-EDM of Ti-alloy [163]. The XRD patterns showing the titanium oxide and nitride layer deposition after electrodischarge coating are elaborated in Figure 22.

**5.4. Hard and Nanoporous Carbide Layer Formation.** The quest for long-term capability of orthopedic or dental implant after implantation remains indispensable. Several reports confirm the premature failure of the implanted hard tissue

such as bone and teeth as elaborated in Section 3.1. Poor corrosion and wear characteristics of the implant material like stainless steel, Co-Cr, and Ti alloys are considered as the root cause of this problem [60–63, 66]. However, formation or deposition of hard carbide layer on the implant surface remained the key remedy to this issue. A carbide film can be generated on the substrate surface by using carbide containing tool electrode, powders, or dielectric fluid. The carbon released by these substances can easily react to the alloying elements in the workpiece-electrode material during AM-EDM process. Algodí et al. [164] investigated the deposition of TiC-Fe layer on the stainless steel using TiC tool-electrode. The results revealed a high deposition of TiC carbide film on the EDMed surface. This carbide coated surface was characterized by a high but variable hardness

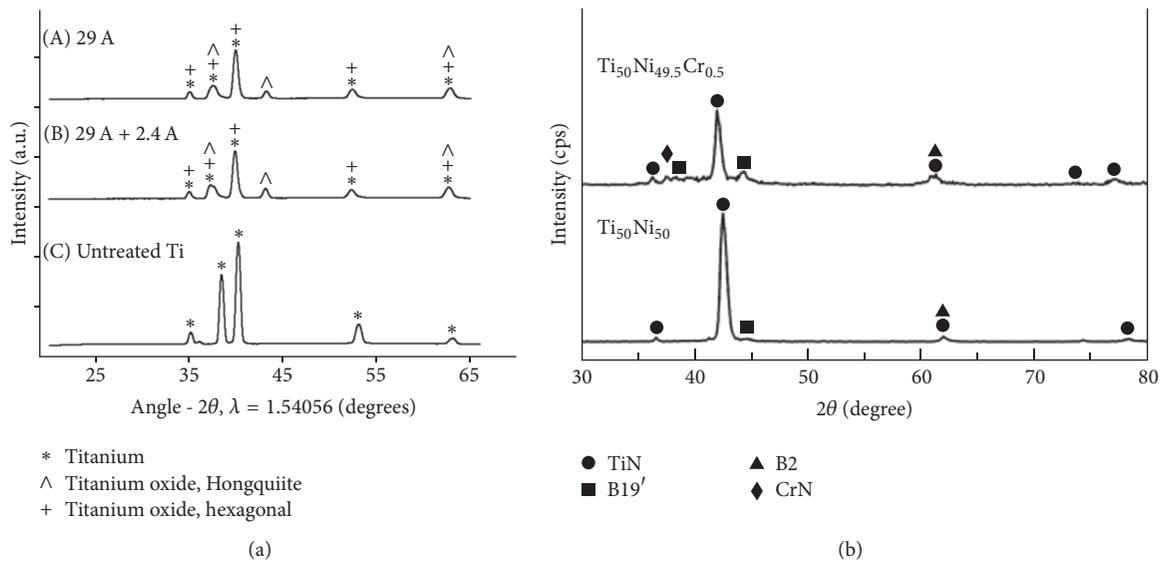


FIGURE 22: It depicts XRD patterns showing titanium oxide and nitride layers deposition after electrodischarge coating. Copyright @Elsevier, reproduced with permission from Elsevier [87, 149].

which may be related to variation of the carbide content on the modified surface. A study presented by Prakash et al. [165] confirmed the production of an extremely hard TiC layer with good wear characteristics during EDM of  $\beta$ -Ti for orthopedic use. The deposition of titanium carbide on the carbon steel work material using graphite powder was successfully achieved [166]. It is found that the added powder not only enhanced the uniformity of the coated layer and the microcracks but also contributed more carbon in the dielectric fluid. In addition, the hardness and wear resistance of the workpiece-electrode material were greatly enhanced. Li et al. [167] observed the Ti-6Al-4V EDMed surface behavior by mixing silicon carbide abrasives with dielectric fluid. The results revealed the formation of carbide film, which is characterized by more than twofold rise in hardness and greatly improves wear resistance. Clear boundary and fine finishing are also noticed. In similar studies, Kuriachen and Mathew [168] reported the formation of hard silicon and tungsten carbide layers on the machined Ti-6Al-4V when SiC powder was mixed with the dielectric fluids using tungsten carbide tool-electrode. Table 6 critically reviews the recent studies for material deposition of metallic implant through the AM-EDM process.

## 6. Challenges and Future Studies

Besides various potentialities of AM-EDM in manufacturing industries and the medical field, there are several issues which are yet to be solved to qualify the application of the EDM process at the clinical stage. These are summarized as follows:

- (i) It is very difficult to control and measure the actual thickness of the layer deposited by AM-EDM. An extensive study in this direction will be helpful.
- (ii) It is quite challenging to set an optimum powder concentration for achieving a required coated layer

size on the implant surface. Thus, there is a need for more researches that will focus on setting the optimum powder concentration for surface coating through AM-EDM.

- (iii) The mechanism of material deposition during AM-EDM is not clearly understood, owing to the complexity of the process. However, future studies need to concentrate on material deposition monitoring during the machining process.
- (iv) There is a great challenge in controlling the uniformity of the deposited layer on the AM-EDMed implant surface, which is supposed to provide sufficient and uniform implant-bone bonding.
- (v) AM-EDM parameter settings depend on the material type. Thus, setting optimum parameters that will suit all material combinations remained challenging.
- (vi) Materials are melted and subsequently quenched during AM-EDM. Thus, there is a considerable risk of transforming the material structure into different phases. This might affect the initial performance and mechanical properties of the substrate materials. Thus, extensive studies are required in this segment.

## 7. Summary

For the long-term performance of an implant, a fast bone-tissue interaction is believed to be dependent on surface characteristic and properties of the deposited layer by the surface modification technique. It has been found that the implant with HA coating has enhanced the bioactivity and promotes the bone-tissue growth and osseointegration at a faster rate. Accordingly, numerous surface treatment/modification techniques like chemical vapor deposition (CVD), physical vapor deposition (PVD), iodization, and laser deposition

TABLE 6: Critical review of recent development of material deposition and surface treatment of metallic implant through AM-EDM.

Electrodes combination	Powder type/conc.	Deposited layers	Findings	Refs.
Grade IV Ti/Ti	Ti/3 & 6 g/L	(i) TiO (ii) Recast layers	(i) With 3 g/l microcracks observed and disappeared when using 6 g/l of Ti (ii) Recast layer thickness increases with increase in current duration and Ti powder concentration (iii) Hydrophilic surface good for dental implant was observed	[86]
Grade II Ti/Ti	—	(i) TiO	(i) Dual surface topography with micron and submicron topographies sufficient for orthopedic and dental applications	[87]
AISI D2 steel/Ti	Ti/2 g/L	(i) TiC	(i) Significant Improvement in surface roughness (ii) Shallow craters and craters diminishes (iii) Ti atoms higher in the cracks area	[88]
$\beta$ -Ti alloy (Ti-Na-Ta-Zr)/Cp-Ti	Si/0, 2, 4, 6 & 8 g/L	(i) Bioceramic oxides (ii) Carbide phases	(i) EDMed surface hardness twice the pure sample (ii) 4 g/L of Si powder produces thin recast layer (iii) A nanoporous, bioactive and biocompatible surface achieved	[11]
Ti-6Al-4V/Ti	HA/5, 10, 15 & 20 g/L	(i) HA	(i) AM-EDM can be used as alternative for HA reach layer deposition on the substrate (ii) AM-EDM produces biocompatible coating for medical applications	[89]
Ti-6Al-4V/Cu	Surfactant + Graphite (Gr) 1–20/gL	(i) Recast layer	(i) RLT layer reduced with 6 g/L of surfactant which becomes evenly distributed with 10 g/L of Gr (ii) Less microcracks were also observed (iii) Surfactant inhibit a powerful material migration	[38]
Ti-6Al-4V/Ti	HA	(i) HA	(i) Moderate pulse-on current and pulse-on duration are possible settings that will produce material deposition	[90]
Grade IV Ti/Cu	—	(i) TiO <sub>2</sub> (ii) Nano TiH	(i) A nanoporous, nanostructured and bioactive TiO <sub>2</sub> layer with short duration (ii) Improved biocompatibility was achieved	[19]
Fe-Al-Mn/Cu	—	(i) Recast layer (ii) Oxide layer (iii) k-carbide phase	(i) Nanostructured recast layer was formed (ii) Increased biocompatibility	[91]
WC90-Co10/Cu	Ti	(i) TiC	(i) Improved hardness with reduced microcracks	[92]

were used to improve the implant stability. However, typically, the coating techniques used for the surface treatment have a few key problems, (i) very thin coating layer and (ii) weak adhesion and bond strength with the substrate, which can deteriorate after some time due to acidic nature of body fluid, causing the implant failure. In contrast to this, the electric discharge machining (EDM) shows better potential for surface treatment of metallic implants. The role of EDM process for surface modification of biomaterials is still at the experimental stage. A few number of research studies have been reported by taking hydroxyapatite powder mixed-EDM as surface modification process for biomaterials; superior biocompatible surface characteristics have been reported and capability of formation of biocompatible nanoporous surface has been reported. This paper reviewed the recent development of AM-EDM and its application as a potential technique for the surface treatment of orthopedic and dental implants. Deposition of bioactive and biocompatible layers on the implant surface through AM-EDM process depends on various factors which include appropriate selection of

electrodes, powder, dielectric fluid, and optimum EDM parameters setting. In addition, AM-EDM can be used to generate a mirror-like finishing and extremely hard surface on the metallic implant surface. However, the powder, dielectric, and electrodes material type are found to have great influence on the thickness and quality of the coated surface. Despite this, there are a gap and the urgent need to study the effect of various input independent parameters of AM-EDM on the machining of magnesium implant, metallic glass, and new  $\beta$ -Ti materials.

### Conflicts of Interest

The authors declare that they have no conflicts of interest.

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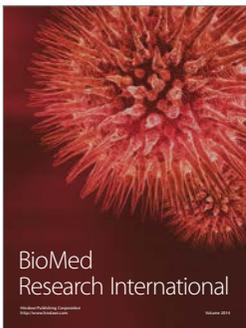
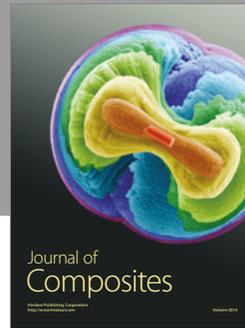
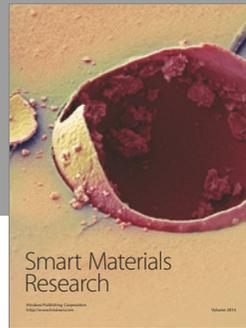
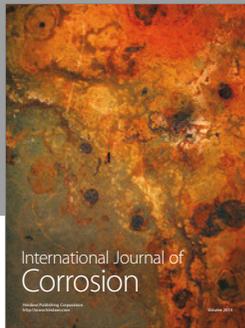
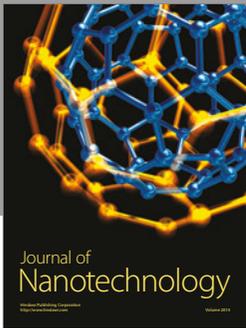
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