

**SURFACE MODIFICATION OF β -PHASE TITANIUM
ALLOY BY BALL BURNISHING ASSISTED
ELECTRICAL DISCHARGE CLADDING FOR
IMPLANTS**

Thesis Submitted For the Award of the Degree of
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(Mechanical Engineering)

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DECLARATION

I, hereby declared that the presented work in the thesis entitled “SURFACE MODIFICATION OF β -PHASE TITANIUM ALLOY BY BALL BURNISHING ASSISTED ELECTRICAL DISCHARGE CLADDING FOR IMPLANTS” in fulfillment of degree of **Doctor of Philosophy (Ph.D.)** is outcome of research work carried out by me under the supervision of Dr. Hitesh Vasudev, working as Professor, in the School of Mechanical Engineering at Lovely Professional University, Punjab, India. & under the Co-supervision of Dr. Chander Prakash, Mechanical Engineering Professor at Chandigarh University, India. In keeping with the general practice of reporting scientific observations, due acknowledgements have been made whenever the work described here is based on the findings of other investigators. This work has not been submitted in part or full to any other University or Institute for the award of any degree.

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Certificate

This is to certify that the work reported in the Ph.D. thesis entitled “SURFACE MODIFICATION OF β -PHASE TITANIUM ALLOY BY BALL BURNISHING ASSISTED ELECTRICAL DISCHARGE CLADDING FOR IMPLANTS” submitted in fulfillment of the requirement for the award of degree of Doctor of Philosophy (Ph.D.) in the School of Mechanical Engineering, is a research work carried out by Rahul Wandra, Registration No. 41800332, is a bonafide record of his original work carried out under my supervision and no part of this thesis has been submitted for any other degree, diploma or equivalent course.

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Abstract

Biomaterials have been extensively utilized for many years to interact with the biological system, restoring the form and functionality of natural organs or tissues in the body. There are two main groups of biomaterials: synthetic and natural. This is mainly due to their excellent mechanical properties and strength. Biomedical implants made from metallic biomaterials like stainless steel (SS-316L), cobalt-chromium (Co-Cr), nickel (Ni), niobium (Nb), magnesium (Mg), titanium (Ti) and its alloys are given huge importance in biomedical industry. In the recent investigations in the field of research, titanium (Ti) and its alloys have gained increasing popularity in the biomedical field due to their impressive thermal conductivity, mechanical attributes, and biocompatibility. The present research work is showing the development of low modulus based β -type titanium alloys (Ti-35Nb-7Ta-5Zr) with value approximately 50 GPa (near to bone), for the fabrication of orthopaedic implants especially total hip and knee joint implants. However, despite its exceptional qualities like a high strength to mass ratio and excellent fatigue endurance, low density and high corrosion resistance as compared with the other metallic biomaterials, β -Ti alloy does not effectively meet the requirement of osseointegration.

(EDM) is considered as a better and long-lasting alternative, to modify the surface of implants for biomedical application. This is mainly because of its ability to produce rough and nano porous surface layer, which enhances biocompatibility, osseointegration, wear and corrosion resistance. However, EDM surface comprised poor surface quality and leads to degradation of corrosion performance and fatigue performance of implant, thus causing joint failure in long term performance. So to overcome above mentioned drawbacks of EDM, an in-house ball-burnishing assisted electrical discharge cladding (BB-EDC) was developed. In (BB-EDC) electrical discharge cladding is combined with ball burnishing machining as a post finishing operation.

In the present research work, β -Ti (53Ti-35Nb-7Ta-5Zr) implant was subjected to a novel surface modification technique using BB-EDC. The main aim of the study is to synthesize a functionalized TiO_2 loaded Hydroxyapatite (HAP) surface to refine

corrosion resistance, wear resistance, fatigue performance and bioactivity of β -Ti. Experiments were performed on BB-EDC setup using a rotary tubular electrode (15mm diameter) made up using commercially pure titanium. The electrode is connected with an arrangement to route pass high compressed slurry toward workpiece, at a pressure of 15 MPa. The cladded layer is composed of blended mixture of TiO_2 (30 wt%) & HAP (70 wt%) of size ranging from 0.110 – 0.200 μm . The slurry required for the purpose of cladding is prepared when the blended mixture (TiO_2 + HAP) is mixed with distilled water at a concentration of 30 g/l. In order to reduce these surface defects, ball-burnishing process using two hard ceramic (ZrO_2) balls of diameter 5mm were equipped or were supported with the help of a tool that rotates in a circular path and subjected a compressive force on the workpiece surface. This ball-burnishing process results in to flattening of cladded surface and produced a dense and compact layer. Taguchi's design of experimental technique was used for organizing the parameters and the levels at which they should vary.

As per the desired level of input process parameters, a mix-mode L-18 orthogonal array (OA) was selected for the DOEs. The impact of main parameters like peak current (I_p), pulse duration (T_{on}), supply voltage (V) and powder concentration (Pc) on surface integrity of β -type titanium alloy has been investigated. Commercially pure titanium (Cp-Ti) alloy based electrode is used as tool electrode in our setup and powder concentration of TiO_2 loaded HAP used is 0 g/l for EDM and 30 g/l for BB-EDC. The average surface roughness measured after the BB-EDC process was around 1.10 μm , this roughness value was less than half of the value obtained through EDM process. This is accountable because, hard ceramic balls (ZrO_2) in BB-EDC, produces compressive forces on the machined surface of specimen, which eliminates the surface irregularities, cracks, and compressed the high ridges of redeposited.

Analysis of variance (ANOVA) technique was utilized to determine the significance level of input parameters while affecting the surface roughness. The process type has the highest contribution (59.28%), the peak-current (10.76%), pulse duration (16.89%), supply- voltage (8.70%) and residual error (4.37%), respectively. As per the best optimal conditions, the parametric combination like the process (BB-EDC), the peak-current (5 A), pulse duration (50 μs), supply voltage (60 V), were found to yield the preferred results. At optimized conditions, the minimum surface roughness was

measured around 0.61 μm . The application of ball burnishing resulted in plastic deformation of the formed layer, leading to the creation of a refined microstructure, which subsequently raised the surface hardness from 285 HV to 848 HV.

Surface hardness was assessed utilizing a Vickers microhardness tester, applying an indentation load of 2.25 N and holding it for 10 seconds. The BB-EDC-treated surface achieved superior fatigue strength of 490 MPa compared to untreated 385 MPa and low corrosion current value $I_{\text{corr}} = -8.18 \mu\text{A}/\text{cm}^2$ compared to the untreated. The BB-EDC modified surface exhibited remarkable superiority in supporting the growth and proliferation of osteoblastic cells. This is attributed to the cladded layer, which contains evenly dispersed TiO_2 -loaded HAP, providing exceptional surface energy that promotes uniform cell growth across all surfaces and initiates faster proliferation.

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(Rahul Wandra)

List of Abbreviations

Abbreviation	Expansion
ANOVA	Analysis of Variance
BM	Biomaterial
BB-EDC	Ball Burnishing Assisted Electrical Discharge Cladding
β -Ti (53Ti-35Nb-7Ta-5Zr)	β -Titanium alloy
CC	Chemical Composition
CP-Ti	Commercially Pure Titanium
C.O.F.	Coefficient of Friction
Icorr	Corrosion Current Density
Ecorr	Corrosion Potential
EDS/EDX	Energy Dispersive X-ray Spectroscopy
EDM	Electrical Discharge Machining
EDC	Electrical Discharge Cladding
FT	Fatigue Testing
FE-SEM	Field Emission Scanning Electron Microscope
IV	In Vitro/ In Vivo
E	Modulus of Elasticity
OA-L18	Orthogonal Array
OFAT	One Factor At a Time
I _p	Peak Current
T _{on}	Pulse Duration
RLT	Recast Layer Thickness
RPM	Rotation Per Minute
HV	Surface Hardness
SCD	Surface Crack Density
Ra	Surface Roughness
V	Supply Voltage

S-N curve	Stress Life Curve
TJR	Total Joint Replacement
TiO ₂ +HAP	Titanium Dioxide Loaded Hydroxyapatite
UTM	Universal Testing Machine

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1.1 Overview (Necessity for Joint Replacement)

In recent decades, the world's population has seen an upward trend in longevity and life expectancy, leading to a rise in the average age of the global populations. It is estimated that around 90% of people aged 40 and above suffer from degenerative and inflammatory conditions, including osteoporosis, osteoarthritis, rheumatoid arthritis, and chondromalacia.

According to (Nag & Banerjee, 2012), these degenerative disorders cause discomfort and a loss of functioning by causing the mechanical properties of bone to deteriorate either due to excessive loading or because the normal biological self-healing process is absent. Total joint replacement (TJR) stands as the sole remedy for these problems, involving the replacement is done to ensure enduring relief from pain and enhanced mobility (Long & Rack, 1998).

According to (Batchelor & Chandrasekaran, 2004), a biomedical implant is an artificial organ used to replace a damaged organ adversely affecting other connected bodily organs. Biomedical implants have a wider range of applications in various parts of the human body. These implants improve the quality of the individuals lifespan (Geetha et al., 2009). Figure 1.1 illustrates the expertise in standard hip and knee replacement. Therefore, there is a pressing requirement for enhanced biomaterials and manufacturing techniques in the area of biomedical implants.

Excellent biocompatibility, remarkable corrosion resistance, a remarkable balance of high strength with low modulus, exceptional fatigue and wear resistance, and non-toxicity to cells are all necessary for orthopaedic implants, especially those meant for load-bearing applications (Geetha et al., 2009; Rack & Qazi, 2006).

Furthermore, with the rise in life expectancy, frequently used materials such as stainless steels, Co-Cr alloys, and Ti-based alloy have demonstrated a tendency to fail over prolonged periods of usage. Notably, not only have replacement surgeries increased, but so too have revision surgeries for hip and knee implants.

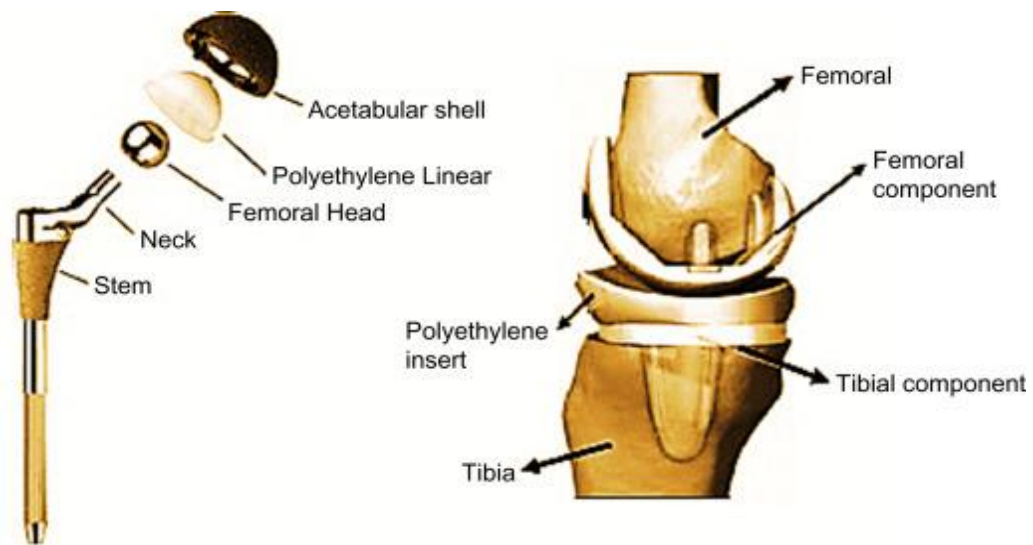


Figure 1.1 Expertise in total hip and knee joint implant replacements (Geetha et al., 2009)

The various reasons necessitating revision surgery are depicted in the Figure 1.2.

1.2 Biomaterials

Biomaterials are substances, either synthetic or natural, that interface with biological entities. These materials are employed to ensure bodily function (Bartolo et al., 2012).

1.2.1 Classes of Biomaterials

Biomaterials can be categorized into two primary groups: synthetic materials and natural materials (Bao Ha et al., 2013).

Figure 1.3 illustrates the commonly acknowledged exemplary features of functional metal-based biomaterials, including Ti and its alloys, Co-Cr alloys, and stainless steels.

1.2.1.1 Application and uses of Biomaterial

Biomaterials have applications in medical industry. Table 1.1 below gives an overview of how biomaterials are used in different areas of medical practice. The field of biomaterials continue to progress, offering creative solutions in these and other medical fields. Researchers are persistently developing new materials and refining existing ones to boost their biocompatibility, durability, and functionality for various medical applications.

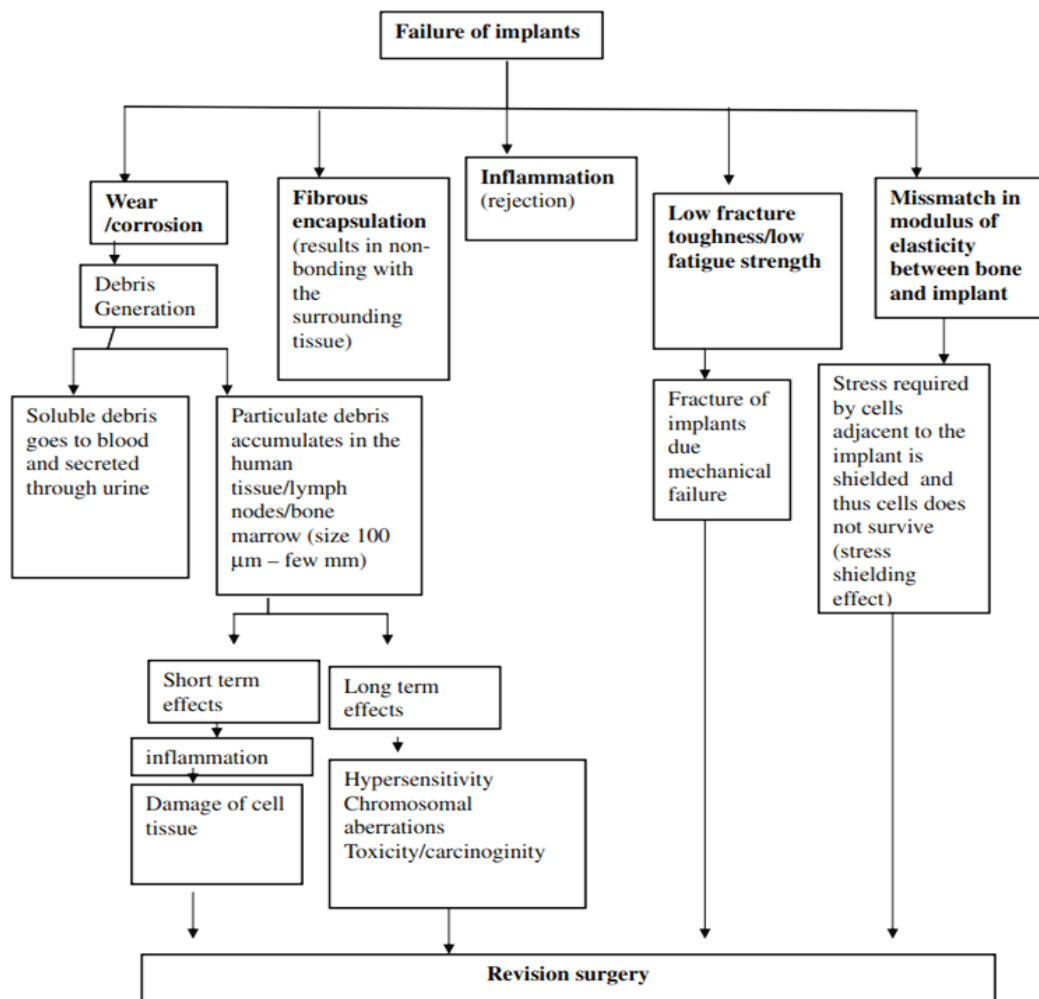


Figure 1.2 Various causes for failure of implants (Li et al., 2014)

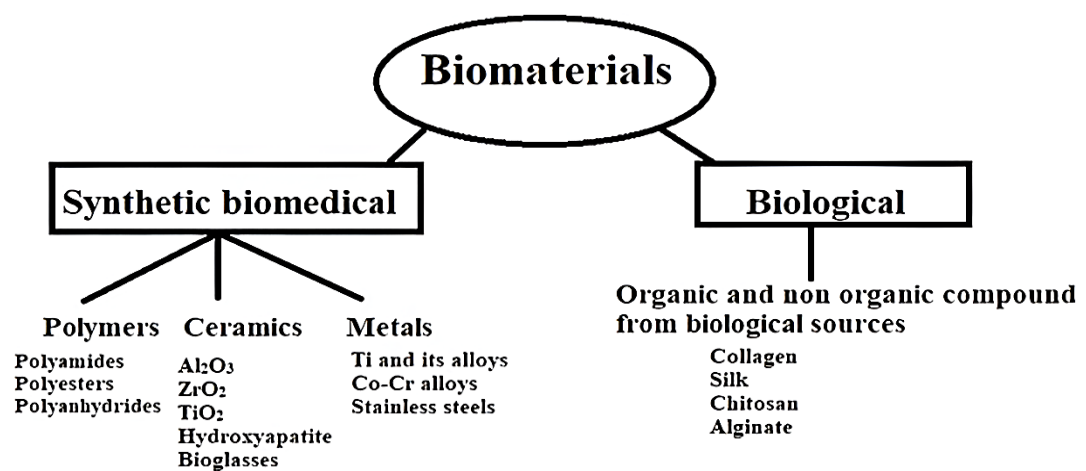


Figure 1.3 Classes of biomaterials (Henao et al., 2019)

These efforts ultimately lead to improved outcomes for patients. Absolutely, the realm of biomaterials is constantly evolving as researchers uncover novel materials, improve existing ones, and develop innovative methodologies.

This ongoing exploration and improvement carry several important implications. Further development of these alloys holds the potential to enhance patient outcomes significantly (Niinomi et al., 2012).

Selecting appropriate characteristics based on surface morphology and chemistry are also important. (Amani et al., 2019; Duraccio et al., 2015).

Table 1.1 Uses of biomaterial (Parida et al., 2012)

Problem	Examples
Replacing an infected or broken part	An artificial hip joint and a kidney dialysis machine.
Healing	Sutures, bone plates, and screws
Diagnostic assistance	Probes and catheters
Treatment of	Catheters, drains

The comparison of Mechanical properties of Ti- alloys are summarized in Table 1.2.

The biocompatibility is demonstrated by the least possible release of ions (Bauer et al., 2013; Mohammed et al., 2014).The wearing and corrosion with Ti-6Al-4V lead to the release of vanadium and aluminium into the body, thus causing hypersensitivity and various clinical complications (Eliaz, 2019).

**Table 1.2 An analysis of the mechanical behavior of a biomedical science titanium alloy
(Geetha et al., 2009)**

Materials	Elastic Modulus (GPa)	Tensile Strength (MPa)	Alloy type
Commercially pure Ti (Cp grade 1–4)	99	239–549	α
Ti–6Al–4V ELI wrought	110	860–965	$\alpha + \beta$
Ti–6Al–4V ELI Standard grade	112	895–930	$\alpha + \beta$
Ti–6Al–7Nb Wrought	109	899–1049	$\alpha + \beta$
Ti–5Al–2.5Fe	109	1019	$\alpha + \beta$
Ti–13Nb–13Zr Wrought	79–85	969–1029	Metastable β
Ti–12Mo–6Zr–2Fe	74–84	1059–1099	β
Ti–35Nb–7Zr–5Ta	56	589	β
Ti–29Nb–13Ta–4.6Zr –	64	959	β
Ti–35Nb–5Ta–7Zr–0.40	66	1010	β
Ti–15Mo–5Zr–3Al	82	–	β

Vanadium is categorized as a toxic substance, while aluminium is classified in the capsule (scar tissue) group (Okazaki, 2001; Velasco-Ortega et al., 2010). Ti, Zr, Nb and Ta exhibit good properties in terms of tissue response (Okazaki, 2001).

The purpose of introducing these alloys, which are based on titanium and vanadium, is to reduce their toxicity (Kopova et al., 2016; Niinomi, 2003). These alloys are classified as α -phase titanium.

The elastic modulus of commercially pure titanium and Ti-6Al-4V (about 104 and 109 GPa, respectively) belong to the α and $\alpha+\beta$ kinds, but are not as good as other quality. Studies have demonstrated that tantalum and niobium possess qualities that make them appropriate for β containing the β phase (Kolli & Devaraj, 2018; Prakash & Uddin, 2017).

Illustration depicting the various interactions between bone and the implant are, as illustrated in Figure 1.4 Diagram showing the varied interactions at different topographical scales between the implant surface and bone (Gittens et al., 2011).

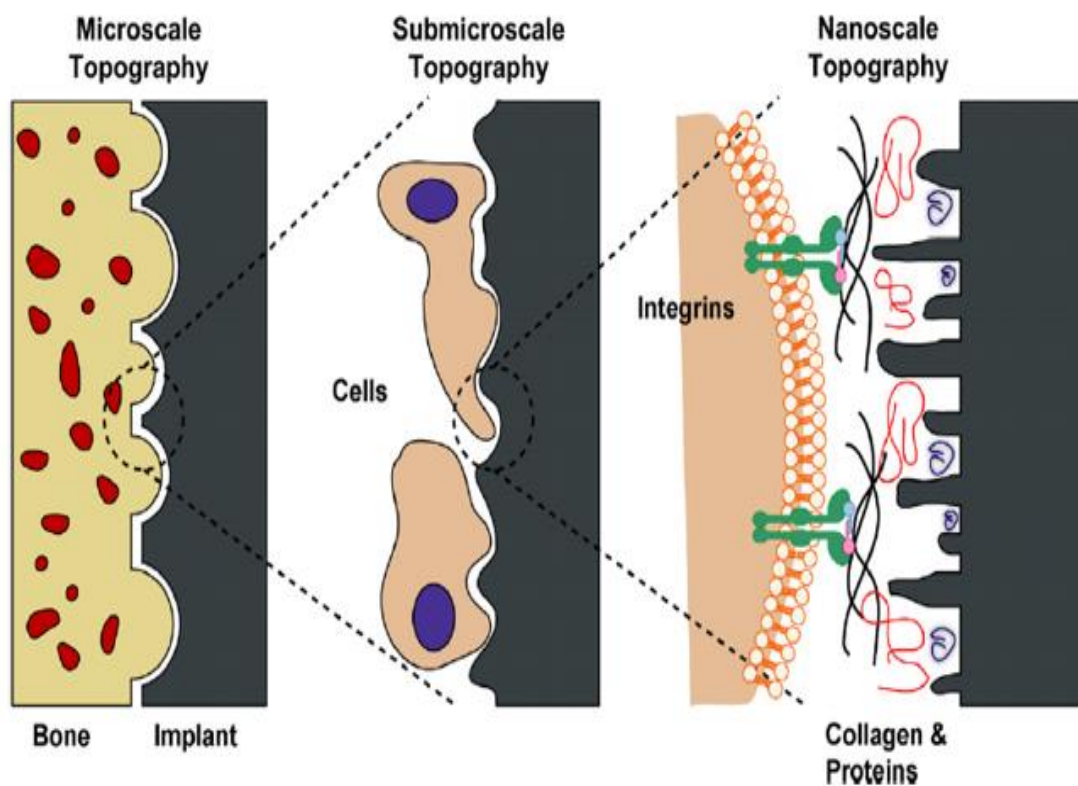


Figure 1.4 Diagram showing the varied interactions at different topographical scales between the implant surface and bone (Gittens et al., 2011)

Previous studies have shown that the characteristics of implants may directly affect how tissues respond by changing the adsorption of proteins and controlling the growth and specialization of cells (Kieswetter et al., 1996; Schwartz & Boyan, 1994). Surface characteristics such as roughness (Boyan et al., 2002; Martin et al., 1995), chemistry (Buser et al., 2004; Sul et al., 2005) and energy (Rupp et al., 2006; Zhao et al., 2005) have been reported for osseointegration (Att et al., 2007).

However, the surface of the implant materials remains in a "native" state, necessitating surface modification before the implantation process. Thus, to activate the surface, surface modification is necessary, resulting in an enhanced osseointegration rate. Strategies for surface modification of metallic implants have aimed to match bone characteristics in order to improve osseointegration (Kubo et al., 2009; Mendonça et al., 2010; Zhao et al., 2006).

These techniques enhance the surface qualities, characteristics, and enhance the biocompatibility of biomedical implants. Macro-roughness refers to visible imperfections that are easily noticeable without any special equipment. These imperfections can range in size from 100 μm to millimeters or even larger. Micro-scale surface roughness pertains to small irregularities that fall within the range of 1 to 100 μm in size. Submicron surface roughness refers to irregularities that fall within the range of 100 nm to 1 μm in size. Flaws smaller than 100 nm can be found in nano roughness. Different surface roughness levels can have distinct impacts on the behavior of implants and the attachment of cells. Macro-roughness is typically seen as advantageous as the larger imperfections can potentially enhance the implant's bonding with the surrounding bone tissue (Harcuba et al., 2012).

The beneficial effects typically outweighing any negative impacts (Gittens et al., 2014). This augmentation also leads to enhanced adsorption of extracellular matrix (ECM) molecules, which promote cell adhesion and thus improve cell adhesion itself (Walma & Yamada, 2020). The nanoscale roughness of the surface may have a significant impact on osteoblast differentiation and tissue regeneration (Ward & Webster, 2007). Numerous research groups have shown a keen interest in investigating the effects of nanoscale roughness (Curtis et al., 2004; Lipski et al., 2008; Riehle et al., 2003; Ward & Webster, 2007).

It increase the susceptibility against the pits formation (Gittens et al., 2011). Notable enhancements were seen in the connection between bone and the implant during trials conducted inside a living organism (Saruta et al., 2019) and improved clinical rates of wound healing (Gittens et al., 2014).

These interactions greatly influence the material's ability to effectively fulfil its intended function over the desired duration within the body. While conceptual in nature rather than immediately practical, this definition underscores three crucial points: (a)

the biomaterial must actively perform rather than merely exist, (b) the desired response from the body's tissues in the given application must be the most suitable among all potential reactions, and (c) biocompatibility must always be defined within the framework of the particular application (Williams, 2012).

The surface of biomaterials ought to be non-toxic and maintain stability over an extended period, without provoking any inflammatory or allergic reactions.

1.2.1.2 Surface Topography, Chemistry and Energy

Surface topography and chemistry are pivotal factors in determining the characteristics of materials, particularly in their interactions with biological systems (Elias et al., 2008). Surface topography encompasses features such as roughness and porosity. Sufficient surface topography ensures a strong biomechanical connection between the implant and surrounding tissues. Typically, an optimal configuration involves a blend of micron- to submicron-scale surface roughness along with an average porosity of 70% (Anselme et al., 2010; Hasegawa et al., 2020; Hong et al., 2020). It determines the surface energy of the implant, which influences its reactivity with surrounding tissues. Apart from surface topography and chemistry, surface energy also plays a significant role in influencing cell behavior and adhesion (Bruschi et al., 2015; Gentleman & Gentleman, 2014).

1.2.1.3 Corrosion and Wear Resistance

The durability of a material is typically assessed by its resistance to wear and abrasion. For implants, it is essential that the material is corrosion-resistant and has good wear resistance (Asri et al., 2017).

Total joint replacement (TJR) stands as a testament to modern medicine's success, offering significant pain relief and enhancing the quality of life for millions suffering from severe joint diseases. However, insufficient wear resistance can lead to the loosening of implants, causing excessive pain and a decline in functional abilities (Takakubo et al., 2013).

1.2.1.4 Fatigue characteristics of Metallic Biomaterial

Artificial hip joints, bone plates, and dental implants are examples of implants that resemble real bones. These are commonly employed in scenarios with significant cyclic stress, particularly when replacing weakened hard tissue. Hence, metallic materials known for their high strength, ductility, and toughness are prime choices.

Concerning the fracturing of structural biomaterials, fatigue fracture occasionally arises, posing a significant challenge among the various types of fractures. The durability of implants following implantation is evaluated by their ability to withstand fracture once the critical infection period has elapsed (Niinomi, 2007). Two primary pathological conditions impact bone tissue. The first involves implant loosening caused by inadequate integration between the implant and the surrounding bone, known as osseointegration and the second involves the regeneration of bone in patients who have suffered from severe bone injury or loss. These two conditions pose significant challenges in both dentistry and orthopaedics (Nobles et al., 2021; Uthayakumar et al., 2019). Figure 1.5 below highlights various events that occur when an oral implant is inserted into the jawbone.

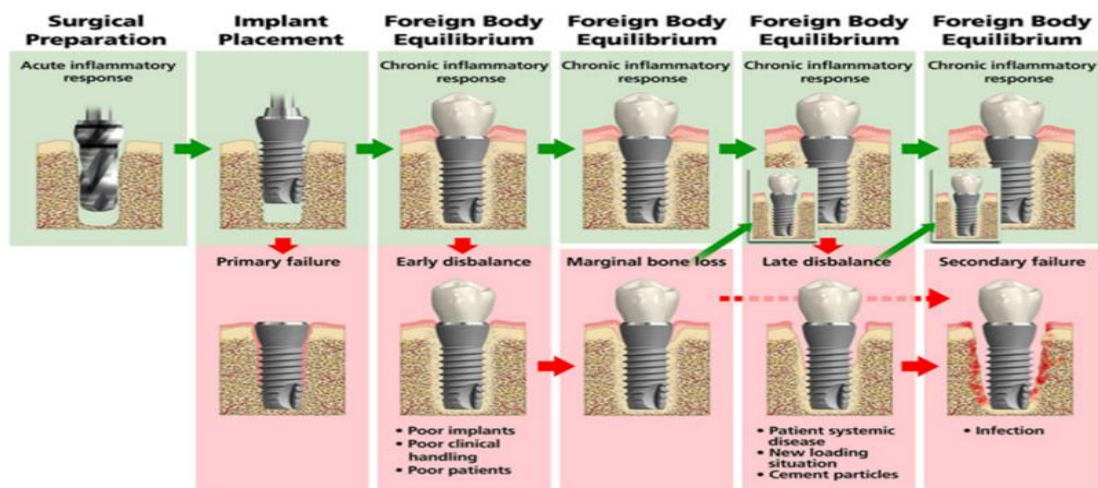


Figure 1.5 Foreign body equilibrium reaction (Albrektsson et al., 2014)

Titanium is a common choice for crafting load-bearing implants in clinical applications, due to its biologically inert properties and exceptional mechanical strength. In order to minimize the risk of implant loosening, surface modifications are designed to enhance

structure, roughness, and biological activity, promoting the crucial process of osteogenic differentiation, which is pivotal for the formation of osteoblasts (Stich et al., 2022). The Occurrences upon the placement of an oral implant in the jawbone involve an unavoidable foreign body reaction, so this phenomenon can be named as foreign body equilibrium reaction as in figure above. Another genuine concern with metallic implants is stress shielding. Figure 1.6 below illustrates the stress shielding mechanism. In vivo, the stress applied is diverted from the surrounding bone.

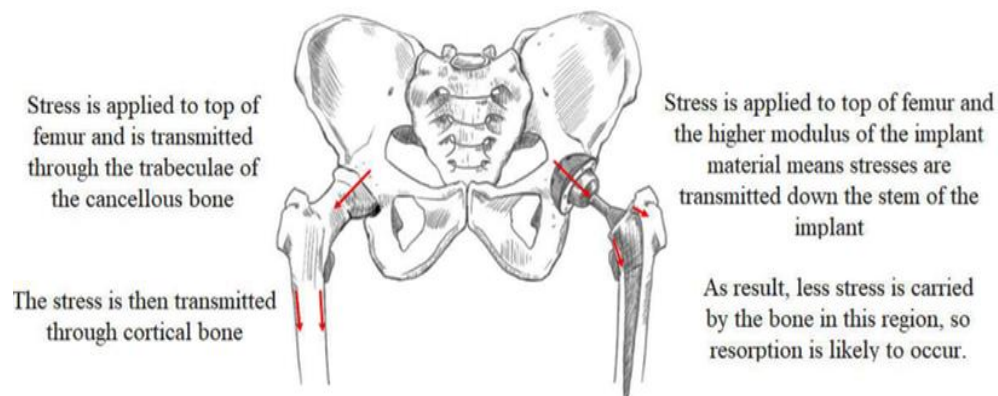


Figure 1.6 Mechanism of stress shielding (Mohammadi et al., 2021)

1.2.1.5 Principle of Bone Modeling and Remodeling

The cells that are involved in modeling and rebuilding bone are depicted in Figure 1.7 below.

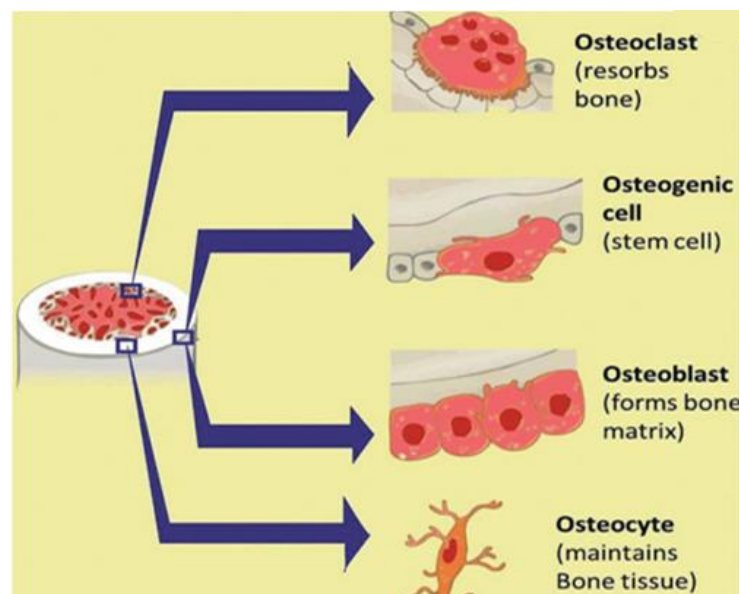


Figure 1.7 Cells for remodeling and modeling (Setiawati & Rahardjo, 2019)

Osteocytes, which are mature bone cells responsible for sustaining bone tissues; osteoblasts, which build the bone matrix; osteoclasts, which break down bones; and osteogenic cells, which are the key stem cells are involved (ElHawary et al., 2021). The balance between bone resorption and bone production, as well as the schematic, are shown in Figure 1.8 below.

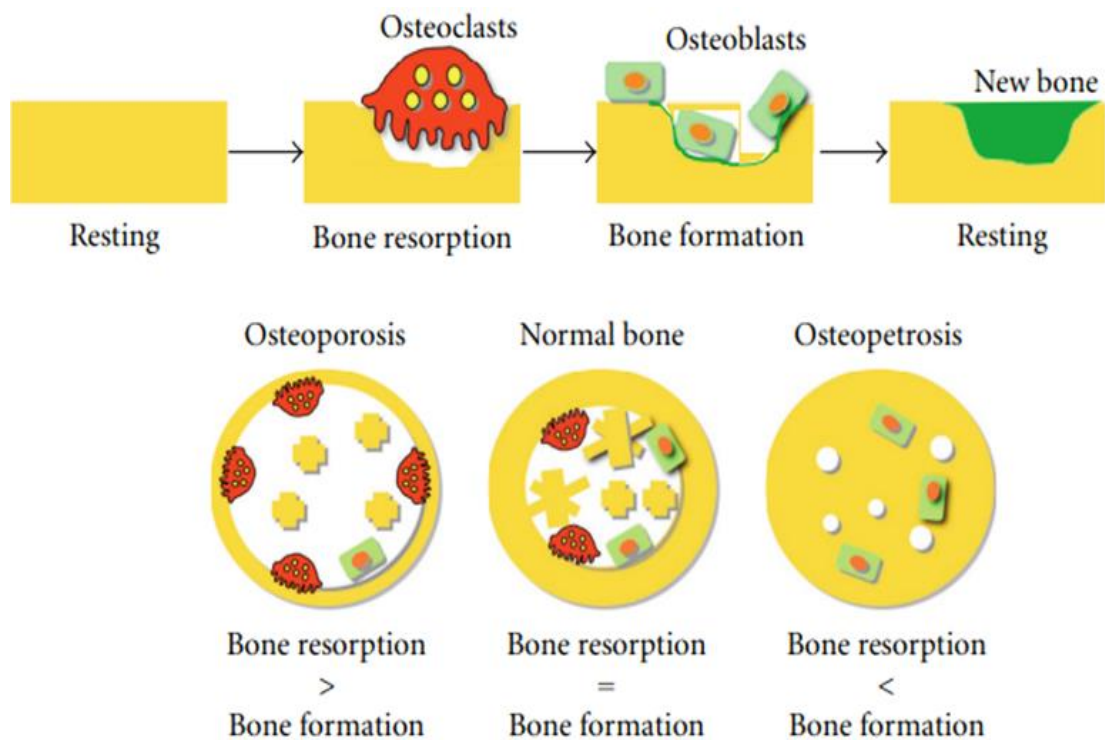


Figure 1.8 Schematic of the bone remodeling (Jimi et al., 2012)

1.3 Need of Surface Modification

The interplay between surface chemistry, topography, and surface energy is of utmost importance, especially during the early stages of the biological response, which ultimately impacts the osseointegration (Feller et al., 2015).

Establishing an appropriate porous and rough morphology constitutes the initial stage in the development of a bone implant surface (Stich et al., 2022). According to (Umeda et al., 2017), the blasting technique also enhanced the expression of key cell-matrix and cell-cell adhesion proteins and raised cellular alkaline phosphatase activity, suggesting enhanced osteogenic potential.

1.4 Surface Modification Techniques

As can be seen, the most common techniques for surface modification include chemical, mechanical, sol-gel, oxidation, carburization, ion implantation, and EDM. Of all these methods, electrical discharge machining (EDM) appears to be the most promising, since it can create a layer of nanopores with nanoscale pores. According to (Alipour et al., 2022; Prakash, Kansal, Pabla, & Puri, 2016a) this arrangement is known to offer extra benefits. Here are the reasons why EDM is regarded as a superior and durable surface modification technique compared to conventional methods:

- EDM has the ability to regulate the thickness of the coating layer in contrast to conventional techniques.
- Surface alloying/modification during EDM offers reduced expenses and increased flexibility.

1.5 Electrical Discharge Machining

Electrical discharge machining (EDM) is a method that has found extensive application in processing hard and tough materials, which pose challenges for conventional machining processes. The experimental setup for EDM is depicted below in Figure 1.9.

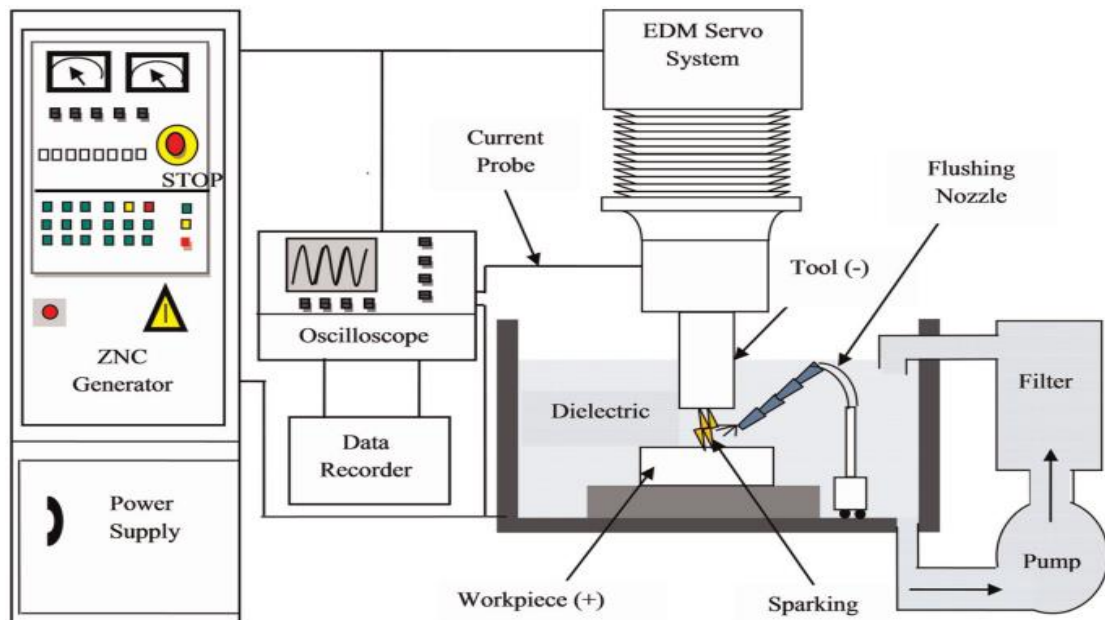


Figure 1.9 Typical EDM setup (Prakash, Kansal, Pabla, Puri, et al., 2016)

Its salient feature is that it may be used to machine electrically conductive objects using thermal energy, regardless of how hard they are.

The production of molds, dies, automotive, aerospace, and surgical components has found great value in this distinctive property. It is also utilized in the manufacturing of surgical components and for polishing parts in the automotive and aerospace industries (Ho & Newman, 2003).

During the EDM process, the insulation of the dielectric is essential for preventing electrode electrolysis. As shown in Figure 1.10, a high voltage greater than the small gap's dielectric breakdown strength causes a spark to be produced at the narrowest point between the electrodes.

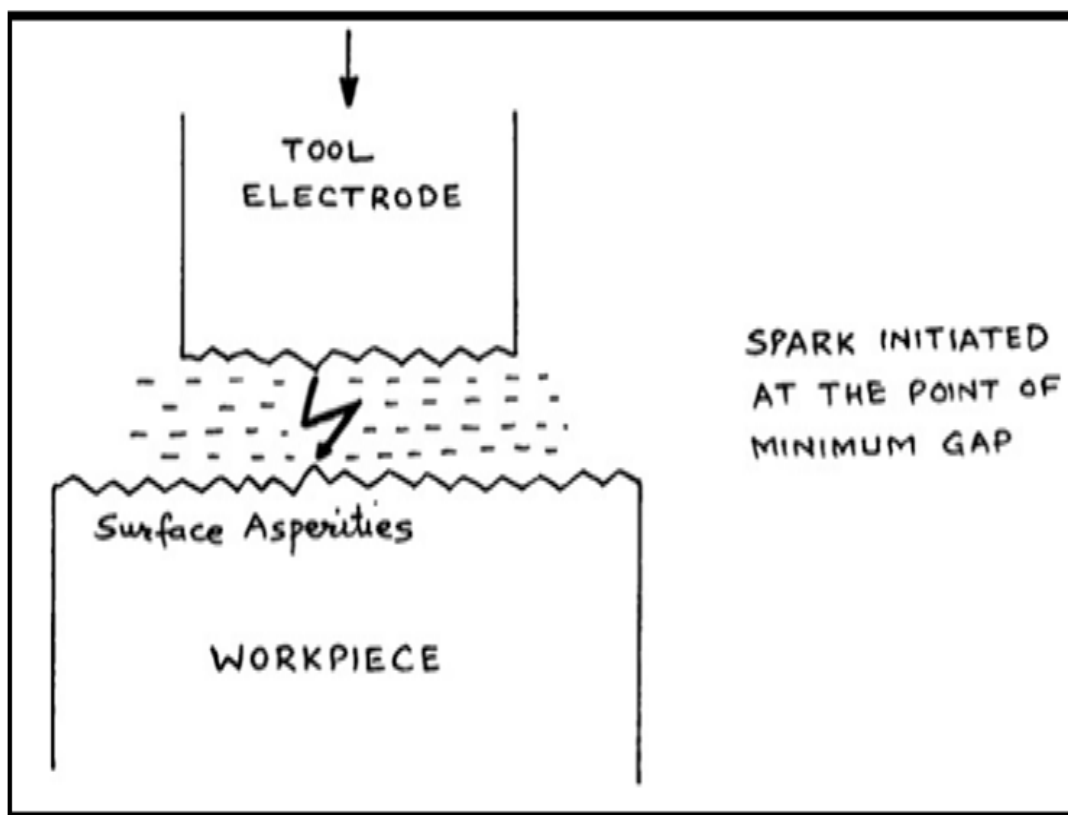


Figure 1.10 Spark initiation in EDM process (Kumar et al., 2009)

Here, there is metal erosion on both electrodes and refilling of capacitor after discharge is from DC source via resistor as in Figure 1.11.

Reducing the peak current and extending the duration of the spark would clearly lead to a decrease in electrode wear and an enhancement in machining efficiency. This was accomplished by using a regulated pulse generator. The typical waveform of this generator is illustrated in Figure 1.12.

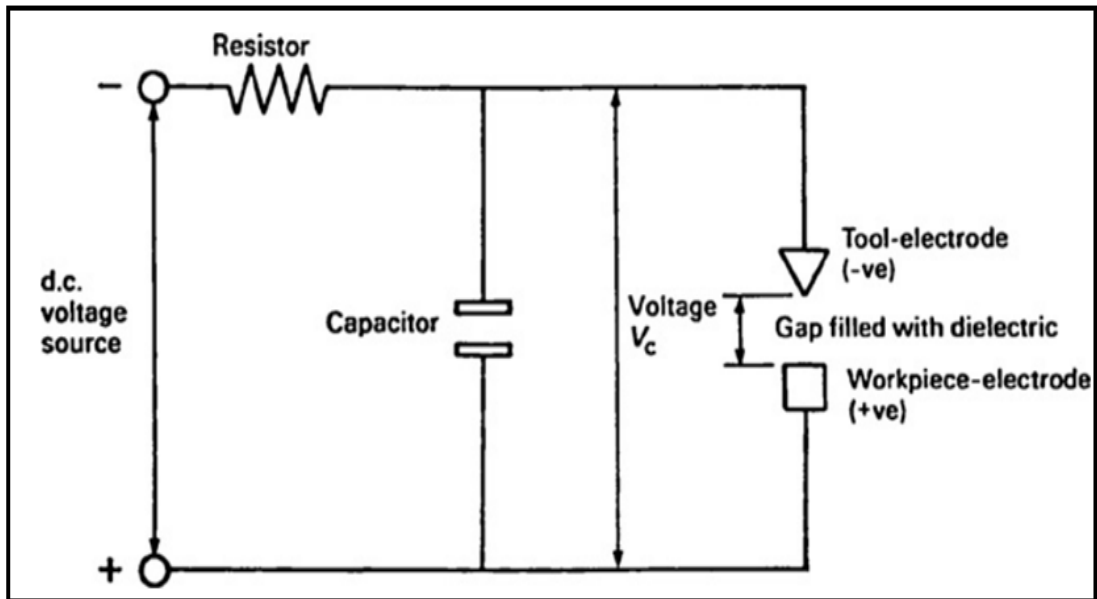


Figure 1.11 The lazarenko RC circuit (Kumar et al., 2009)

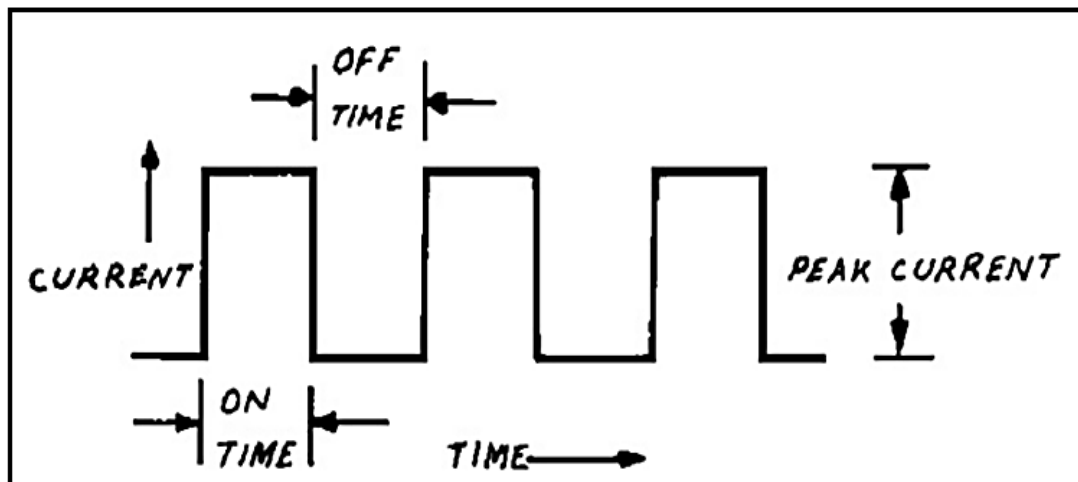


Figure 1.12 Pulse waveform of controlled pulse generator (Kumar et al., 2009)

As per Figure 1.12, as soon as the electrode is moved away from the workpiece, the potential normally increases to the open circuit voltage, which is approximately 100 V. A reduction in potential to about 35 V occurs when the dielectric starts to ionize and produces an electric current. Plotting voltage and current against time yields a more complete representation, as Figure 1.13 illustrates. Ionization time is represented by (a), discharge time by (b), deionization time by (c), and idle time by (d) in this instance.

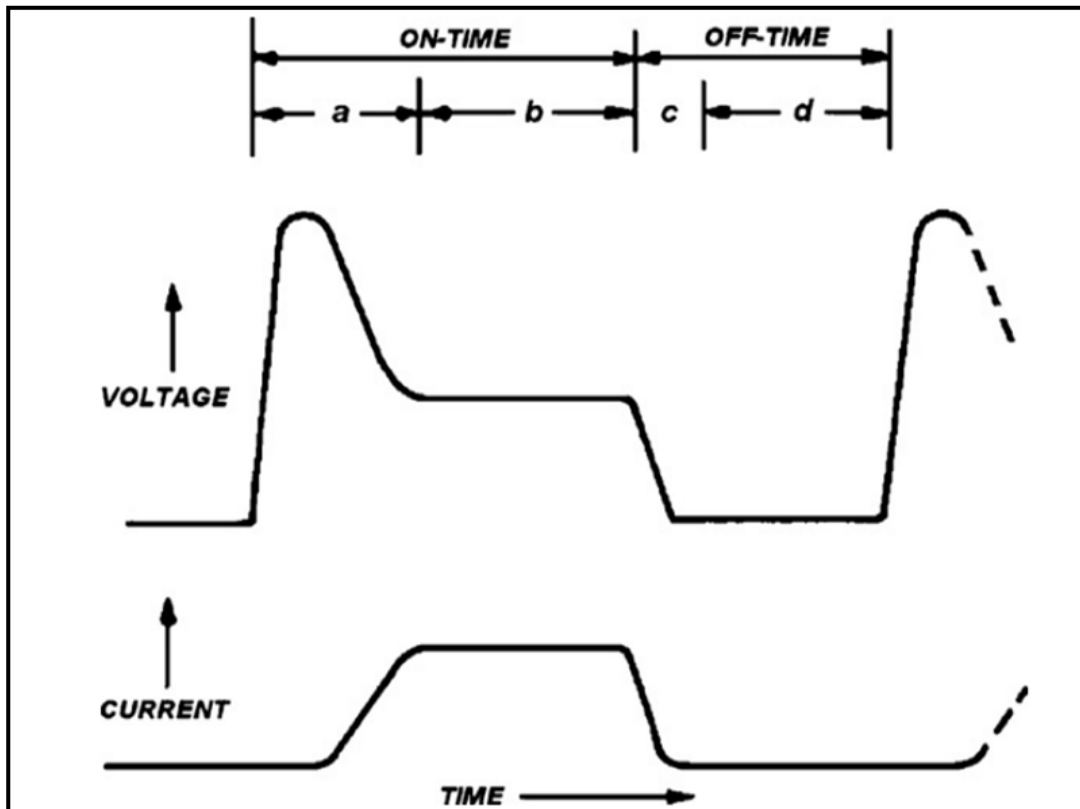


Figure 1.13 Profile for a single pulse of EDM (Kumar et al., 2009)

The ionization phase is when the majority of electrode wear occurs, although the RC circuit normally uses negative polarity for the tool electrode, positive polarity is preferred when with a pulse generator power source (McGeough, 1988).

During the process of EDM, a tremendous amount of heat is generated, reaching temperatures as high as 20,000°C. The intense heat due to striking of millions of electrons causes the melting at both the tool electrode and the workpiece (Kunieda et al., 2005).

The resulting debris, solidify on the workpiece surface, forming a recast layer consisting of oxides and various carbides (Bojorquez et al., 2002; Furutania et al., 2001; Luo, 1997).

This causes the machined surface to develop a variety of surface and subsurface defects, which restricts its further use. In 2007 (Mohd Abbas et al., 2007) reviewed with a critical eye the several developments that researchers have made to improve EDM performance. A few of these are as follows: (a) giving the electrode rotary or orbital

motion (Gohil & Puri, 2018); (b) helping with ultrasonic vibration (Abdullah et al., 2009); and (c) incorporating powder via dielectric fluid in EDM (Kansal et al., 2005).

1.5.1 Reverse polarity EDM

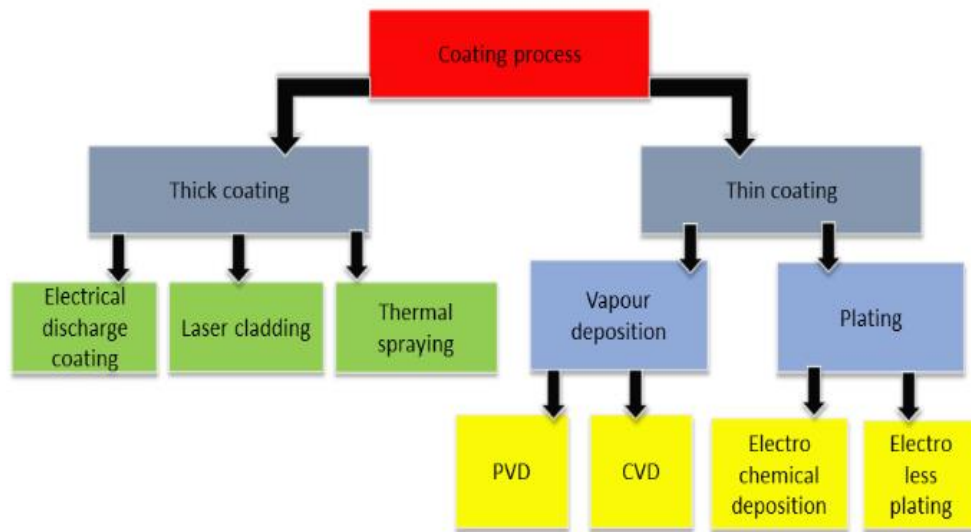
Reverse Polarity Electric Discharge Machining (EDM) is a specialized technique utilized in the manufacturing industry for machining conductive materials. EDM represents a non-traditional machining method wherein material is extracted from a workpiece. In reverse polarity EDM, the electrical charge polarity is reversed compared to the conventional EDM process.

1.5.2 Need for coatings on biomaterial surfaces

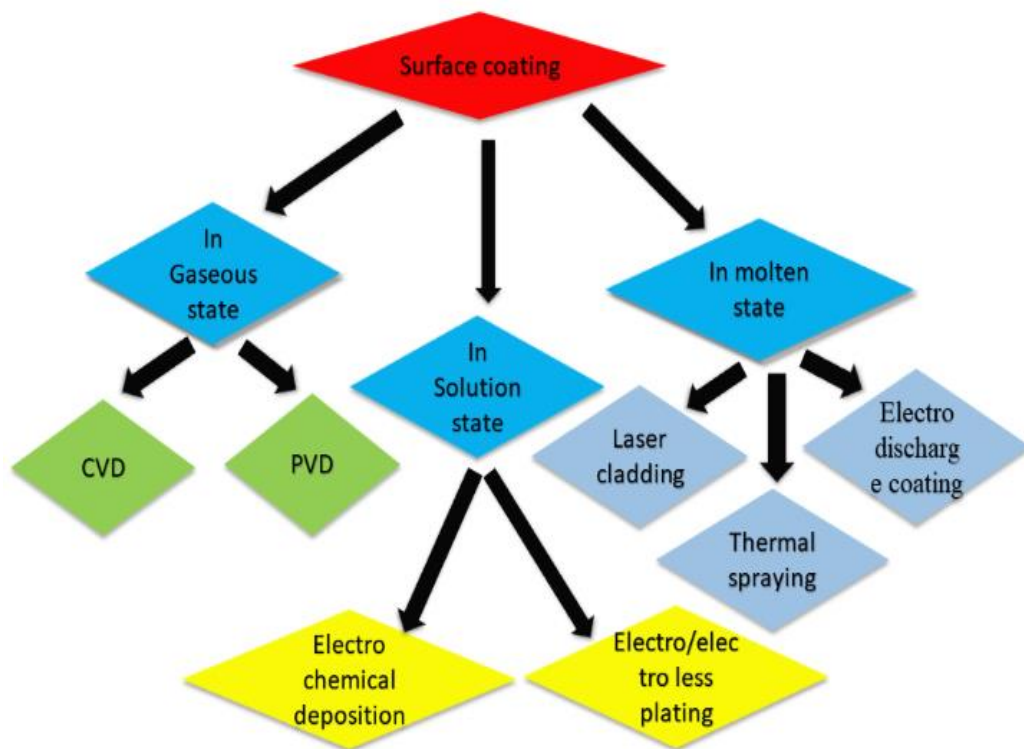
Conversely, in recent times, there has been a rapid and extensive adoption of surface modifications for biomaterials, aimed at augmenting biocompatibility and optimizing their environmental attributes. Additionally, the application of Electric Discharge Machining (EDM) has expanded to include orthopaedic implants due to the incorporation of new techniques where dielectric fluids are blended with specific additives such as metal powders, surfactants, and various gases. Apart from improving wear and corrosion, the incorporation of additives has significantly bolstered the mechanical properties and fatigue lifespan of orthopaedic implants (Mohammed et al., 2014; P. et al., 2022). Moreover, utilizing additive mixed processes enables development of a biocompatible and nanoporous coating layer over the surface of implants (Peng et al., 2010; Zou et al., 2016). In conclusion, Electrical Discharge Coating (EDC) show a significant role in modifying material surfaces and improving functionalization for biomedical applications (S.-L. Chen et al., 2014; P. et al., 2022; Sales et al., 2016; Wandra et al., 2022).

1.5.3 Methods of Coating, Advantages, and Limitations

Tribological characteristics can be significantly enhanced by utilizing one of the numerous coating methods that are currently available for surface modification. Figure 1.14 shows how these coating techniques can be divided into groups based on how the material was deposited during the procedure.



(a)



(b)

Figure 1.14 Coating process aa per: (a) state of material in the deposition process and (b) coating thickness (Tyagi et al., 2022).

1.5.3.1 Electrical Discharge Coating (EDC)

Electrical Discharge Coating (EDC) is widely employed as a cost-effective method for coating, utilizing electrical discharge processes to create coating layer onto the surface of a substrate. This technique enhances wear resistance and hardness properties by utilizing the negative polarity of Electric Discharge Machining (EDM) (Tyagi et al., 2022).

Within this method, the thickness of the coating can be substantially regulated without the requirement for specialized equipment (Tyagi et al., 2018). Moreover, EDC operates without the need for complex setups and arrangements.

This disparity arises because water emulsion has a higher viscosity than kerosene, making it difficult to remove the molten material, resulting in its accumulation over the sample surface (Tyagi et al., 2022). Figure 1.15 (b) displays a schematic illustration of the process, in the powder suspension technique, as shown below. Metallic carbide can be deposited directly, as in Figure 1.15 (a), or it can form as the process goes on, as in Figure 1.15 (b).

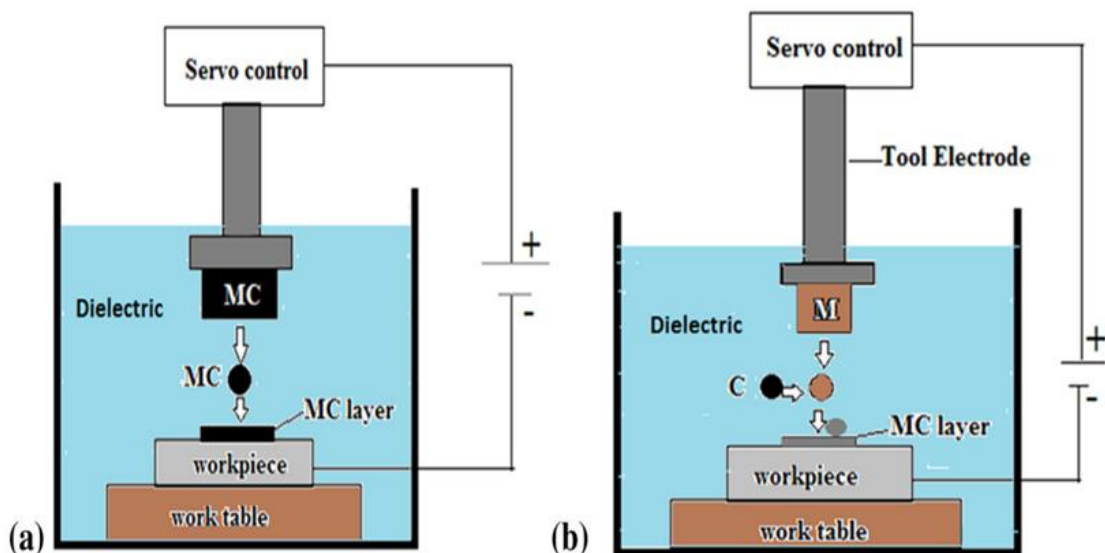


Figure 1.15 The EDC process's material transfer schematic representation, utilizing an electrode from a powder metallurgy tool (Tijo et al., 2017)

The function, polarity, and machined surfaces of the EDM and EDC procedures are different, as shown in Table 1.3.

Table 1.3 Differences between EDM and EDC (Liew et al., 2020)

	Electrical Discharge Machining (EDM)	Electrical Discharge Cladding (EDC)
Definition	A material removal process that shapes a workpiece by eliminating excess material to achieve the desired form.	A coating process that applies a layer of material to a workpiece for surface modification.
Electrode polarity	The electrode or tool can be connected to either the positive or negative terminal, depending on the parametric conditions, such as tool wear rate and material removal rate.	The electrode or tool is placed at the anode (positive terminal), while the workpiece is positioned at the cathode (negative terminal).
Weight of work piece after processing	decrease because the material has been removed.	Increase due to the deposited material (or decrease in some cases due to crater formation).
Function	To remove material from the surface of the workpiece.	To enhance the surface properties of the workpiece (reduces cracks).

1.5.4 EDC Vs. BB-EDC

Among the techniques explored previously, Electric-Discharge Coating (EDC) stands out as a widely adopted cost-effective coating method. It employs electrical discharge procedures that use the negative polarity of Electrical Discharge Machining (EDM) to apply coating material onto the substrate surface, improving wear resistance and hardness qualities (Tyagi et al., 2018, 2022).

1.5.4.1 Electrical Discharge Coating (EDC)

Electrical Discharge Coating (EDC) has emerged as a transformative technique for enhancing the surface properties of metallic biomaterials, particularly titanium alloys used in orthopedic implants. The process involves the use of electrical discharges to deposit bioactive materials onto the substrate, thereby improving properties such as biocompatibility, corrosion resistance, and mechanical strength. EDC-treated surfaces exhibit significant alterations in surface morphology and microstructure. Prakash et al. (2015) reported the formation of a nanoporous bioceramic layer on β -Ti implants using hydroxyapatite (HAp) powder mixed in the dielectric fluid. The resulting surface demonstrated enhanced roughness and porosity, which are conducive to cell adhesion and proliferation. Similarly (Ekmekci & Ekmekci, 2016) observed the formation of a uniform and adherent HAp layer on Ti6Al4V substrates, contributing to improved surface characteristics. The mechanical properties of EDC-treated surfaces have shown notable improvements. (Prakash et al., 2015) found that the incorporation of HAp in the dielectric fluid during EDC led to increased surface hardness and wear resistance. The enhanced mechanical properties are attributed to the formation of a hard ceramic layer and the refinement of the surface microstructure. EDC has been effective in enhancing the corrosion resistance of metallic implants.

(Prakash & Uddin, 2017) demonstrated that EDC-treated β -Ti implants exhibited superior corrosion resistance in simulated body fluid (SBF) environments compared to untreated samples. The improved corrosion resistance is linked to the formation of a stable and protective oxide layer on the implant surface. The biocompatibility and bioactivity of EDC-treated surfaces have been extensively studied. (Prakash et al., 2015) reported enhanced cell proliferation and differentiation on EDC-treated β -Ti implants, indicating improved bioactivity. The presence of bioactive phases such as HAp and TiO_2 on the surface contributes to better osseointegration and overall implant performance. Despite the advantages, EDC processes can introduce surface defects such as microcracks and residual stresses due to rapid cooling rates. These defects may compromise the mechanical integrity and long-term performance of the implants. Therefore, optimizing process parameters is crucial to minimize such drawbacks.

1.5.4.2 Ball Burnishing-Assisted Electrical Discharge Coating (BB-EDC)

Ball Burnishing-Assisted Electrical Discharge Coating (BB-EDC) integrates mechanical burnishing with EDC to further enhance surface properties. This hybrid approach aims to mitigate the limitations of conventional EDC by refining the surface finish and improving coating adhesion. The incorporation of ball burnishing in the EDC process leads to significant surface refinement. (Lin et al., 2001) demonstrated that combining EDC with ball burnishing resulted in smoother surfaces with reduced roughness. The mechanical action of burnishing helps in closing microcracks and eliminating surface irregularities. BB-EDC-treated surfaces exhibit superior mechanical properties compared to those treated with EDC alone. (Yan et al., 2002) reported increased surface hardness and wear resistance in aluminium alloys treated with BB-EDC. The synergistic effect of electrical discharge and mechanical deformation contributes to the enhanced mechanical performance. The mechanical burnishing action in BB-EDC enhances the adhesion strength of the deposited coatings. (Prakash et al., 2022) observed that BB-EDC-treated β -Ti implants exhibited better coating adherence, leading to improved durability and longevity of the implants. BB-EDC-treated surfaces have shown promising results in terms of biocompatibility and osseointegration. (Prakash et al., 2022) reported enhanced cell proliferation and differentiation on BB-EDC-treated β -Ti implants. The refined surface topography and improved coating integrity contribute to better biological responses. BB-EDC offers numerous advantages, optimizing process parameters such as burnishing force, speed, and duration is essential to achieve desired surface characteristics. Further research is needed to establish standardized protocols for BB-EDC in biomedical applications.

In this chapter, we examine how electrical discharge machining (EDM) might be used in biomedical settings to modify or treat surfaces. The chapter is divided into three main sections, each of which focuses on research on the surface modification or treatment of implants employing EDM in dental, orthopaedic, and shape memory alloys, among other biomedical domains. Also included in these sections are conclusions from literature reviews and the robustness of bone-implant interfaces. They explore the potential use of EDM in biomedical area, aiming to enhance biocompatibility, corrosion resistance, improve wear, and fatigue performance. Additionally, the sections discuss the impact of surface porosities on the strength of bone-implant interfaces.

2.1 Research in to the Surface Alteration of Metallic Implant Through EDM for Orthopaedic Implant

The EDM technique can be widely used to machine and modify, the intelligent and hard materials. A comprehensive study of the surface modification of metallic implants using Electrical Discharge Machining (EDM) for orthopaedic applications reveals a significant body of research aimed at enhancing the performance, longevity, and biocompatibility of orthopaedic implants.

Numerous studies have investigated the application of EDM for surface modification of metallic implants, particularly focusing on materials such as titanium and its alloys, stainless steel, and cobalt-chromium alloys. Titanium (Ti) and its alloys stand out in this regard and are heavily utilized in the biomedical sector. The increasing use of titanium alloys can be attributed to outstanding properties and compatibility with biological systems. These investigations have explored various EDM parameters including discharge current, pulse duration, electrode material, dielectric fluid, and machining strategies to tailor the surface properties of implants according to specific orthopaedic requirements. This section focuses on the several key benefits of EDM surface modification for orthopaedic implants, including the ability to create precise surface topographies, improve surface roughness, control surface chemistry, and induce

compressive residual stresses. These modifications can promote osseointegration, reduce wear and corrosion, enhance biocompatibility, and minimize the risk of implant failure or loosening.

2.1.1 The Impact of EDM for Enhancing the Osseointegration

The subsequent studies have been thoroughly examined to assess the effectiveness of EDM-derived surface modifications on osseointegration, across both in vitro and in vivo settings.

2.1.1.1 Study Conducted In Vitro

Increased hydrophilicity and surface roughness in the nanoporous layers are favorable for cell adhesion and growth. In this investigation, machined titanium, polished titanium, and titanium that underwent EDM treatment were designated as M-Ti, P-Ti, & EDM-Ti, respectively, to facilitate identification. Figure 2.1 displays SEM images of Ti, EDM-Ti-1, EDM-Ti-3, & EDM-Ti-5. The Ti surface is seen in Figure 2.1 (a), where machining tracks are visible.

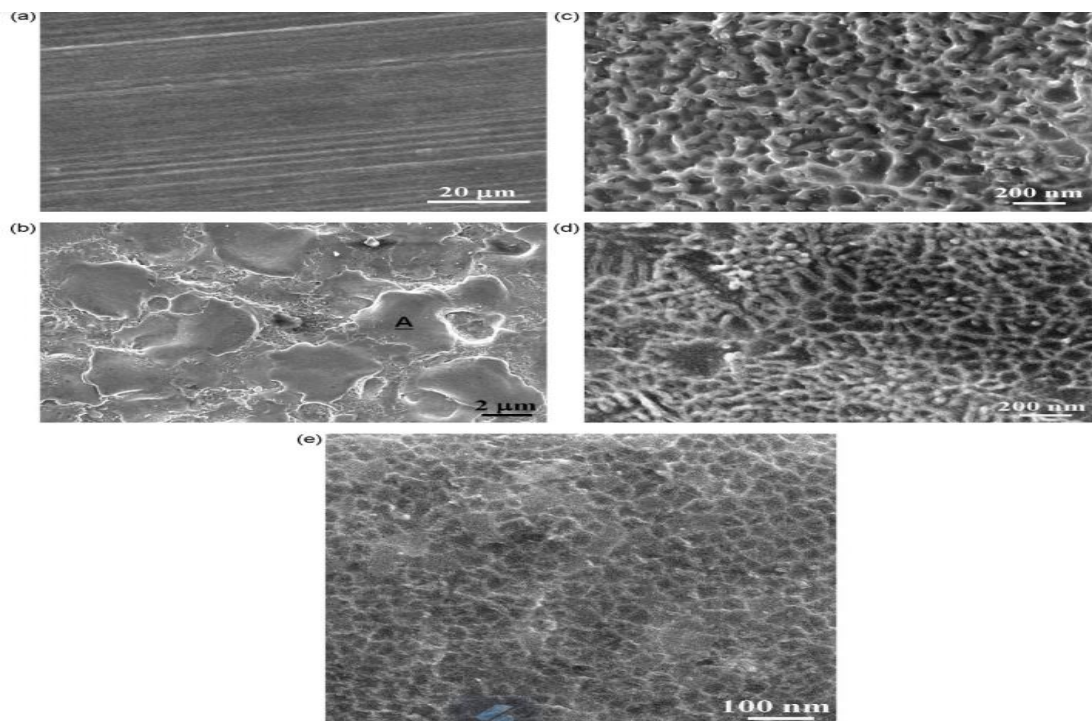


Figure 2.1 (a) Ti, (b) and (c) EDM-Ti-1, (d) EDM-Ti-3, and (e) EDM-Ti-5 shown in SEM images (Peng et al., 2010).

With molten metals that have solidified again, the EDM-Ti-1 surface takes on the form of a volcano in Figure 2.1 (b). Significant scaffold nano porosity on the EDM-Ti-1 surface (A in Figure 2.1 (b))) is shown in Figure 2.1 (c). After an extended period of immersion in an alkaline solution, this porosity structure became apparent. Additionally, as shown in Figure 2.1 (d & e), nanoporous forms were found on EDM-Ti-3 & EDM-Ti-5.

The findings have implications for the optimization of surface modification processes and the development of advanced implant materials with enhanced properties for biomedical applications.

(Chang-bin et al., 2011) conducted research which offers significant contributions to the field of surface modification techniques for titanium alloys in biomedical contexts. The results of osteoblast cell attachment as in Figure 2.2.

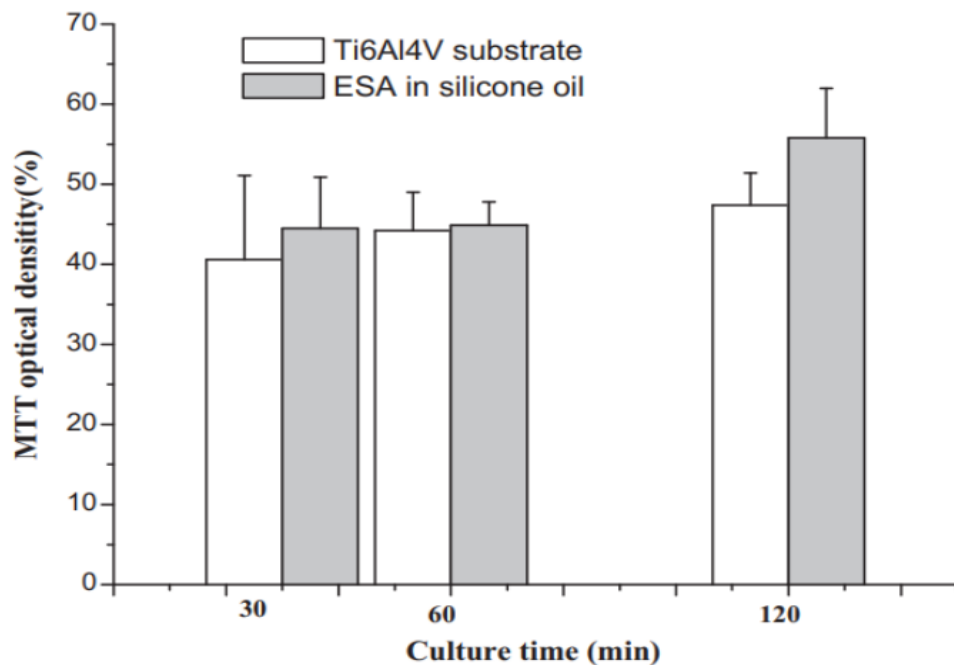


Figure 2.2 Cell attachment of osteoblasts on Ti6Al4V (Chang-bin et al., 2011)

(T.-S. Yang et al., 2013) evaluated biological characteristics of specimens treated with Electrical Discharge Machining (EDM), by examining the in vitro cellular responses, as depicted in There was a notable increase in cell proliferation following EDM functionalization, with a significant upregulation observed at higher current levels as in Figure 2.3

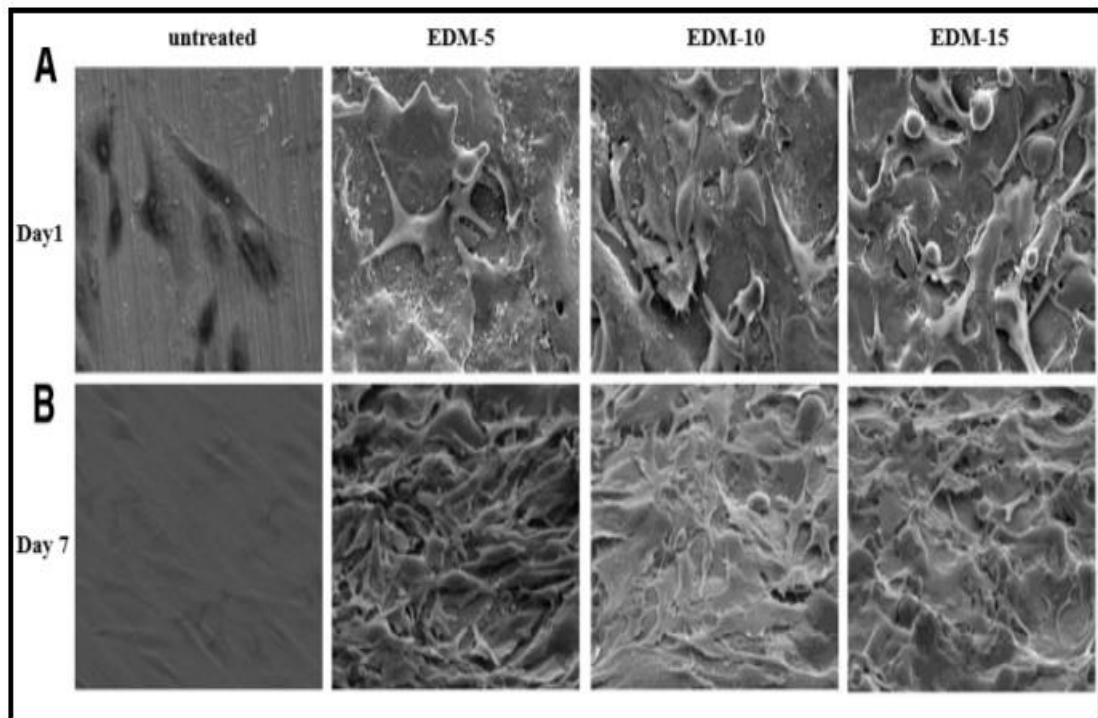


Figure 2.3 EDM-5, EDM-10, and EDM-15 untreated cells were used to test MG 63 cell adhesion and proliferation on (A) day 1 and (B) day 7 (T.-S. Yang et al., 2013)

The innovative surface modification method proposed by (Havlikova et al., 2014) holds promise for enhancing osteoblast proliferation and fatigue performance, thereby contributing to the development of more effective and reliable biomedical implants. The notably elevated levels of carbon and oxygen as in Figure 2.4.

(Prakash et al., 2015) explored nanoporous layer directly on β -Ti implants has been explored using powder mixed electrical discharge machining. This innovative approach involves the production of microscale pits, approximately 70 μm in size, and the creation of a biomimetic interconnected porous layer within these pits, ranging from 200–500 nm in size. The utilization of PMEDM not only introduces nanoporous layer but also induces alterations in surface chemistry conducive to enhanced bioactivity. The use of PMEDM to machine β -Ti alloy was studied (Prakash et al., 2017). The SEM images of β -Ti after PMEDM is shown in Figure 2.5. PMEDM produces a cleaner surface morphology with connected discharge craters that are smaller in size than traditional EDM.

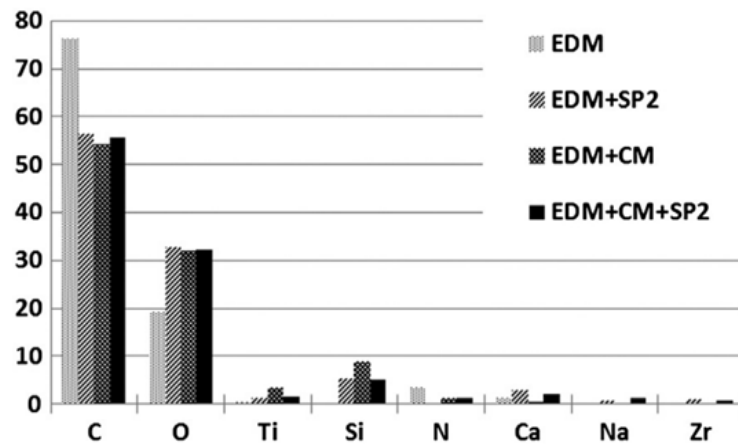


Figure 2.4 Chemical composition of the surface measured by X-ray photoelectron (Havlikova et al., 2014)

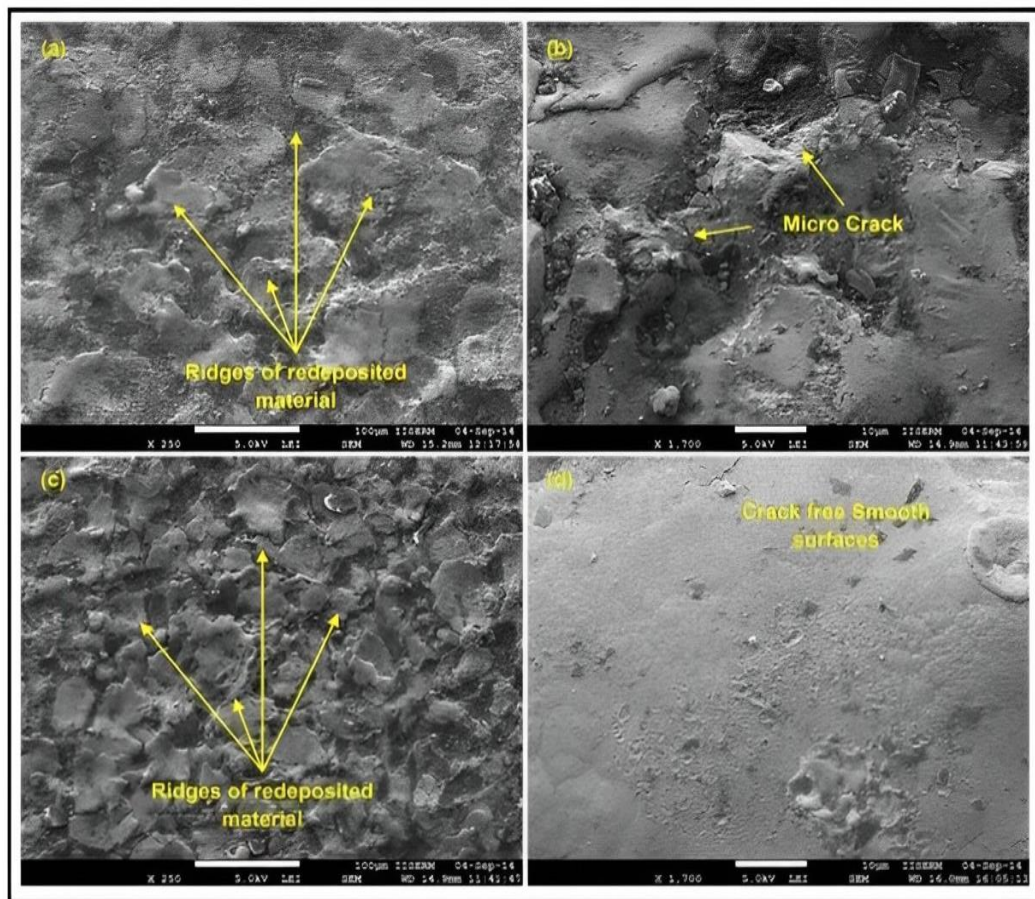


Figure 2.5 SEM images of the modified β -Ti alloy with pulse current of 10A and a duration of 10 μ s, with Si powder concentrations of (a-b): 2 g/l and (c-d): 4 g/l (Prakash et al., 2017)

(Jain & Parashar, 2021) examined the EDM process in relation to biomedical materials, considering their essential properties for biocompatibility, which dictate the appropriate machining procedures. Figure 2.6 illustrates cell proliferation.

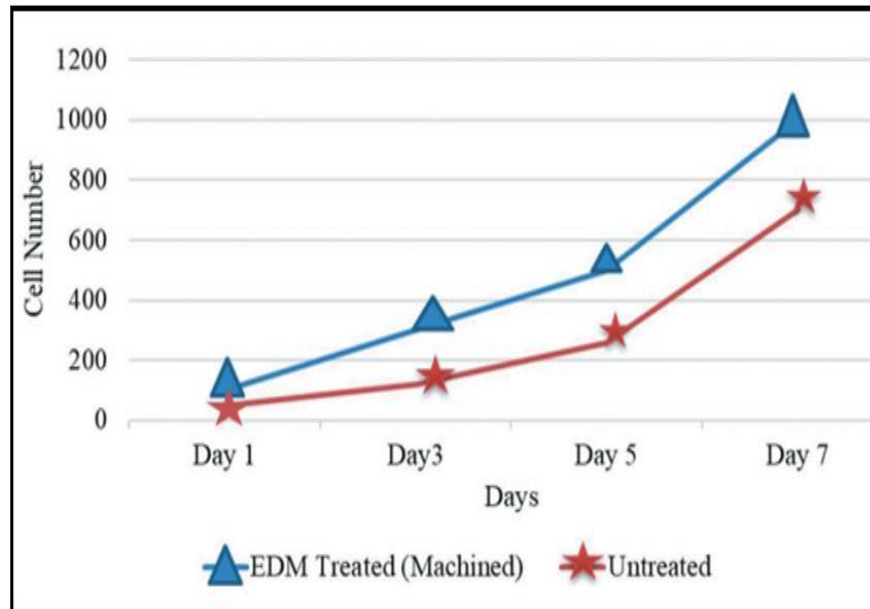


Figure 2.6 Cell growth assessment (Jain & Parashar, 2021)

The findings indicate that the machined surfaces treated in the aforementioned studies exhibit a higher cell count compared to non-machined surfaces.

2.1.1.2 Study Conducted In Vivo

In vivo study of four Japanese white rabbits were employed to assess bone formation around screw-shaped implants manufactured using conventional machining and EDM processes, as depicted in Figure 2.7.

White rabbits have long been recognized as a valuable preclinical animal model for evaluating osseointegration and peri-implant bone response due to their manageable size, well-characterized bone physiology, and relatively rapid skeletal turnover. In the present study, white rabbits (*Oryctolagus cuniculus*) were employed to assess bone formation around screw-shaped implants fabricated through conventional machining techniques. These implants were manufactured using standard lathe and threading operations, ensuring precise dimensional control, reproducibility, and surface finish that mimics typical clinical-grade implants. The rationale behind selecting conventionally machined screw-shaped implants lies in their clinical relevance, given

that such implants are extensively used in dental and orthopaedic applications for their mechanical interlocking capacity and surface topography that supports primary stability. The use of white rabbits offers several advantages, including sufficient tibial or femoral cortical bone volume for implantation, ease of surgical manipulation, and an immune response profile that closely mimics human bone healing at the early and intermediate stages. The experimental design involved bilateral implantation of machined implants into the tibial metaphysis under sterile surgical conditions, followed by euthanasia at predetermined time intervals—typically 2, 4, 8, and 12 weeks post-surgery—to evaluate time-dependent bone-implant interactions. Standard protocols for anesthesia, analgesia, and post-operative care were strictly followed to maintain animal welfare and comply with ethical guidelines.

The screw-shaped implants were typically composed of biocompatible titanium or titanium alloy (such as Ti-6Al-4V), known for their excellent corrosion resistance, low elastic modulus closer to that of bone, and inherent biocompatibility. The machined surface, although devoid of micro-roughness enhancements that are characteristic of advanced surface treatments like sandblasting, acid etching, or anodization, presents a moderately smooth interface that allows for the study of intrinsic bone response uninfluenced by complex surface modifications. This makes the model particularly valuable for baseline assessments or comparative studies where the goal is to evaluate how surface engineering strategies improve upon the conventional machined surface. Histological and histomorphometric analyses were the cornerstone methods employed to quantify bone formation around the implants. After retrieval, the tibiae containing the implants were fixed in neutral-buffered formalin, dehydrated, embedded in polymethylmethacrylate (PMMA), and sectioned longitudinally along the axis of the implant using precision saws and microtomes.

Stained sections (commonly with toluidine blue or Masson's trichrome) were then examined under light microscopy to visualize newly formed bone, the bone-implant interface, and cellular responses such as osteoblast activity and vascular infiltration. Quantitative parameters such as bone-to-implant contact (BIC) percentage, bone area fraction occupancy (BAFO), and bone volume/tissue volume (BV/TV) were assessed using image analysis software to evaluate the degree and pattern of osseointegration. The presence of woven bone at earlier time points and its gradual replacement by

lamellar bone at later stages served as indicators of natural remodeling processes and implant integration. Additionally, micro-computed tomography (micro-CT) scanning was conducted in some protocols to obtain three-dimensional reconstructions of the peri-implant region, allowing for non-destructive analysis of trabecular microarchitecture parameters such as trabecular thickness, number, separation, and connectivity density.

Mechanical push-out or pull-out tests were also employed in parallel to histological evaluations to correlate the degree of bone formation with the mechanical anchorage of the implant. These tests involved applying a uniaxial force to the implant until failure occurred, with the peak load and energy-to-failure serving as indicators of mechanical integration. The implants' primary stability was primarily governed by their geometry and the quality of the rabbit tibial bone at the time of placement, while secondary stability resulted from the biological bonding through bone apposition over time. The results of such studies generally reveal a time-dependent increase in BIC and mechanical strength, with a plateau often reached around the 8–12 week mark, indicating the maturation of the bone-implant interface.

In addition to the standard analyses, immunohistochemical staining for markers such as osteocalcin, collagen type I, and alkaline phosphatase was performed in some studies to elucidate the molecular underpinnings of osteogenic activity around the implant surface.

These markers help identify the differentiation status of osteoblasts, matrix deposition, and mineralization dynamics, thereby offering a deeper insight into the regenerative capacity of the host bone in response to the machined implant surface. Furthermore, scanning electron microscopy (SEM) and energy-dispersive X-ray spectroscopy (EDS) analyses were used to characterize the surface morphology and elemental composition of the implants before and after implantation to determine any alterations or deposition of bone-related minerals, such as calcium and phosphorus, on the implant surface.

The use of white rabbits for such studies not only provides vital data regarding the osseointegration of conventionally machined implants but also serves as a benchmark for comparing modified implants with enhanced surface topographies, coatings, or biofunctionalization strategies. The healing process in rabbits is rapid enough to observe meaningful bone remodeling within a relatively short experimental

period, yet consistent and predictable, thus ensuring the validity and reproducibility of results. The combination of histological, mechanical, and molecular assessments in this model provides a comprehensive understanding of how the host bone responds to clinically relevant implant surfaces. This approach establishes a solid foundation for future innovations in implant design and surface modification aimed at enhancing bone healing and long-term implant success. These markers help identify the differentiation status of osteoblasts, matrix deposition, and mineralization dynamics, thereby offering a deeper insight into the regenerative capacity of the host bone in response to the machined implant surface. The implants' primary stability was primarily governed by their geometry and the quality of the rabbit tibial bone at the time of placement, while secondary stability resulted from the biological bonding through bone apposition over time.

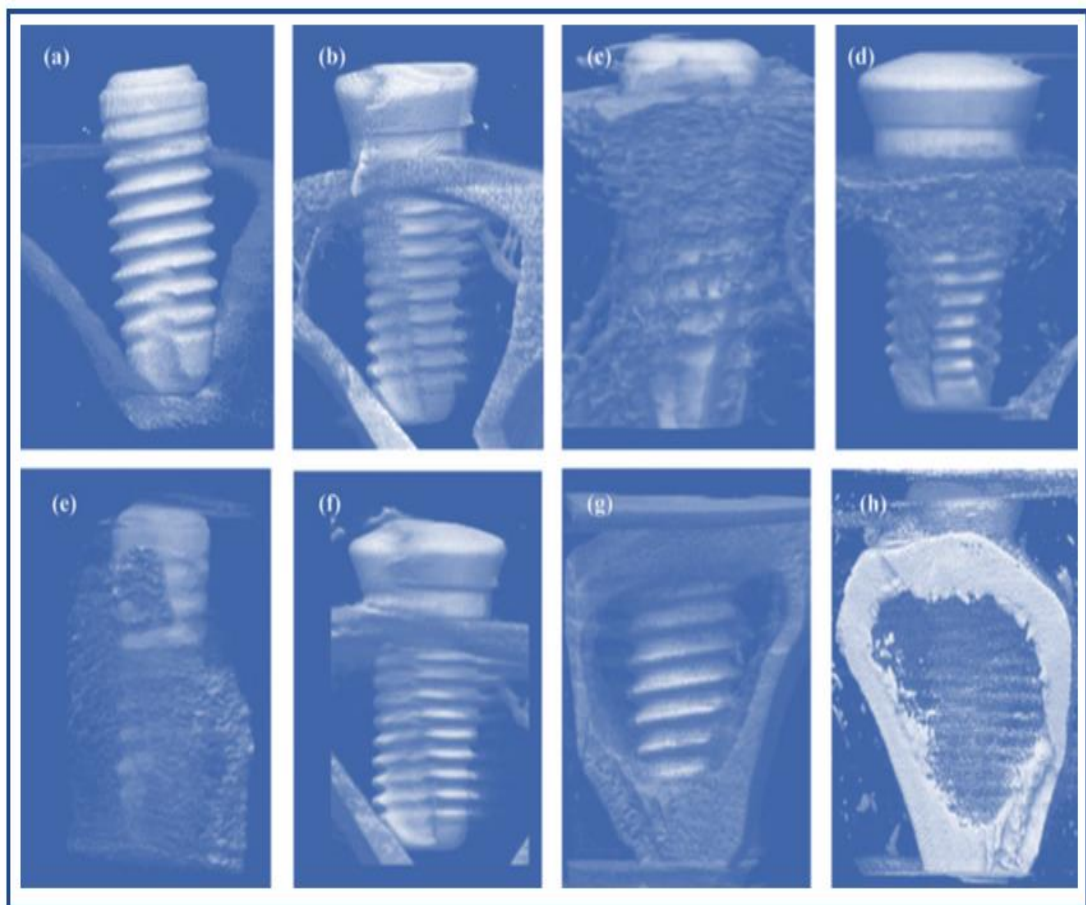


Figure 2.7 Post-implantation Micro-CT images of EDM-treated implant covered by bone tissue (Singh et al., 2022)

The figure highlights a greater degree of bone formation around surfaces treated with EDM compared to those processed using conventional machining methods. Additionally, in another investigation involving live animals, as illustrated in Figure 2.8.



Figure 2.8 In vivo study of implantation on porous Ti alloy implant (Singh et al., 2022)

(Bahl et al., 2018) explored TNS-2 and TNS-4 cast pancakes, comprising Ti-32Nb-2Sn and Ti-32Nb-4Sn compositions, which were produced using non-consumable vacuum arc melting, employing elemental. The fluorescent micrographs in Figure 2.9 show the hMSCs at one, three, and five days. One day later, all specimens show the same widely distributed appearance of the cells. Thus, as time goes on, the number of cells grows.

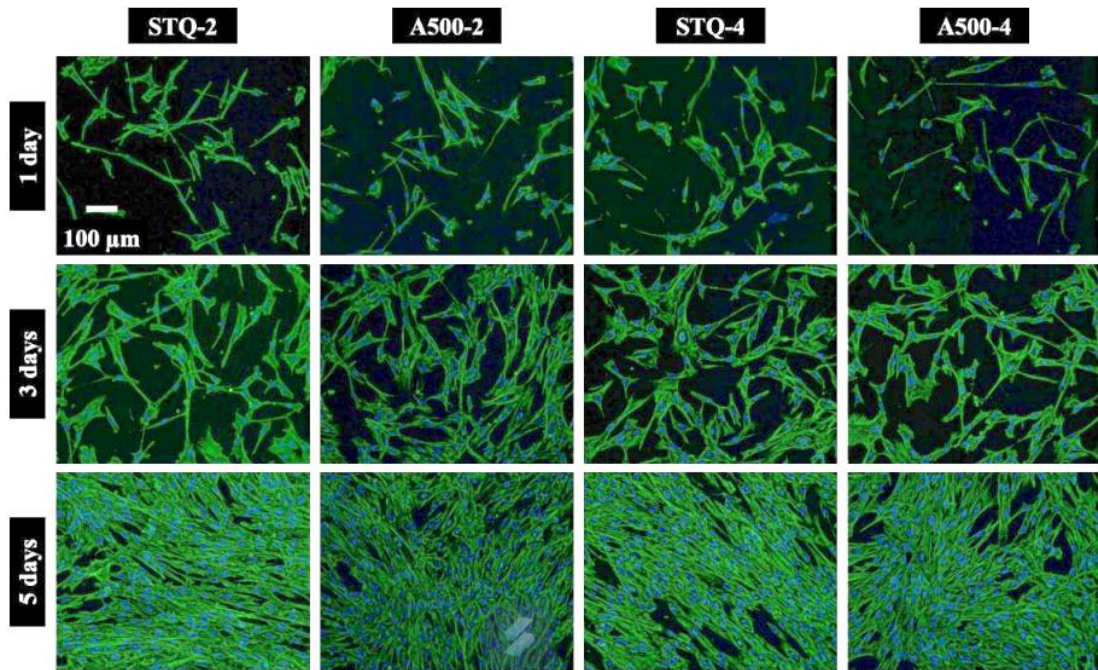


Figure 2.9 Human mesenchymal stem cells were used to create fluorescence micrographs at one, three, and five days following cell seeding. Actin filaments are displayed in green, and nuclei are displayed in blue (Bahl et al., 2018).

2.1.2 Utilization of Electrical Discharge Machining EDM to Improve Surface Integrity

Several optimal solutions were identified through optimization of all responses. (Prakash et al., 2018) involved creating a biomimetic HA-containing layer with a low surface roughness (SR) of 0.7 μm , low rough layer thickness (RLT) of 11.80 μm , and high microhardness (MH) of 246 HV. These were achieved with a hydroxyapatite concentration (CHA) of 5.20 g/l, a pulse current (I_p) of 3.40 A, a pulse-on time (T_{on}) of 40.30 μs , and a pulse-off time (T_{off}) of 109.20 μs . Furthermore, the characterized specimens exhibited the presence of a hydroxyapatite (HA) layer with interconnected pores ranging from 5 to 10 μm in size, as observed morphologically.

The surface properties of Electrical Discharge Machining (EDM) on Ti6Al4V ELI were investigated by (Umar Farooq et al., 2020) and they have used Si powder mixed in dielectric. The study examined how surface topography at the micro and nanoscale was affected by powder concentration (5 g/L, 10 g/L, and 20 g/L), pulse current, and pulse

length. A biocompatible surface that is nanostructured (50–200 nm) has also been designed to improve cell adhesion and proliferation. The use of Energy Dispersive Spectroscopy allowed for the confirmation of increased adhesion and proliferation in human osteoblasts (EDS), highlighting a highly favourable carbon-enriched surface. In their discussion of the impact of nano and micro hydroxyapatite powder additives on the surface integrity during the electrical discharge of Ti6Al4V alloys.

(Ekmekci & Efe, 2022) came to the conclusion that the micro additive dielectric liquid was used to achieve the highest rate of HA deposition to surface at a pulse current of 22A and a pulse duration of 100 μ s. The authors also came to the conclusion that the material mentioned above had a hardness rating of 1800 HV.

2.1.2.1 Regulated Surface Roughness and Hardness

Hydrocarbon oil that has been used as the dielectric or the carbon based increases the propensity being formation of the carbide increases. There are a lot of other parameters that has been chosen for increase in the machining parameters as well.

(Manjaiah et al., 2014) worked on the wire electric discharge machining properties of titanium nickel and the authors found out that the combination of the 1 μ s pulse on time, 3.8 μ s pulse off time and 40 V servo voltage were the main parameters. The authors also concluded that the development of cracks on the surface were due to the large pulse timings.

(Thakur et al., 2021) studied and optimized the surface roughness parameters using EDM by machining titanium (Ti-6246) alloys & concluded optimized parameters using the Box Behnken Design of RSM techniques. The authors also concluded that for the best optimization the surface roughness value obtained were 0.99 μ m respectively.

(Xu et al., 2022) studied the surface integrity of the titanium alloy machined with the EDM based milling systems and the authors concluded that the improvement in the roughness as compared to the conventional milling systems and the also found out an effective method of the machining of the titanium alloys.

(Baroi et al., 2022) studied the various parameters in electric discharge machining of the titanium alloys in the distilled water and the authors concluded that copper as the tool electrode with distilled water proved to be successful replacing the hazardous hydrocarbon oil. The authors also concluded that the MRR varies from 2 to 10 mg/min,

TWR varies from 0.2 to 1.3 mg/min and SR ranging from 4.1 to 9.6 μm . This study is very much significant in providing the insights to these processing parameters and output responses.

2.1.2.2 Surface Topography, Chemistry and Energy

(Lin et al., 2001) studied the improvement of Al-Zn-Mg alloy, using a combination of EDM and ball burnishing (BBM). The author concluded the improved hardened modified layer after processing the same material, through combined process of EDM with BBM. Reduction in micropores and cracks using combined Process of EDM with BBM under positive polarity EDM (Lin et al., 2001) is shown Figure 2.10

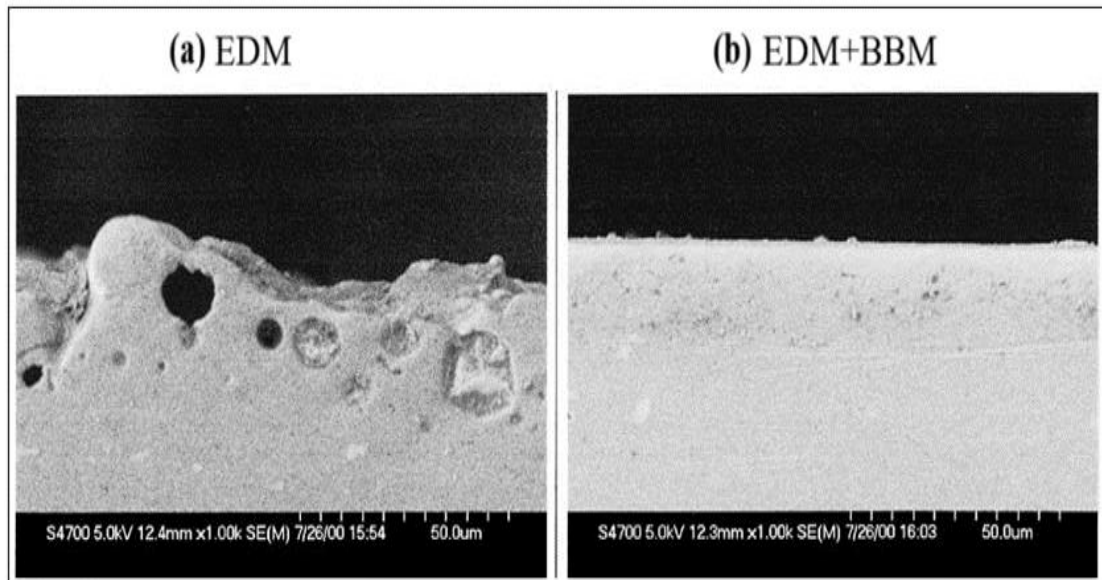


Figure 2.10 Reduction in micropores and cracks using combined Process of EDM with BBM under positive polarity EDM (Lin et al., 2001)

(Kumar et al., 2009) reviewed the paper on the surface modification by EDM and concluded the significance of intrinsic nature of the process. The selection of the electrode is very much significant in the EDM and the pyrolysis of the dielectric influence the surface composition and various other property parameters. The composite electrodes or the metallic powder are the two such material that has been used for the material transfer during the EDM.

According to (Devgan & Sidhu, 2020), the surface modification of β -titanium using a mixed EDM treatment of powdered μ -hydroxyapatite (μHAP) and multiwalled carbon

nanotubes (MWCNTs) produced a bio-favorable substrate. The authors concluded that these new development techniques were successful in achieving this goal. Furthermore, the authors determined that the graphite tool's polarity value of 15 A represents the best suitable process parameters for the bioactive surface growth.

2.1.3 Utilization of EDM to Enhance Mechanical Properties

According to (Prakash, Kansal, Pabla, & Puri, 2016b), significant improvement was observed by the author, with improved fatigue limit of 280 MPa, for the application of the same biomaterial in orthopedics as in Figure 2.11.

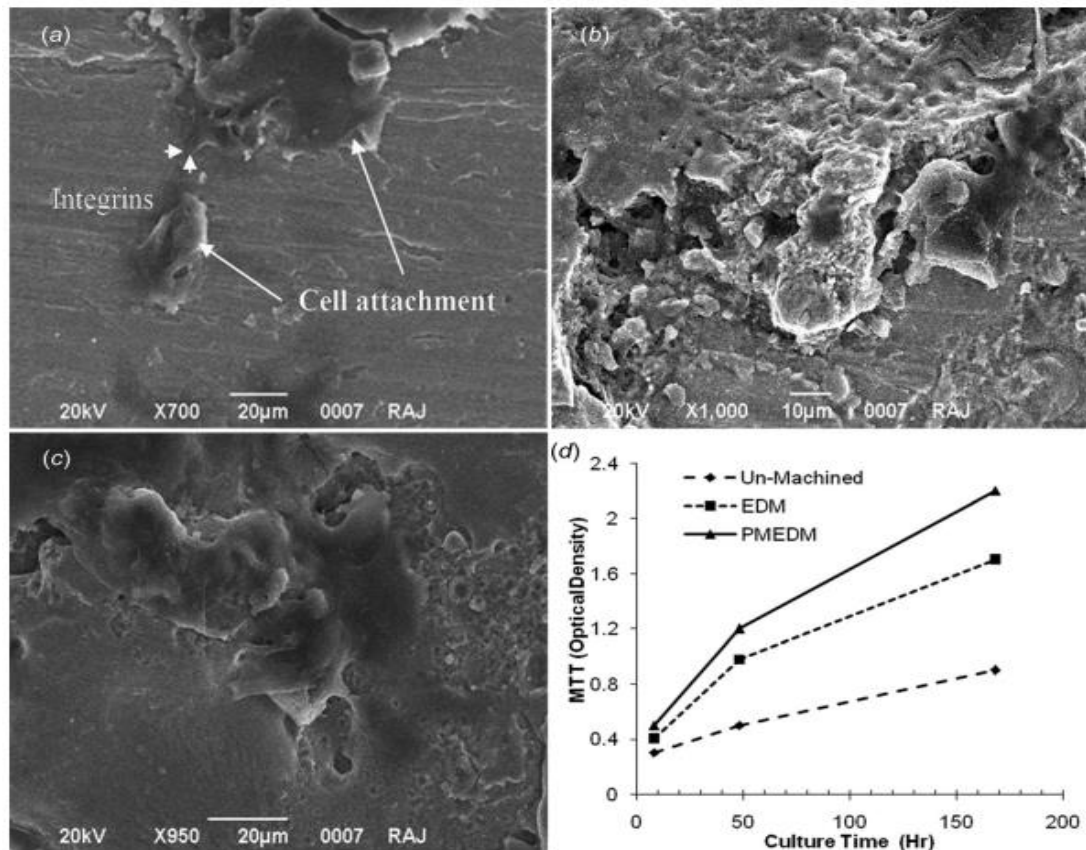


Figure 2.11 Behavior of β -Ti after processing it through MG-63 bone cells (a) polished, (b) EDM surface, and (c) PMEDM surface, and (d) culturing cells for 1,3, and 7 days using the same MG-63 cells (Prakash, Kansal, Pabla, & Puri, 2016b).

(Sharma et al., 2020) discussed the effect of the hBN (hexagonal boron nitride) powder mixed on titanium alloy through the micro-electric discharge machining and concluded

that there was reduction of wear rate of the coated surface to one forth to that of the base parent metal.

(Das et al., 2021) discussed the textural defect of the processed surface on Gap active electrical discharge machining and the authors concluded that the system consists of gap sensible and automatic adjustment of the electrode withdrawal system. The authors also discussed the employment of the auto sensible and synchronized with the provision of the flushing systems. The authors concluded that the adaptation of the secondary dielectric enriches the carbon containment and formation of the surface which is being harder than the conventional EDM systems.

(Munir et al., 2022) discussed the effect of equal channel angular pressed Ti-28Nb-35.4Zr alloys and all the mechanical, corrosion, nanotribological and biocompatible properties were investigated, and the authors concluded that the properties were enhanced and the process being suitable for the load bearing implant applications.

2.1.3.1 Fatigue performance and corrosion resistance

The potentiodynamic polarization graphs of untreated titanium, anodized heat-treated titanium with 1 hour gap, and titanium immersed in Hank's solution for seven days are shown in Figure 2.12

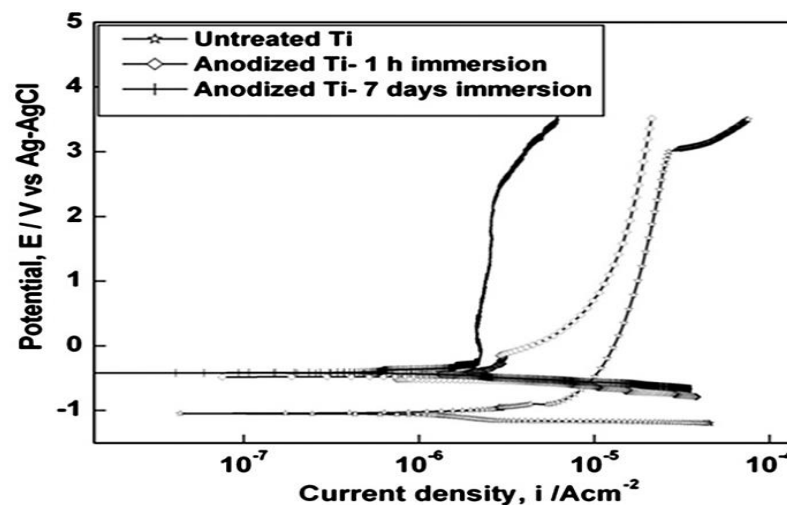


Figure 2.12 Untreated Ti, anodized Ti after one hour, and seven days submerged in Hank's solution polarization curves (Indira et al., 2013)

Table 2.1 presents an overview of the outcomes. After seven days in Hank's solution, anodized titanium exhibits enhanced potential due to surface hydroxyapatite.

Table 2.1 Hank's solution characteristics of untreated Ti, anodized Ti after 1 hour, and seven days (Indira et al., 2013)

Specimen	$-E_{\text{Corr}}$ (Volts)	I_{Corr} (A cm ⁻²)	R_p , (k Ω cm ²)
Untreated Ti	1.04	1.29×10^{-6}	20.05
Anodized Ti—1 h immersion	0.48	1.37×10^{-6}	20.40
Anodized Ti—seven day immersion	0.42	8.98×10^{-7}	67.93

(Lee et al., 2013) investigates the formation and properties of nanoporous layers on Ti–6Al–4V alloys and their influence on osteoblast-like cell response. The findings reveal enhanced osteoblast-like cell response, including increased cell adhesion & proliferation. A notable adherence of MG-63 cells was observed on the untreated surfaces, as depicted in part A of Figure 2.13, in contrast to its part B, which reveal that the cells exhibited widespread distribution and fully covered the surfaces of the EDM specimen.

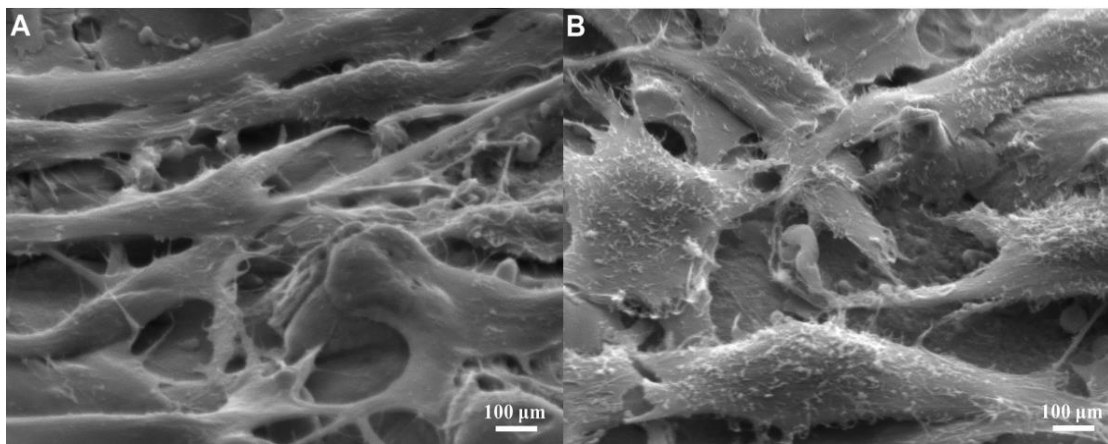


Figure 2.13 Adherence and growth of MG-63 cells on the (A) untreated specimen and (B) EDM specimen (Lee et al., 2013)

(Prakash, Kansal, Pabla, & Puri, 2016b) Studied different specimens of β -Ti alloy using various modifying methods- PMEDM, EDM, and polishing. Author concluded the improved high-cycle fatigue performance at a frequency of 50 Hz and stress ratio 0.1, with its variation due to surface roughness across different materials. The depicted values of surface roughness across different specimens are shown in Figure 2.14 and

similarly the S-N curve in Figure 2.15 compares the improved fatigue of PMEDM surface.

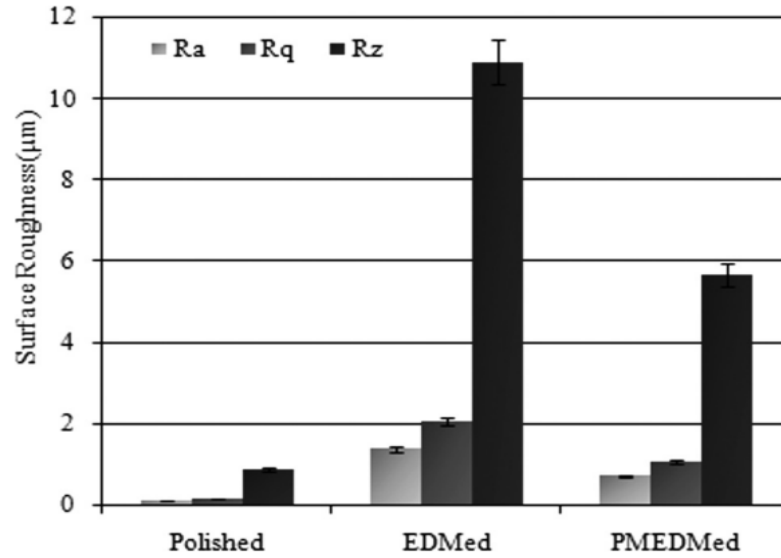


Figure 2.14 Roughness on the surface of the polished, EDMed, and PMEDMed β Ti alloy (Prakash, Kansal, Pabla, & Puri, 2016b)

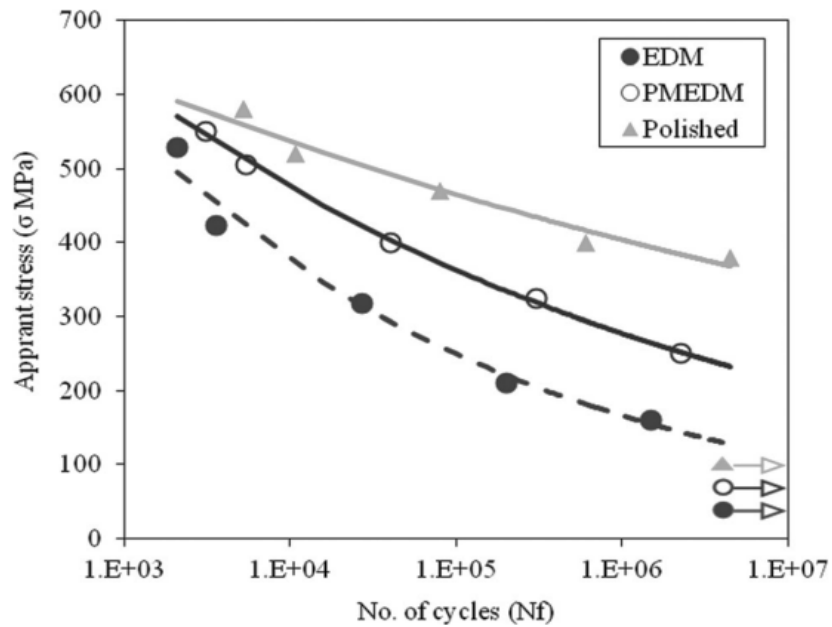


Figure 2.15 S–N curve for polished, EDM, and PMEDM β - Ti alloy (Prakash, Kansal, Pabla, & Puri, 2016b)

2.2 Application of EDM in Machining Shape Memory Alloys and Other Materials

The application of EDM in the machining of Shape Memory Alloys (SMAs) and other materials involves using electrical discharges to erode the material. This technique is particularly useful for SMAs, which are difficult to machine using traditional methods due to their unique properties, such as the ability to return to a pre-defined shape when heated. EDM offers several advantages for machining SMAs and other challenging materials:

(Huang et al., 2003) studied the machining of micro holes in Nitinol using ultrasonic vibration-assisted EDM. This technique was having minimal impact on electrode wear, but the efficiency of machining was improved by more than 60 times.

(Theisen & Schuermann, 2004) explored the machining of NiTi-SMA using the EDM process. The presence of titanium carbide precipitations in the melting zone suggests that the shape memory properties might have been compromised in the Heat Affected Zone (HAZ).

(S. L. Chen et al., 2008) investigated surface modification of NiAlFe ternary shape memory alloy (SMA) through EDM. The roughness on the surface of the EDM-treated NiAlFe SMA follows the empirical form $R_a = \lambda(IP \times \tau P)^\beta$. Enhanced surface hardness is attributed to the presence of oxides such as Fe_2O_3 , Al_2O_3 , and NiO, along with deposited particles on the modified surface.

In order to create silicon nitride-titanium nitride (Si_3N_4 -TiN) composites for implantable biomedical devices, (Bucciotti et al., 2010) looked into the EDM machining process. The recast layer thickness and its roughness was calculated to be 10–20 μm and 1.4–0.15 μm respectively. There were no cracks found and no indications of extensive flaws, around the machined holes.

(J. F. Liu & Guo, 2015) examined wire electrical discharge machining (W-EDM) of Nitinol alloy. A porous recast layer, 2–8 μm thick, was created on the Nitinol alloy using W-EDM. The microtexture on the machined surface is specifically designed to enhance cell attachment. Because nickel elements are present in the recast layer, the changed surface has a microhardness that is about 50% higher than the unmachined surface.

2.3 Some Studies on EDC and ball burnishing assisted EDC

Electrical Discharge Coating (EDC) and its advanced hybrid, Ball Burnishing-Assisted Electrical Discharge Coating (BB-EDC), have gained substantial attention for surface modification of biomedical implants, especially those made from titanium and its alloys. EDC is a non-traditional process that utilizes high-energy electrical discharges in a dielectric medium, often mixed with bioactive powders like hydroxyapatite (HAp) or titanium dioxide (TiO₂), to create a recast layer embedded with functional particles. This technique has demonstrated remarkable improvements in surface roughness, porosity, hardness, wear resistance, and corrosion resistance due to the formation of ceramic-rich coatings. These properties enhance osteoblast adhesion, proliferation, and overall osseointegration, making EDC a viable method for biomedical surface functionalization (Prakash et al., 2015; Ekmekci & Ekmekci, 2016; Prakash & Uddin, 2017). However, EDC is also associated with limitations such as microcrack formation and residual thermal stresses due to rapid cooling, which can adversely affect coating integrity and mechanical performance. To overcome these challenges, BB-EDC has been introduced, combining mechanical burnishing with electrical discharge coating. The burnishing action applies a compressive plastic deformation force using a hardened ball, which smoothens the surface, seals microcracks, and enhances coating adhesion and surface finish. This hybrid method has been shown to significantly improve fatigue strength, wear performance, and biocompatibility of titanium implants (Lin et al., 2001; Yan et al., 2002; Prakash et al., 2022). Additionally, BB-EDC-treated surfaces promote superior cell response and osseointegration due to a synergistic effect of improved topography and chemical bioactivity. Despite the promise, optimal results with BB-EDC depend on carefully controlled parameters such as discharge energy, burnishing speed, and force. Nevertheless, the ability of BB-EDC to enhance both mechanical and biological surface characteristics, while addressing the inherent drawbacks of standalone EDC, positions it as a forward-looking and sustainable solution for next-generation orthopaedic implants.

METHODOLOGY AND ANALYSIS

This chapter presents a brief overview of the study and delineates the format of the report. Research gaps have been found through the introduction and literature analysis, which has resulted in the creation of research objectives meant to fill these gaps. The majority of the literature that currently exists concentrates on the EDM machining of titanium alloys, with a special emphasis on the production of biocompatible surfaces. It also entails examining the β -Ti alloy's corrosion resistance and biocompatibility. Thus, for orthopaedic applications, the current study suggests creating a unique HAP coated nanoporous surface on β -Ti alloy by BB-EDC technique.

3.1 Research gaps

Following the analysis of a thorough literature review, the following research gaps have been found:

- 1) The earlier findings on surface modification of biomaterials by EDM reveal that the improved surface was comprising surface cracks and high ridges of recast layer, which deteriorate the performance of implants.
- 2) No research reported on the electrical discharge cladding process.
- 3) Surface modification of biomedical β -Ti alloy material by EDM has been reported. But, no research on the hybridization of ball burnishing with electric discharge cladding process.
- 4) Analysing of the machining characteristics that contribute to the biocompatibility of Ti alloy such as Recast Layer Thickness (RLT), Surface hardness (HV), Surface Crack density (SCD) and Surface roughness (Ra) needs to reinvestigate by reestablishing of the facts.
- 5) There hasn't been any research on how biological response in terms of Osseointegration, is affected by nano-scale surface roughness on the surface of β -Ti alloy treated by BB- EDC.
- 6) The stress distribution at loading conditions, osseointegration, and implant strength with surrounding bone tissues of BB-EDC β -Ti alloy surface is remaining.

7) There is an inadequacy of information in the existing literature regarding the optimization of BB-EDC parameters for titanium alloys machining.

3.2 The Proposed Research and Formulation of problem

The exceptional specific strength, reduced modulus of elasticity (50 GPa), enhanced compatibility of tissue, and increased corrosion resistance of β -phase Ti-Nb-Ta-Zr titanium alloys have led to their increasing popularity.

Orthopaedic implant success is contingent upon titanium (Ti) osseointegration. In order to promote bone in-growth or osseointegration, implant material must be surface modified to increase biocompatibility and make the surface active. Numerous studies using titanium alloy implant materials focus on surface modification using CVD, PVD, alkali treatment, sol-gel, oxidation, and EDM.

A wear-and corrosion-resistant surface that is nanoporous and biocompatible can be produced by the EDM method, according to earlier research. According to earlier research, powder suspended in PMEDM or EDM's dielectric fluid has the potential to be an effective approach for surface modification by changing the substrate's surface integrity. The result was an even, nanostructured surface that was resistant to corrosion and abrasion.

However, under a corrosion and fatigue environment, the surface featured high ridges and microcracks in the recast layer, which reduce the effectiveness of implants. According to earlier research, ball burnishing is a promising method for improving surface hardness and removing high ridges of surface roughness, both of which enhance a material's resilience to corrosion and wear.

Using ball burnishing aided electrical discharge cladding (BB-EDC), the current study aims to establish a biocompatible surface with varying surface texture scale on β -phase Ti alloy for biomedical implant applications.

3.3 General hypothesis

Surface modification is our hypothesis with main focus is to alter surface topography & surface chemistry by utilizing the ball burnishing assisted electrical discharge cladding, so as to improve top surface layer. The TiO₂ loaded HAP nanoporous layer is cladded and the ball burnishing setup is used as a post finishing operation to improve

the surface features. This hypothesis is however in its infancy and further work will be needed to verify it.

3.4 Research objective

After a thorough review of the literature, the various objectives which have been chosen to fill the research gaps in the present research work are as follows:

1. To design and develop an experimental setup for the ball burnishing assisted electrical discharge cladding (BB-EDC).
2. To determine the effect of various process parameters such as pulse current, pulse duration, supply voltage, powder concentration, spindle speed on the surface integrity of β -Ti alloy using BB-EDC. The surface morphology, surface roughness, recast layer thickness (RLT), micro-cracks density, and surface hardness (HV) are the evaluations against the surface integrity check.
3. To determine the optimal condition of process parameters of BB-EDC for the synthesis of biocompatible layer on β -Ti alloy.
4. To study the effect of surface characteristics and morphology obtained by BB-EDC- treatment on the corrosion resistance, fatigue resistance, wear resistance of β -Phase Ti alloy.
5. To evaluate the biological response of BB-EDC-treated surface on the bioactivity of β -Phase Ti alloy for implant applications using apatite growth analysis.

3.5 Chapter Wise Details

Chapter 1: Introduction

This chapter introduces the growing need for orthopedic implants due to age-related bone diseases. It highlights the limitations of existing implant materials and the critical role of biocompatibility, corrosion resistance, and mechanical strength. Special focus is given to β -Ti alloys, known for their low modulus and excellent compatibility. The chapter stresses the need for advanced surface modification techniques and presents Electrical Discharge Cladding (EDC) and Ball Burnishing as promising solutions for improving implant surface integrity and performance.

Chapter 2: Literature Review

This chapter reviews past research on surface modification of biomaterials using EDM, focusing on titanium and its alloys. It discusses in vitro and in vivo studies, showing improved osseointegration, wear resistance, and corrosion protection. Limitations like surface cracks and recast layers in conventional EDM are noted. The review identifies a lack of studies on combining ball burnishing with EDC, setting the stage for the proposed hybrid BB-EDC approach for enhancing the biofunctional surfaces of β -Ti implants.

Chapter 3: Methodology and Analysis

Here, the research methodology is defined, beginning with identification of gaps and formulation of objectives. The BB-EDC process is introduced as a hybrid surface treatment method aimed at reducing surface defects and improving biocompatibility. Objectives include evaluating process parameters, surface features, corrosion, fatigue, wear behavior, and biological response. The chapter lays out a clear research path for investigating BB-EDC as a novel technique for biomedical surface engineering.

Chapter 4: Experimental Design and Analysis

This chapter presents the experimental setup and process design for BB-EDC, including the materials used (β -Ti alloy, TiO_2 -HAP powders), machine configurations, and testing protocols. The Taguchi method is used for process optimization. Parameters such as surface roughness, hardness, recast layer thickness, corrosion resistance, and fatigue strength are measured. Advanced characterization tools like SEM, EDS, and profilometers are used to evaluate the modified surfaces, confirming BB-EDC's ability to create dense, biocompatible layers.

Chapter 5: Results and Discussion

This chapter outlines the input process parameters (peak current, pulse duration, voltage) and their effects on output responses (surface roughness, hardness). A cause-effect analysis identifies key machining factors. Pilot experiments help establish parameter ranges. It emphasizes the importance of controlling these parameters to achieve optimal surface features for enhanced osseointegration, wear resistance, and corrosion resistance in β -Ti implants.

Chapter 6: Conclusions and Future Scope

The final chapter evaluates the performance of the BB-EDC treated β -Ti alloy. Corrosion testing, wear analysis, fatigue testing, and biocompatibility assessments are conducted. Results confirm that BB-EDC enhances implant surface integrity, improves biological response, and increases durability. The technique shows potential for clinical applications, combining the benefits of both electrical discharge and mechanical surface treatments.

EXPERIMENTAL DESIGN AND ANALYSIS

This chapter initially explores the specifics of the experimental configuration for BB-EDC process. The workpiece, tool electrode, dielectric, powder additives used in the experimental test equipment, and EDM machine are all discussed with the experimental setup.

4.1 Research design and Methodology

- I. To begin with, the BB-EDC experimental setup was created in CITCO – Industrial Development Cum Facility Centre (IDFC), Industrial Area, Phase – I, Chandigarh.
- II. The β -Ti alloy was obtained under the guidance of my co-guide, Dr. Chander Prakash, who had previously completed its casting during his earlier research at the DMRL lab located at Telangana, Hyderabad, India. The alloy was produced by blending powders of titanium (Ti), niobium (Nb), tantalum (Ta), and zirconium (Zr). The Ti–35Nb–7Ta–5Zr (TNTZ) alloy was created via vacuum arc melting technique.
- III. The additive powders (TiO₂ loaded HAP used as 0 gm/l for EDM and 30 gm/l for BB-EDC) were sourced from NANOSHEL, which is located at Derabassi, Punjab.
- IV. Commercially pure titanium (CP-Ti) alloy tool electrode is utilized for testing.
- V. Next, using our experimental setup of a BB-EDC machine, experimentation was carried out by suspending the powder in a dielectric fluid with varying concentrations and powder types, and by adjusting the various input process parameters like peak current (I_p), pulse duration (T_{on}), supply voltage (V), and spindle speed.
- VI. Surface roughness (SR), surface hardness, thickness of the recast layer, corrosion resistance, cracks, resistance to wear and fatigue resistance have been used to gauge process performance.

- VII. A connection has been made between different input/output process parameters with help of Taguchi design, followed by an analysis of variance (ANOVA).
- VIII. A variety of techniques, including FE-SEM, EDS/EDX analysis, Surface Hardness and Surface Roughness Testers, and Universal Testing Machines (UTM) 100 KN for fatigue testing, have been used to analyze the machined surface in order to comprehend the surface topography. After the fatigue test, surface cracks were further examined using FE-SEM fractography analysis on the failed specimens.
- IX. The mechanical properties like surface roughness, surface hardness & the fatigue strength of BB-EDC β -Ti alloy has been determined with different batches of test specimen such as untreated, EDM, BB-EDC.
- X. Next, a cell culture test (in vitro) was completed to evaluate the biological properties of the BB-EDC surface of the β -Phase Ti implant in terms of osseointegration. The β -Phase Ti test specimens, including untreated, EDM and BB-EDC, were analyzed in vitro utilizing many metrics for cell growth, proliferation, and viability.

An appropriate experimental plan is essential for achieving high-quality research results.

An appropriate design matrix, based on mixed orthogonal array L-18 was used to conduct experiments, so as to infer quality.

A summary of the research process is illustrated in Figure 4.1. This clearly indicates the basic flow of the whole process.

4.2 Design and Fabrication of Experimental Setup of BB-EDC

In this study, a novel β -Phase Titanium alloy (Ti-Nb-Ta-Zr), chosen for its extensive use in the biomedical field, serves as the work material.

The elemental composition of the β -Phase titanium alloy is provided in the Table 4.1 Composition of β -Ti alloy.

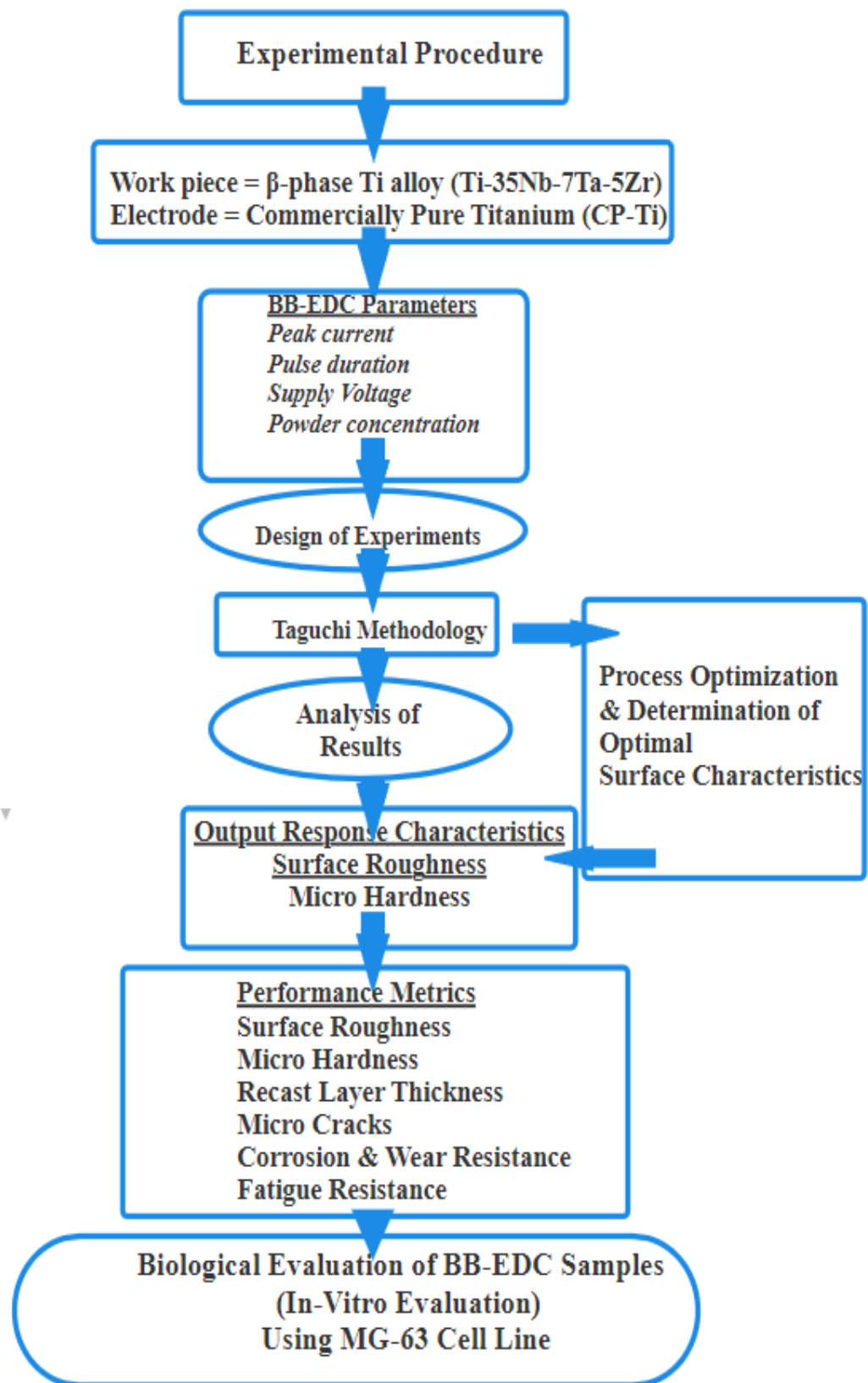


Figure 4.1 Flowchart depicting the steps undertaken in the current research study.

Table 4.1 Composition of β -Ti alloy

Chemical composition of β -Ti				
Elements	Titanium (Ti)	Niobium (Nb)	Tantalum (Ta)	Zirconium (Zr)
Wt %	53	35	7	5

In a typical scenario, and as depicted in Figure 4.2, the schematic representation of the BB-EDC setup illustrates the use of (HAP-Distilled Water) (5) as a slurry mixture.

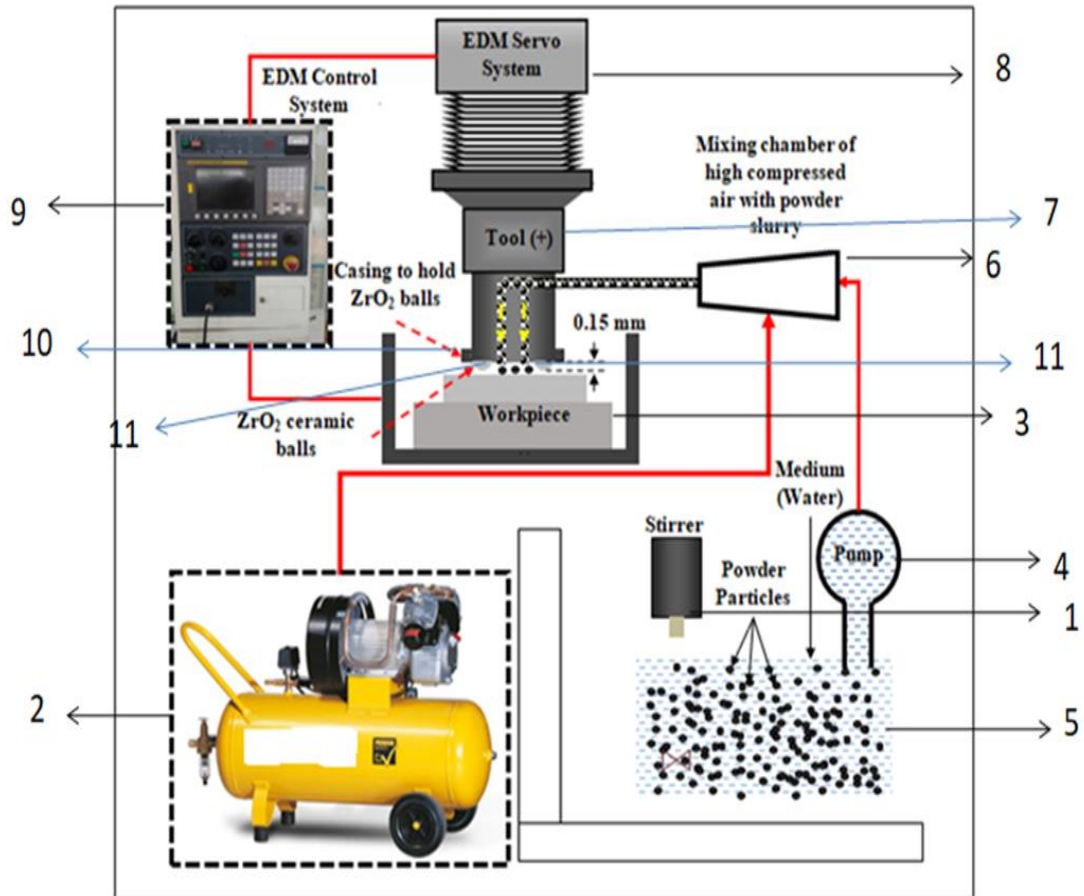


Figure 4.2 Schematic Diagram illustrating the Ball-Burnishing Assisted Electrical Discharge Cladding (BB-EDC) setup

A stirrer (1) aids in maintaining a homogeneous mixture. Additionally, a pump (4) is employed to transport the homogeneous mixture to a mixing chamber (6). Compressed

air, generated by a compressor (2), is combined with the slurry and directed onto the workpiece surface (3).

Electrical potential is applied through a tool electrode (7), resulting in the generation of a significant amount of sparks and thermal energy, further causing plastic deformation of the specimen's top surface.

Simultaneously, a tubular electrode directs highly compressed slurry into the processing zone, while (ZrO₂) balls (11) rotate at high speed to execute the BB-EDC process.

A casing (10) is utilized to secure the (ZrO₂) balls (11) in place. An EDM servo system (8) automatically adjusts the gap, while an EDM control system (9) regulates the EDM servo system. Very fine and small powder particles (ranging from 0.110 to 0.200 µm) are employed, including pulse current (ranging from 5 to 15 A), pulse duration (ranging from 50 to 200), supply voltage (ranging from 60 to 120), duty cycle (0.5%), rotational speed (1200 RPM), and the composition of the Hydroxyapatite layer using a blended mixture of TiO₂ (30 wt%) & HAP (70 wt%) at 30g/l.

Similarly another Figure 4.3 (a) shows the general Setup of EDM and Figure 4.3 (b) shows the rotary tool electrode setup used for BB-EDC.

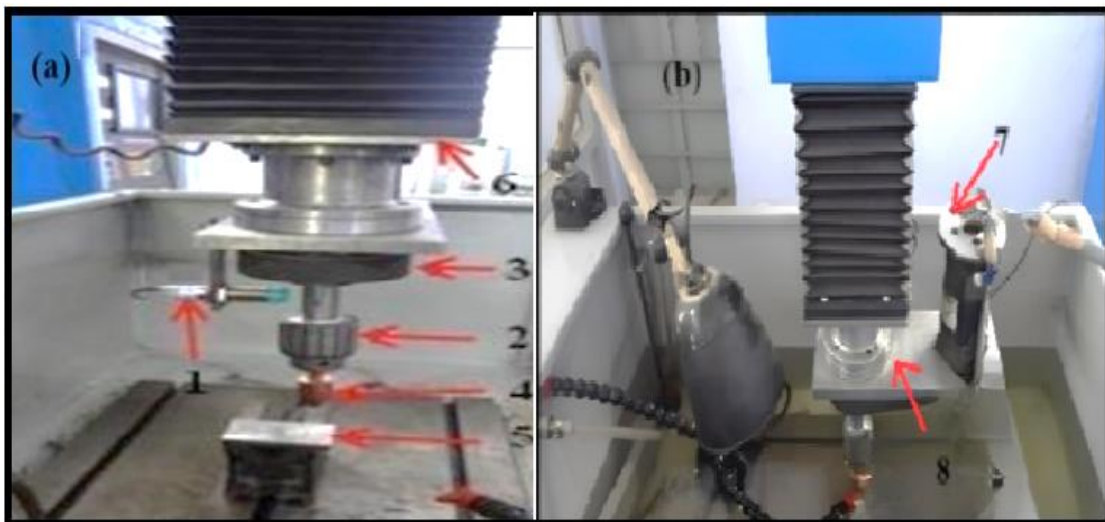


Figure 4.3 General appearance (a) EDM setup: 1. Speed detector 2. Chuck 3. Belt 4. Holder for tool 5. Workpiece 6. Upper plate (b) BB-EDC setup: 7. DC motor 8. Setup plate for BB-EDC Tool electrode holder

The process parameters for BB-EDC are outlined in

Table 4.2.**Table 4.2 Processing parameters for BB-EDC**

Parameters	Descriptions
Tool Electrode	Commercially pure Ti
Workpiece	β -Ti
Type of Process	EDM, BB-EDC
Peak current (Ip)	5, 10, 15
Pulse duration (Ton)	50, 100, 200
Supply Voltage (V)	60, 90, 120
Duty cycle (%)	0.5
Rotational speed	1200
Servo reference voltage (V)	20
Powder and size (μ m)	Hydroxyapatite (HA) and <5
Concentration (gm/l)	0 (EDM) and 30 gm/l

In another Figure 4.4 illustrates the formation of a functionalized concentrated HAP-cladded layer.

The experimental setup is explained in regard to the workpiece, tool electrode, dielectric, powder additives utilized in the experiment, experimental test equipment, and EDM machine. To regulate the homogeneous mixture of HAP and distilled water, a stirrer is used with motor assistance.

The tubular electrode is used to process high compressed slurry towards the workpiece, where the molten metal is quenched by high compressed air and water, leading to the deposition of molten metal on the specimen's surface. HAP-powder particles (9) are cladded onto the specimen's top surface.

The EDC technique creates a recast layer, consisting of discharge craters (8), high ridges of re-deposited metal (4), and microcracks. Amid BB-EDC process, ZrO₂ balls (3) rotating at high speed exert compressive load on the deformed superficial layer resulted out of the EDC process. This compressive action by the ZrO₂ balls compacts the re-deposited metal, resulting in a dense and compact layer of HAP (7).

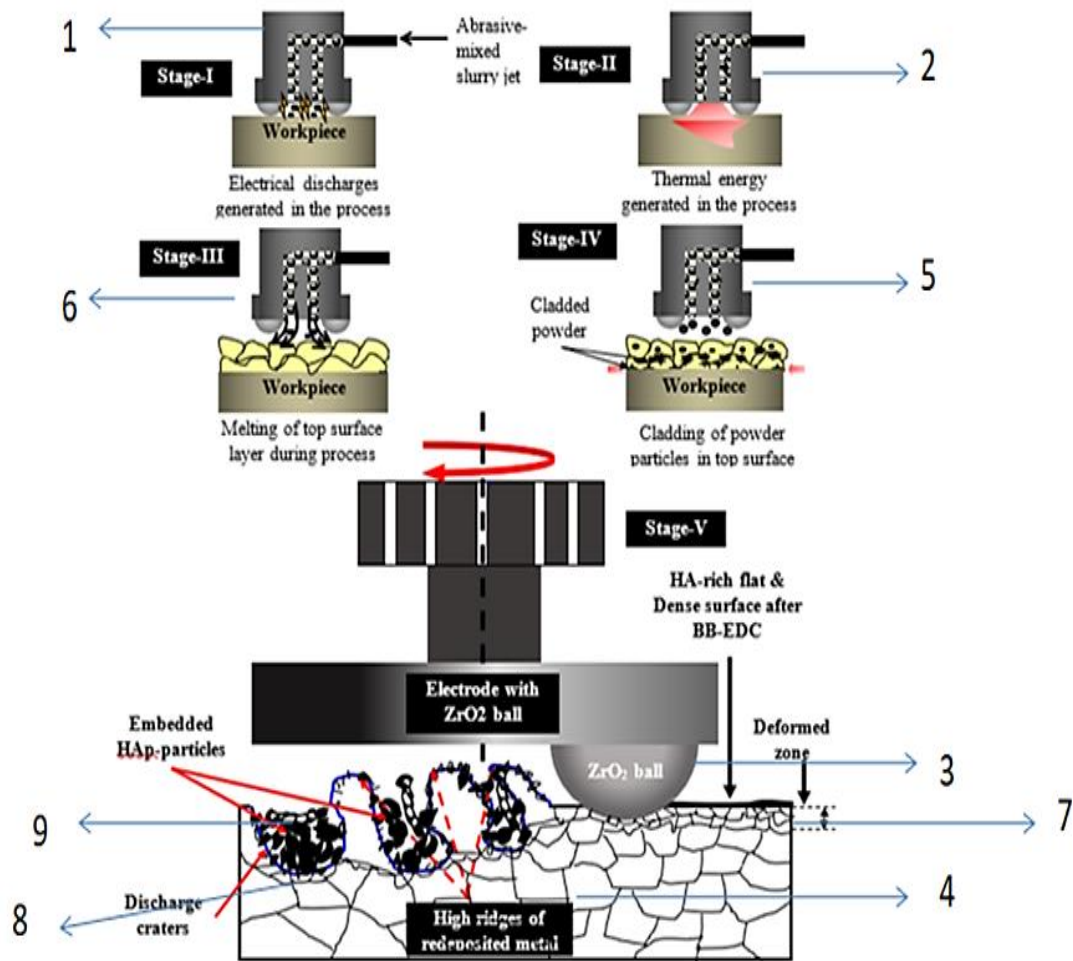


Figure 4.4 Mechanism of BB-EDC process

4.3 Choosing the workpiece material and its characterization

The β -Ti alloy was selected for use as the implant material. The chemical makeup has already been discussed in the previous section. The β -Ti alloy was obtained under the guidance of my co-guide, Dr. Chander Prakash, who had already completed its casting during his earlier research at DMRL located at Telangana, Hyderabad, India. The alloy was produced by blending powders of Ti, Nb, Ta, and Zr. The Ti-35Nb-7Ta-5Zr (TNTZ) alloy was created via vacuum arc melting technique to have compact and homogeneous ingots. Figure 4.5 displays the as-cast β -Ti in pancake form, weighing 600 grams.



Figure 4.5 As cast β -phase Ti alloy by vacuum arc melting Process

The mechanical and physical properties of the implant compared to human bone are shown in Table 4.3.

Table 4.3 Comparison of β -Ti implant compared to human bone (Prakash, Kansal, Pabla, & Puri, 2016b)

Properties	β -Ti alloy	Human bone
Hardness	380 Hv	—
Density (kg/m^3)	5700	2000
Elastic modulus (GPa)	55	15–20
Strength (MPa)	560–950	150
Poisson's ratio	0.33	0.3
Melting point ($^{\circ}\text{C}$)	1765	—
Coefficient of expansion ($10^{-6} \text{ }^{\circ}\text{C}^{-1}$)	9.03	—

The β -Ti alloy stands out as a biomaterial for crafting implants when compared to other metallic biomaterials such as SS-316L and Ti-6Al-4V alloy. This distinction arises from its closely matched elastic modulus to that of bone, alongside its outstanding mechanical properties and biocompatibility.

Energy dispersive spectroscopy (EDS) was used for elemental composition and for morphological composition, FE-SEM-7600F by JEOL and optical microscope was utilized. The test specimens were mirror polished to get appropriate microstructure measurements. After that, samples were etched for roughly about 15 seconds using Kroll's reagent (10 vol.% HF and 5 vol.% HNO₃ and remaining water). Figure 4.6 shows the microstructure of the polished (un-machined) surface of the as-cast ingot of

the β -phase Ti alloy through an optical SEM micrograph and an optical microscopic picture.

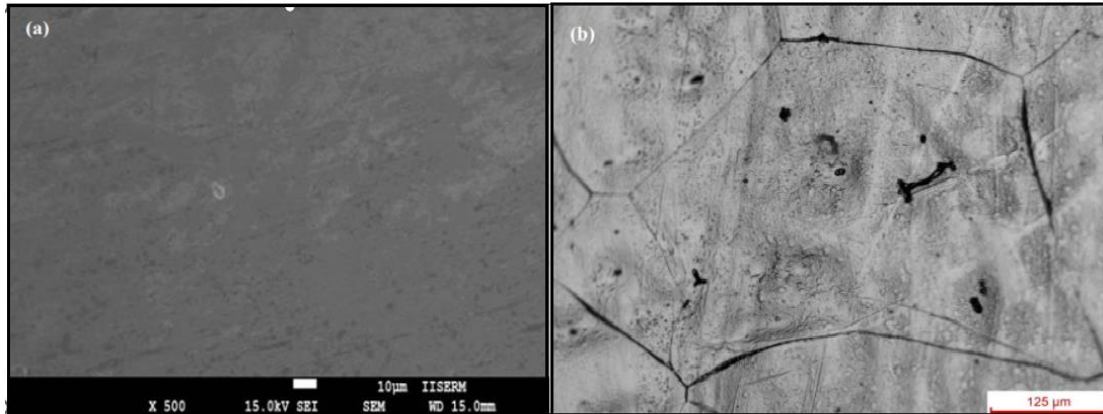


Figure 4.6 Microstructure of β -Ti alloy (a) SEM micrograph and (b) Optical image (Prakash, Kansal, Pabla, & Puri, 2016b)

4.4 Selection of tool electrode material

A rotary tubular electrode (15 mm diameter) made up of commercially pure titanium was used and connected with an arrangement to supply high compressed slurry at a pressure of 15 MPa.

The blended mixture of TiO_2 (30 wt%) loaded HAP (70 wt%) of size $\sim 0.112 \mu\text{m}$ was mixed in distilled water at a concentration of 30 g/l to prepare the slurry for cladding purpose.

To manufacture a tool electrode with a diameter of 15 mm, a cylindrical rod was stepped turned. This rod was then employed in machining tests to help place the tool electrode at the workpiece surface. In Figure 4.7, the electrode picture is shown. To help remove the carbon that had been deposited, the electrode's machining surface was ground after each experiment.



Figure 4.7 Shape and size of tool electrode.

The chemical makeup of the tool material is described in full in Table 4.4 Chemical composition of tool electrode material.

Table 4.4 Chemical composition of tool electrode material

Material	Commercially Pure Ti (CP-Ti)			
Elements Present	Ti	Fe	O	C
Percentage (%)	99.55	0.25	0.2	.08

4.5 Dielectric fluid and powder Particles

Distilled water served as the dielectric fluid in this study. The cladded layer consists of a blended mixture of TiO₂ (30 wt%) and HAP (70 wt%). The TiO₂-loaded HAP powder particles have sizes ranging from 0.110 to 0.200 µm. The slurry required for cladding is prepared by mixing the blended mixture (TiO₂ + HAP) with distilled water at a concentration of 30 g/l. Using distilled water as the machining fluid eliminates the risk of fire and ensures a safe working environment. However, employing distilled water as the machining fluid in EDM may lead to a lower material removal rate when machining large surfaces and increased erosion of the machine tool (Y. Liu et al., 2010).

4.6 Tool Electrode utilized in Ball Burnishing

The recast layer formed following the EDC process have craters, micro cracks, and raised ridges of redeposited metals. To mitigate these surface defects, a ball burnishing process was employed using two hard ceramic (ZrO₂) balls with a diameter of 5mm as in Figure 4.8. The balls are mounted symmetrically using a rotary tool of diameter 15 mm, within a ring electrode casing, where only 0.15 mm of the spherical balls are extending out for the purpose of ball burnishing by applying compressive force to the workpiece surface. A central hole of 1 mm is provided to facilitate the flow of TiO₂-loaded HAP slurry. This ball-burnishing process flattens the cladded surface and yields a dense, compact layer.

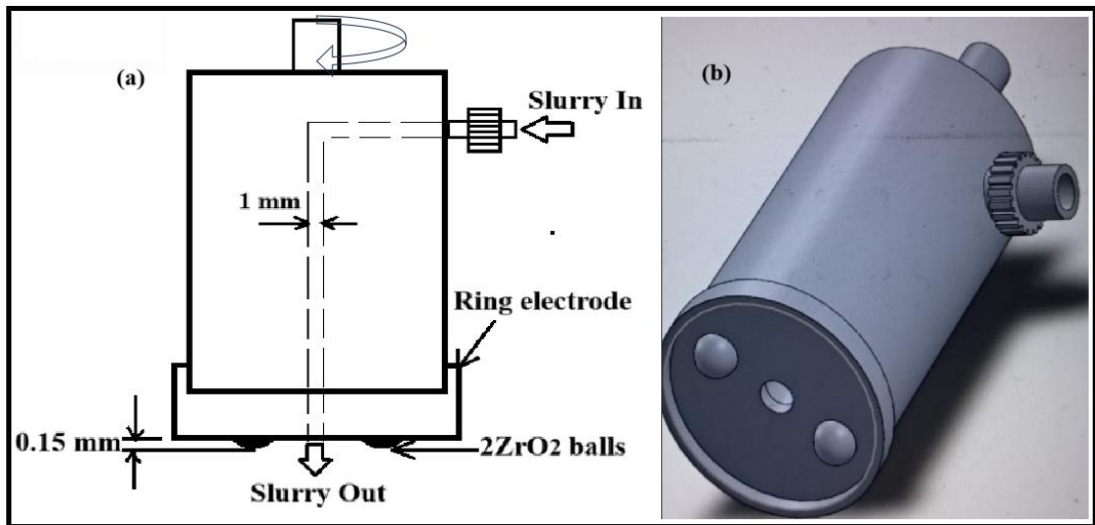


Figure 4.8 (a) 2D Schematic of BB-EDC tool electrode (b) 3D Schematic

4.7 PROCESS PARAMETER SELECTION

Based on Taguchi's orthogonal array (OA-L 18), the input process parameters were set at the range and level after the output response characteristics and input process parameters were chosen.

4.7.1 Choosing Output Response Characteristics

A cause-and-effect diagram was created to identify potential factors that could impact machining responses like surface roughness and microhardness, as illustrated in Figure 4.9.

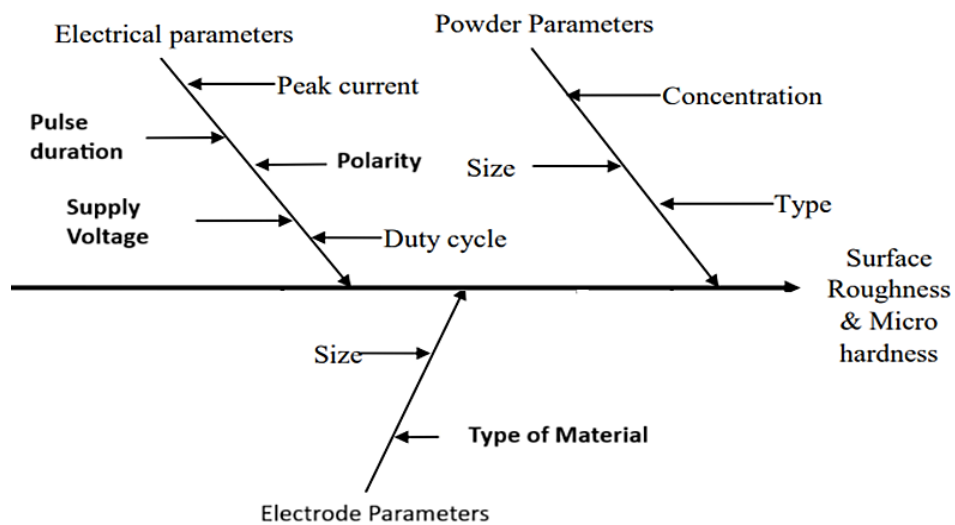


Figure 4.9 Cause and effect diagram of possible parameters in BB-EDC

Surface roughness, topography, chemistry, and hardness are pivotal factors influencing the biocompatibility of implant materials. Surface roughness, in particular, facilitating the growth, spreading, and proliferation of bone tissues. A properly textured surface can promote osseointegration. Additionally, surface hardness is essential for ensuring a strong and durable bond between the bone and the implant interface. A sufficiently hard surface can withstand the mechanical stresses and strains encountered in the body, enhancing the longevity and stability of the implant. In essence, optimizing surface characteristics, including roughness and hardness, is essential for enhancing the biocompatibility and performance of implants, ultimately contributing to their success and stability within the body.

4.8 Output Response Characteristics

The response parameters of the machined surface selected for this investigation are surface roughness and microhardness.

4.8.1 Surface Roughness (SR)

The average roughness (R_a) is determined from the center line across the entire surface area, as illustrated in Figure 4.10.

$$R_a = 1/L \int |y(x)| dx \text{ from } 0 \text{ to } L$$

4.1

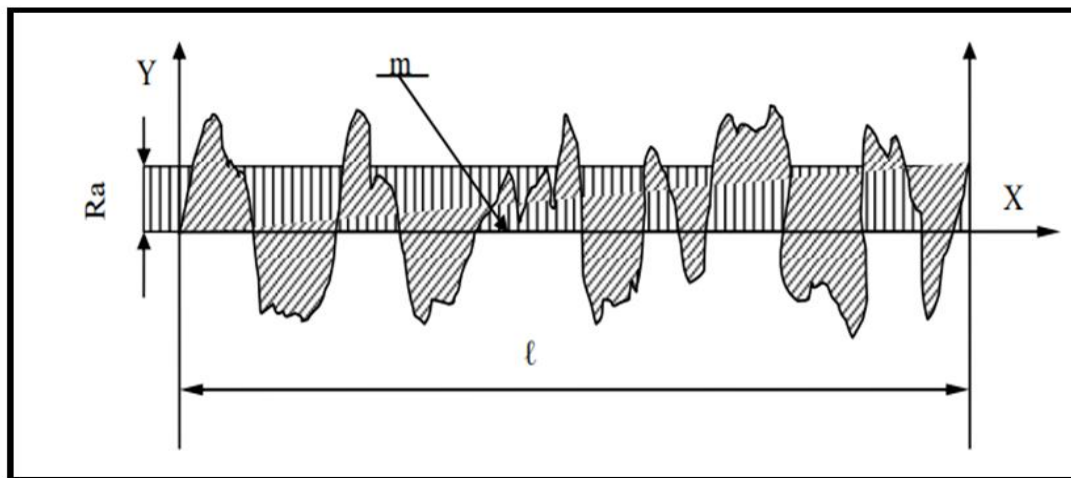


Figure 4.10 Representation of all variance in the roughness profile.

In this context, L represents sampling length, and y denotes the profile curve. The average surface roughness (R_a) of the machined Ti-alloy specimens was measured using a contact-type Surface Profilometer (manufactured by Mitutoyo Surface Roughness Tester, SJ210, Japan), following the ISO-1997 standard. Three measurements were taken at different locations to determine the average R_a value.

4.8.2 Microhardness (MH)

Microhardness refers to the hardness of a material determined through instruments equipped with small indenters. As a result, these instruments use less force than conventional measuring tools, making it easier to take measurements on thinner sheets or smaller specimens that conventional instruments might not be able to accurately measure. A schematic illustration of the indentation mark left by the tip of a diamond indenter for measuring microhardness is shown in Figure 4.11.

The microhardness for the surface of the samples was examined using a Vickers hardness tester (SHIMADZU, Japan: HMV-G21ST). An indentation loading of 2.25 N was applied for 10 s to assure the indentation.

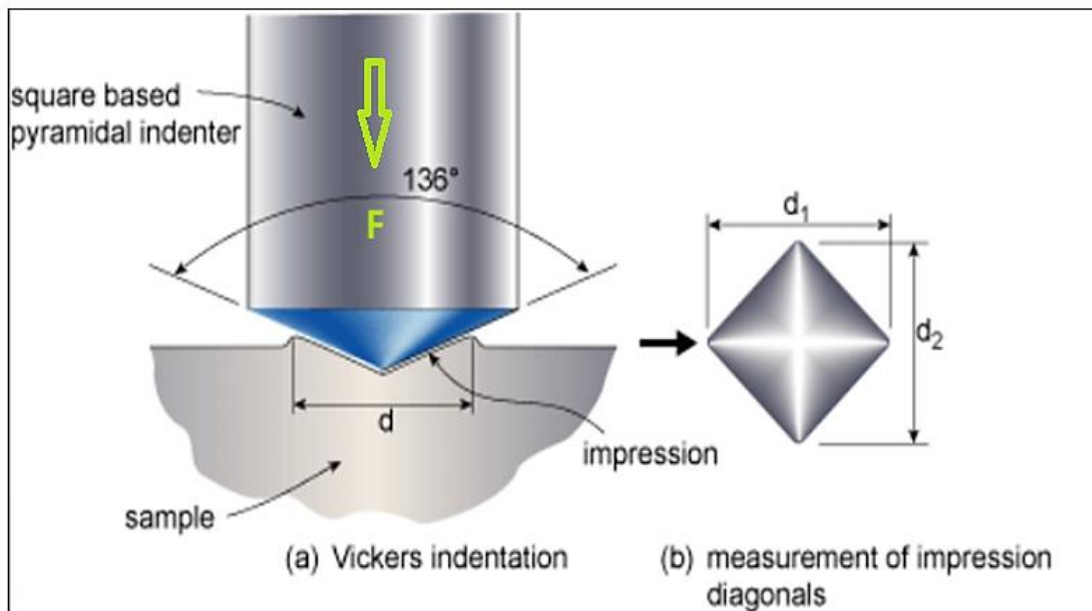


Figure 4.11 Schematic representation of indentation mark of diamond-indenter.

$$HV = 1.854 (F/d^2)$$

4.2

4.9 Input process parameters and pilot study

The selection of process parameter ranges for the experiments was decided using a literature review and initial experiments conducted through (OFAT) approach. Also osseointegration is significantly influenced by microscale roughness and nanoscale topological features. Below are some key process parameters known to impact surface roughness responses:

4.9.1 Peak Current (I_p)

This current results in localized melting and vaporization of the workpiece material, due to high-energy electrical discharge. As per Figure 4.12 (Prakash et al., 2017) illustrate the effect of pulse current on surface roughness (SR) of β -Ti alloy. In standard EDM, SR increases with higher pulse current and duration due to intense heat causing significant melting and damage to the workpiece surface, resulting in larger and deeper craters. Conversely, in powder-mixed EDM (PMEDM), SR is lower for the same parameters. At a 4 g/l powder concentration, SR remains minimal regardless of pulse current settings.

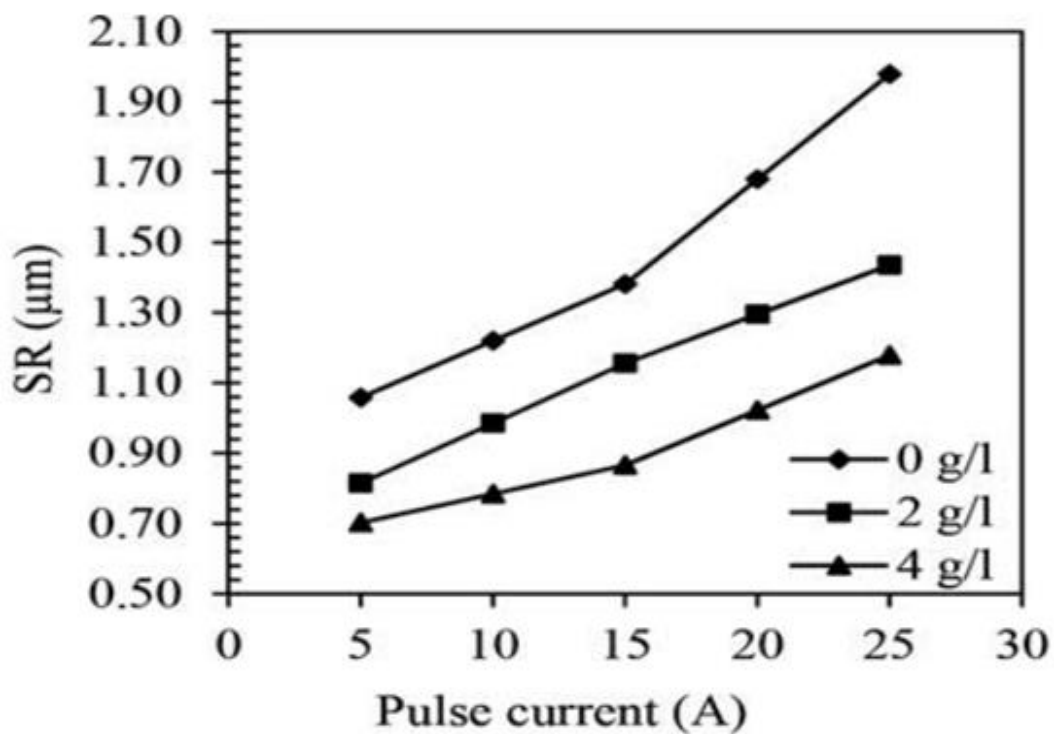


Figure 4.12 —Effect of pulse current on SR (Prakash et al., 2017)

Also as per the available literature on Ti-based alloy, researchers have observed that machining of Ti-based alloy typically commences at peak currents exceeding 5A. Operations below this threshold prolong machining times and lead to inefficient results. Furthermore, higher peak currents correlate with increased surface roughness. (Peng et al., 2010) also found EDM using a peak current of 5A demonstrates good cell adhesion and growth. (Harcuba et al., 2012) noted EDM using a peak current of 25A exhibits favourable bone ingrowth.

Conversely, (Ndaliman et al., 2013) observed that surface microhardness initially rises with peak current, but beyond 15A, it begins to decline. According to literature, surface microhardness can be enhanced with peak currents exceeding 15A as operational parameters.

4.9.2 Pulse Duration (Ton)

Pulse duration in EDM refers to the length of time during which the electrical discharge occurs. It represents the duration of the voltage application between the electrodes in a single discharge cycle.

This duration encompasses the actual machining process and is expressed in microseconds (μs). The voltage between the electrodes is applied for a duration of T_{on} during the pulse on-time and pause times of each discharge cycle. Wider and deeper craters are produced as a result of increased spark energy produced by longer pulse durations.

As per Figure 4.13 (Prakash et al., 2017) illustrates (SR) of β -phase titanium (β -Ti) alloy. In standard EDM, SR increases with longer pulse durations due to the intense heat causing significant melting and damage to the workpiece surface, resulting in larger and deeper craters.

Conversely, in powder-mixed EDM (PMEDM), SR is lower under the same conditions. At a 4 g/l powder concentration, SR remains minimal regardless of pulse duration settings.

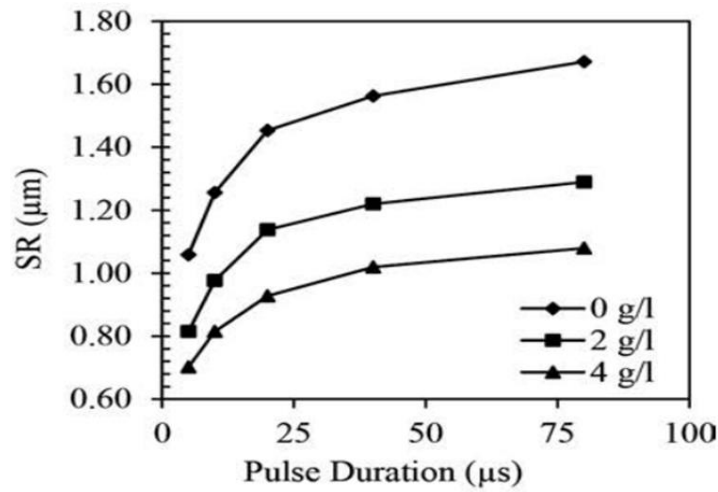


Figure 4.13 Effect of pulse duration on SR (Prakash et al., 2017)

Also (Peng et al., 2010) stated that for durations ranging from 10 to 60μs also exhibits favourable cell adhesion and growth. (S.-L. Chen et al., 2008) noted that EDM is capable of generating a biocompatible surface with pulse durations under 60μs. (Prakash et al., 2017) studied the effect of extended pulse length and current, for β-Ti alloy.

4.9.3 Effect of voltage on surface roughness

(Daneshmand et al., 2013) Figure 4.14 illustrates how varying voltage impacts surface roughness in NiTi alloy through rotational and traditional spark methods. As the voltage is raised to 250V, there is a corresponding increase in surface roughness.

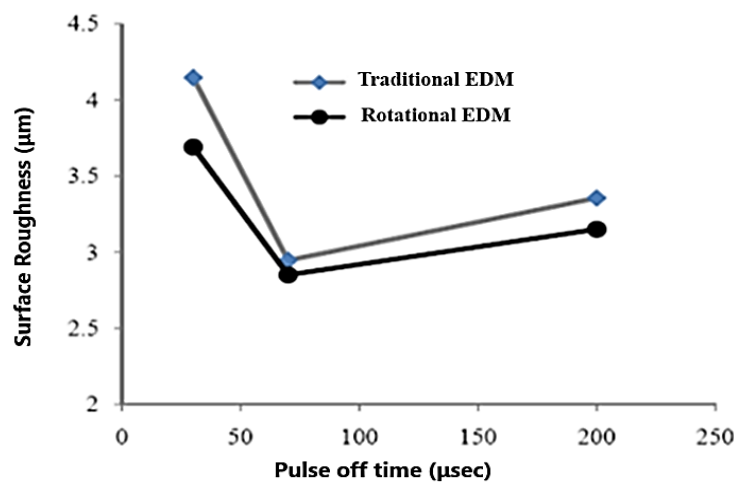


Figure 4.14 Surface roughness comparison: traditional EDM and rotational EDM at various Gap voltage (Daneshmand et al., 2013)

4.9.4 Selection of Range of Process Parameter

The process parameter ranges are detailed in Table 4.5. Within these ranges, specific levels of process parameters will be chosen for implementation in the Taguchi experimental design methodology.

4.10 Choosing Performance Metrics

4.10.1 Corrosion Resistance

Corrosion refers to the gradual degradation of unprotected metal surfaces due to exposure to elements such as atmosphere, acids, gases, alkalies, and more.

This process leads to the formation of pits, perforations, and eventual dissolution of metal parts. Materials with high corrosion resistance exhibit minimal or no signs of corrosion, ensuring longevity and reliability in various applications.

The capacity of implants or implant surfaces to resist damage from oxidation or other chemical processes is known as corrosion resistance. In this work, we use potentiodynamic polarization measurements to evaluate the corrosion resistance of BB-EDC surfaces.

Table 4.5 Process parameters

Parameters	Symbols	Units	levels		
			1	2	3
Process type			EDM	BB-EDC	
Peak Current	I_p	Ampere	5	10	15
Pulse duration	T_{on}	μs	50	100	200
Voltage	V	Volt	60	90	120
Duty cycle	τ	%	0.5	Constant parameters	
Powder Concentration	P_c	g/l	0 (EDM) and 30 g/l (BB-EDC)		

Powder type			TiO ₂ loaded HAp	
Servo reference voltage (V)	V	Volt	20	
Rotational speed	n	Rpm	1200	
Dielectric fluid			Distilled water	
Pressure	P	MPa	15	
Polarity			Reverse (Tool positive)	

4.10.2 Fatigue Performance

The ultimate tensile stress limit or yield stress limit of the material may not always be reached by the stress levels generating this damage. The largest stress occurring a material for a predetermined number of cycles without breaking is called fatigue strength, commonly referred to as endurance strength or fatigue limit.

4.10.3 Wear test

Pin-on-disk type friction test was used to measure the wear of untreated, EDM and BB-EDC specimens of β -phase Ti alloy. The wear test is conducted in accordance with ASTM G 99 standards (Bheemappa et al., 2023) as shown as Figure 4.15 below.

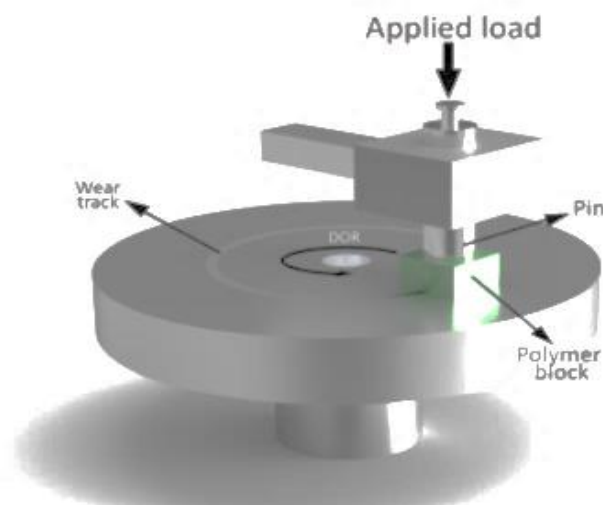


Figure 4.15 Wear Setup

4.10.4 Biocompatibility

The biocompatibility of a material can be assessed by observing the biological responses elicited when human osteoblast cells comes in contact with the implant surface. In this research, the biocompatibility of BB-EDC surfaces is evaluated through in vitro testing using MG-63 human osteoblastic cell line, focusing on cell attachment, proliferation, and differentiation activities.

4.11 Taguchi Analysis

Widely used in engineering, the Taguchi method optimizes performance by carefully selecting process parameters and their levels. It combines mathematical and statistical tools to enhance process improvement and product development, ensuring optimal parameter combinations with minimal variability.

The Taguchi technique is a useful tool for creating intricate systems and procedures that are of the highest caliber. It enables the development of robust designs by minimizing the impact of variation caused by uncontrolled factors on response parameters. The methodology is utilized in the current study, incorporating the steps outlined in Figure 4.16.

4.12 Experimentation Design using Taguchi's Analysis

Based on Taguchi parametric design, the standard L-18 orthogonal array was employed (Lin et al., 2001).

Also L-18 utilized one two-level factor and seven three-level factors. The objective was to examine the effects of key factors of the developed process on quality characteristics with minimal experiments. Four input parameters were considered: "Type of Process," "Peak Current," "Pulse Duration," and "Supply Voltage." For the "Type of Process" parameter, two levels were selected (EDM and BB-EDC), while the remaining input parameters were varied at three levels each (Peak Current: 5 amperes, 10 amperes, 15 amperes; Pulse Duration: 50 μ s, 100 μ s, 200 μ s; Supply Voltage: 60 volts, 90 volts, 120 volts). Surface Roughness (in μ m) was measured as the response parameter for each trial condition and replication. To ascertain how different parameters affected the chosen quality attributes, the main effects were plotted. The highest S/N Ratio

technique was used to optimize the parameters, and confirmation experiments were carried out to confirm the ideal quality attributes.

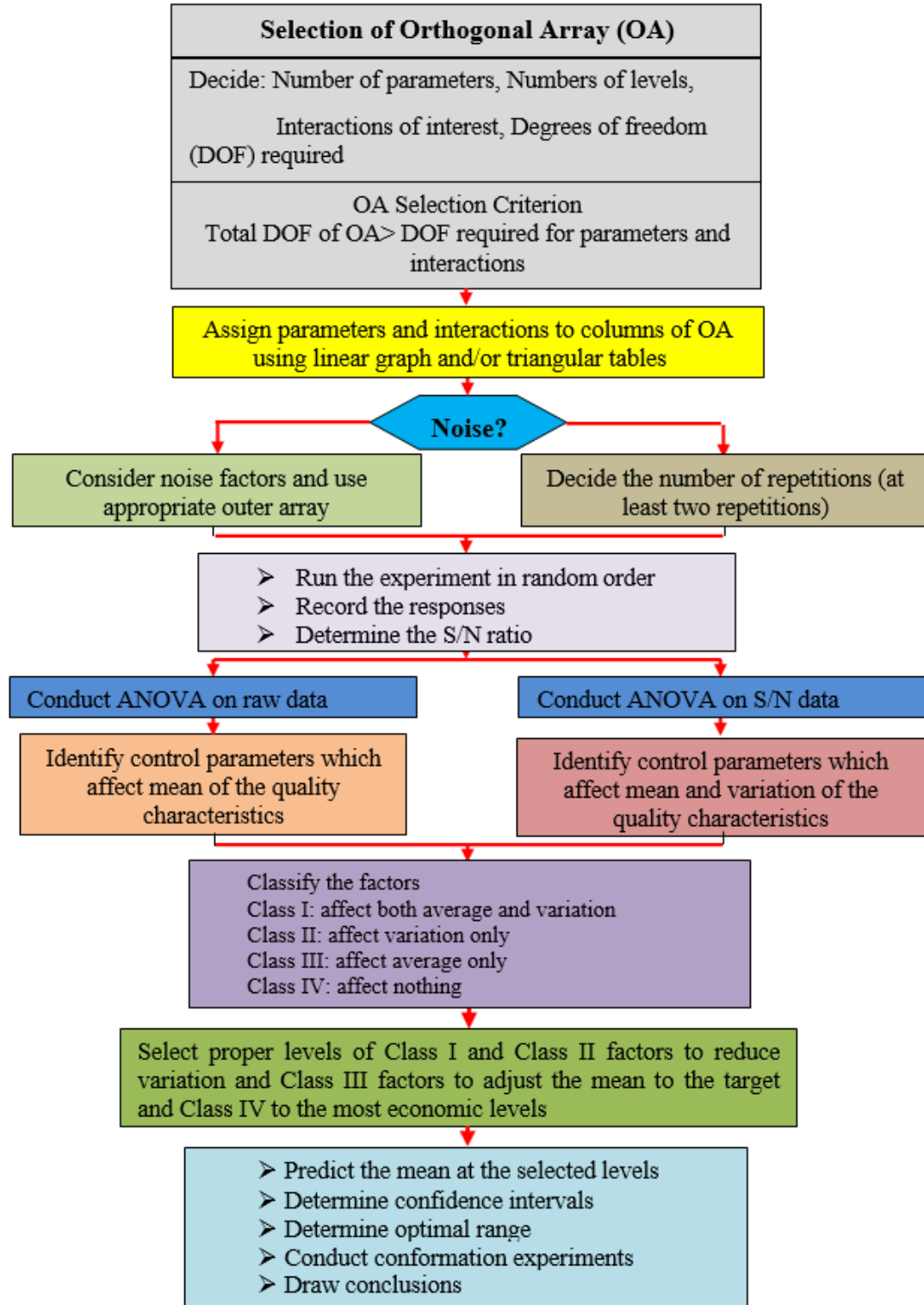


Figure 4.16 Flow chart for Taguchi design of experiment and analysis

4.13 Characterization Techniques

Characterization techniques are methods used to measure and/or inspect features of interest so that they can be described either quantitatively or qualitatively. Quantitative results involve precise measurements against a known scale.

4.13.1 Fatigue Testing

A 100kN Universal Testing Machine (UTM, BIS), which was used for fatigue testing, is as shown in Figure 4.17 (a).

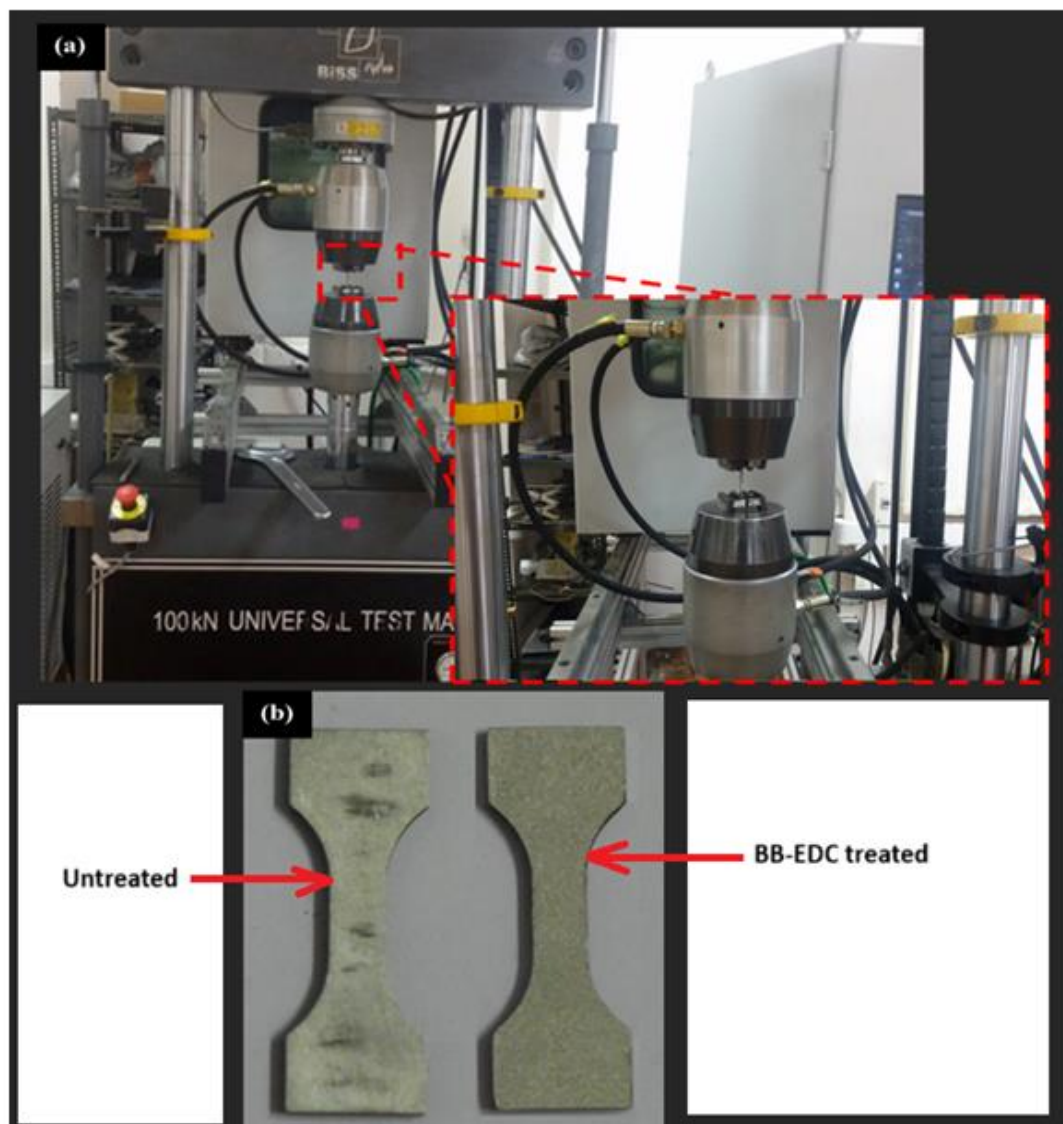


Figure 4.17 (a) Universal Testing Machine with a capacity of 100kN for high cycle fatigue testing (b) untreated and BB-EDC modified specimens

Fatigue testing performed with a Universal Testing Machine (UTM) of 100 kN, at 50 Hz frequency with 0.1 stress ratio, to process material with a cyclic (Tensile-Tensile) loading repeatedly until failure ensues. In this high cycle fatigue (HCF) testing, the UTM applies cyclic tensile loading to the specimen in a sinusoidal waveform, which oscillates between a maximum and minimum stress value (tension-tension loading), over a large number of cycles in the range of 10^4 to 10^7 cycles.

The collected data from the fatigue test, which includes stress-life (S-N) curves, undergoes analysis to ascertain critical parameters like fatigue limit. Specimen blanks for HCF were crafted from the sheet of β -Ti alloy.

Standard tensile “dog-bones” measuring 3 mm $\times$ 1.5 mm in load-bearing cross-section and 80 mm in total length, as per ASTM Standard E466-07, were used in all specimens used in this study for fatigue testing Figure 4.17 (b).

4.14 Corrosion Testing

The electrochemical test setup for corrosion testing is shown in Figure 4.18. Potentiodynamic polarization measurements are a widely used electrochemical technique employed to investigate the corrosion characteristics of materials. These measurements aid in assessing parameters such as corrosion rate, corrosion potential, and other electrochemical properties of materials under corrosive conditions, thereby providing insights into their corrosion resistance.

In this analysis, Corrosion Potential (E_{corr}) refers to the potential at which the anodic gets equal to cathodic currents, serving as an indicator of the material's susceptibility to corrosion in the specific environment. Conversely, Corrosion Current Density (I_{corr}) represents the current density at the corrosion potential, offering a measure of the corrosion rate. This testing is valuable for selecting materials with enhanced corrosion resistance for biomedical applications, as it enables the assessment of the protective efficacy of coatings on various materials.

Potentiodynamic polarization measurements provide quantitative data on corrosion rates and electrochemical properties.

They enable comparative analysis by allowing the evaluation of different materials or treatments under similar conditions. In summary, potentiodynamic polarization measurements are a valuable tool in corrosion science, offering detailed insights into

the electrochemical behaviour of materials in corrosive environments. Table 4.6 below provides chemical constituents of the SBF solution.

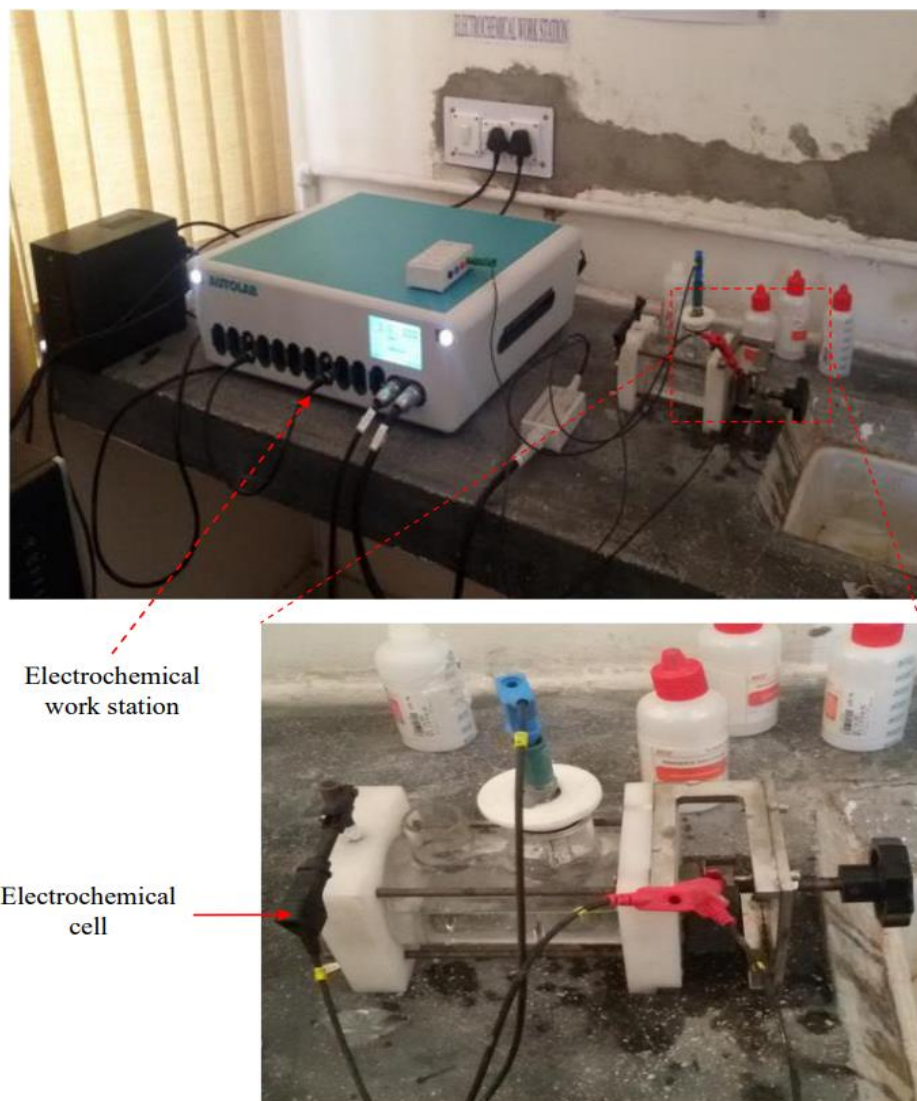


Figure 4.18 Photograph: Electrochemical test setup for corrosion testing

Table 4.6 Chemical constituents of SBF solution

Sr. No.	Reagents Used	Grams in 1000 ml
1	NaCl	8.03
2	NaHCO ₃	0.35
3	KCl	0.22
4	K ₂ HPO ₄ ·3H ₂ O	0.23
5	MgCl ₂ ·6H ₂ O	0.31

6	CaCl_2	0.29
7	Na_2SO_4	0.07
8	$((\text{HOCH}_2)_3\text{CNH}_2)$	6.11
9	1MHCL	1-4

4.15 Wear Testing

A frictional wear test was used for assessing the tribological performance of the untreated, EDM, and BB-EDC samples of the β -phase titanium alloy. This test was conducted under a 5 N load in room temperature ambient air, utilizing a pin-on-disc system of frictional wear-testing, (DUCOM, Instruments Pvt. Ltd., Bangalore, India), as shown in Figure 4.19.

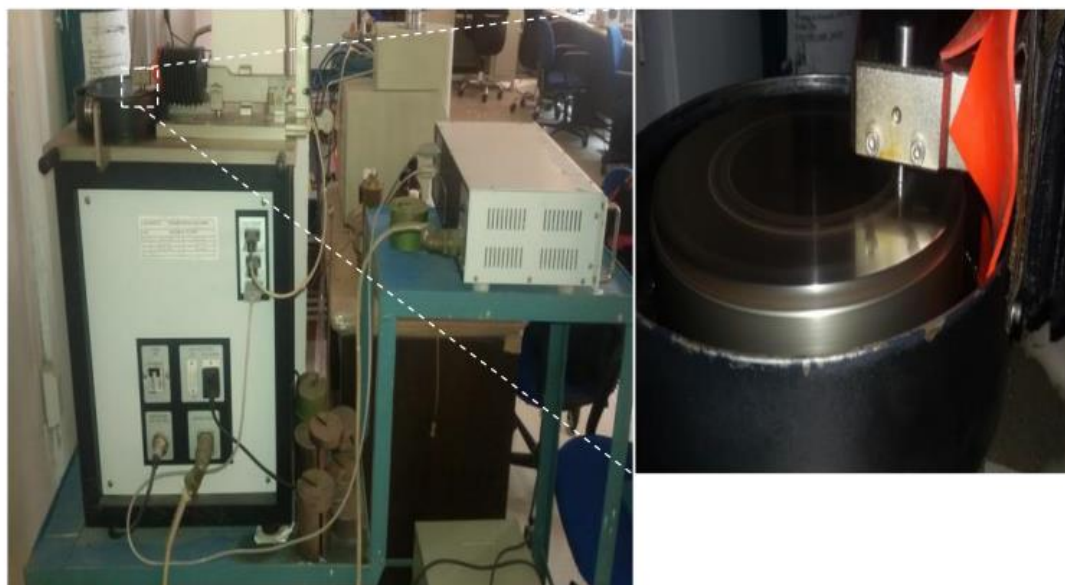


Figure 4.19 Photograph of the wear test apparatus

The wear experiments were conducted using a β -Ti alloy pin with a diameter of 5 mm and a length of 10 mm, combined with an EN-31 material based disc as the counterface. Each test involved a sliding distance of 500 m and a sliding velocity of 0.1 m/s.

4.16 In-Vitro Biocompatibility Assessment

In-vitro biocompatibility assessment refers to quality evaluation of a material interaction with biological systems under controlled laboratory conditions. This

assessment involves studying the material's compatibility with living cells, tissues, and organisms outside of their natural environment, typically within laboratory settings.

4.17 Evaluation of Cell Proliferation

Cell proliferation was assessed using DNA quantification and MTT assays. After MG-63 cells growing on tissues for 1, 3, and 7 days, the MTT test was carried out. After the allotted time, take out the culture media and fill the plate with 50 mL of MTT reagent (M2128; Thiazolyl Blue Tetrazolium Bromide, Sigma-Aldrich). Incubate for four hours at 37°C.

4.18 Evaluation of Cell Differentiation

Assessment of cellular differentiation entails examining the mechanism through which cells transition from one type to a more specialized form. This evaluation is essential for comprehending the progression and function of cells across diverse biological scenarios, including tissue generation, restoration, and reaction to external triggers.

4.19 Morphology of a cell

Pictures of the dried cultivated specimens after fixation are shown in Figure 4.20.

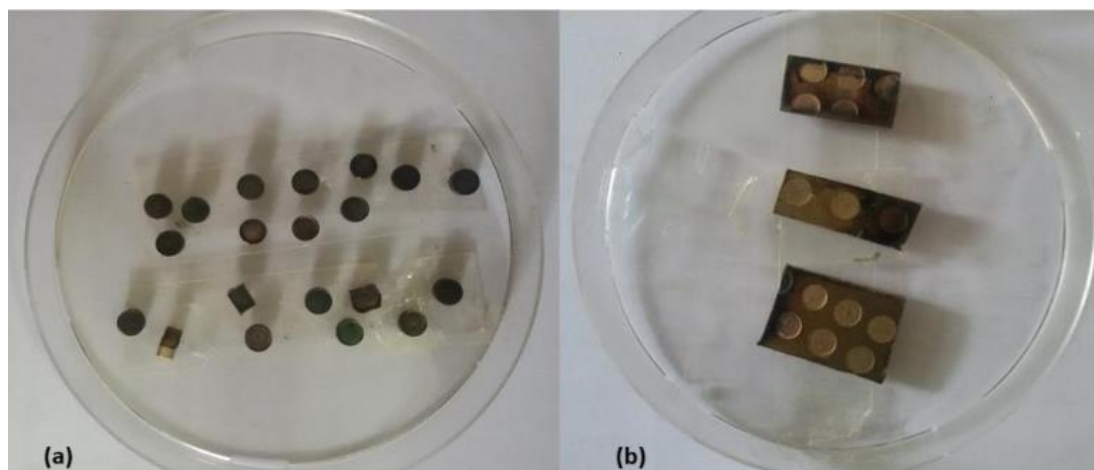


Figure 4.20 Picture of cell cultured specimens following fixation.

Cell morphology involves examining the physical structure and configuration of cells, encompassing aspects such as their dimensions, form, and internal arrangement. This examination aids in distinguishing diverse cell varieties, evaluating their well-being and investigating their reactions to different stimuli or intervention.

The cultured specimens were taken out of the incubator at predetermined intervals, and immobilized cells were prepared for 30 minutes in a 2.5% solution of glutaraldehyde. They were then rinsed three times with phosphate buffered saline solution and gradually dehydrated for ten minutes at a time in ethanol at different concentrations (30%, 50%, 70%, 90%, and 100%). The samples were then air-dried in a desiccator for a full day before gold-sputter coating for FE-SEM analysis.

4.20 Statistical Evaluation

The data are displayed as mean $\pm$ standard deviation (SD). Each experiment was replicated thrice to confirm the reliability of the findings. Statistical analysis was performed using the Taguchi method followed by ANOVA to identify significant differences between factors. A confidence level of 95% was maintained, and p-values less than 0.05 were considered statistically significant.

RESULTS AND DISCUSSION

This chapter explores the discussions pertaining to results and analysis in two main phases. The initial phase revolves around the examination of experimental work results and analysis, focusing on the surface attributes.

A thorough and stringent investigation of surface morphology, surface roughness, surface chemistry, surface microhardness, coating layer thickness, and surface cracks, was conducted in order to assess the effect. The impact of these surface characteristics on the changed surface's bioactivity, fatigue endurance, and corrosion resistance is discussed in the following phase.

5.1 Findings and Examination of β -Phase Titanium Alloy

In this study, a novel category of β -Ti alloy has been chosen as the material of focus due to its extensive utilization in the biomedical sector.

According to the level of input process parameters, subsequent to obtained the experimental data results from the mixed mode L-18 orthogonal array (OA), analysis was conducted utilizing the MINITAB-17 software.

5.1.1 Analysis of Surface Roughness

To conduct experiments, an appropriate design matrix of mixed L-18 orthogonal array was used, and surface roughness was measured for each corresponding parametric setting. The observed average values of mean and S/N ratios for surface roughness (SR) resulted out of experiments are shown in Table 5.1.

The impact of four input parameters: ("Type of Process," "Peak Current," "Pulse Duration," and "Supply Voltage.") were considered. For the "Type of Process" parameter, two levels were selected (EDM and BB-EDC), while the remaining input parameters were varied at three levels each (Peak Current: 5 amperes, 10 amperes, 15 amperes; Pulse Duration: 50 μ s, 100 μ s, 200 μ s; Supply Voltage: 60 volts, 90 volts, 120 volts). Three measurements were taken at different locations to determine the average

Ra value. Further, the Surface roughness (SR) was examined by reviewing the mean and S/N ratios plots.

Table 5.1 Design of experiment and observed surface roughness as per L-18 OA

Sr. No.	Type of Process	Peak Current (Ip)	Pulse Duration (Ton)	Supply Voltage (V)	Surface Roughness (μm)	
					Mean	S/N Ratio
1	EDM	5	50	60	1.19	-1.51
2	EDM	5	100	90	1.99	-5.97
3	EDM	5	200	120	2.96	-9.43
4	EDM	10	50	60	1.61	-4.15
5	EDM	10	100	90	2.81	-8.98
6	EDM	10	200	120	3.72	-11.42
7	EDM	15	50	90	2.44	-7.74
8	EDM	15	100	120	3.67	-11.29
9	EDM	15	200	60	3.17	-10.01
10	BB-EDC	5	50	120	0.79	2.05
11	BB-EDC	5	100	60	0.66	3.64
12	BB-EDC	5	200	90	1.01	-0.11
13	BB-EDC	10	50	90	0.94	0.52
14	BB-EDC	10	100	120	1.17	-1.36
15	BB-EDC	10	200	60	1.37	2.77
16	BB-EDC	15	50	120	1.09	0.78
17	BB-EDC	15	100	60	1.16	1.32
18	BB-EDC	15	200	90	1.7392	4.81

The analysis employed the "smaller-the-better" criterion to assess the effect of input parameters on SR.

Figure 5.1 (a) and (c) illustrate the mean and S/N ratio plots for the surface roughness of β -phase titanium alloy following EDM and BB-EDC treatments. Both plots show similar trends, validating the accuracy of the observations. The figures specify that the average surface roughness on the EDM surface is notably high, around 2.61 μm . EDM's high discharge energy penetrates the workpiece surface deeply, removing material as molten debris and forming micro-sized craters (Prakash et al., 2015; Prakash, Kansal, Pabla, & Puri, 2016b).

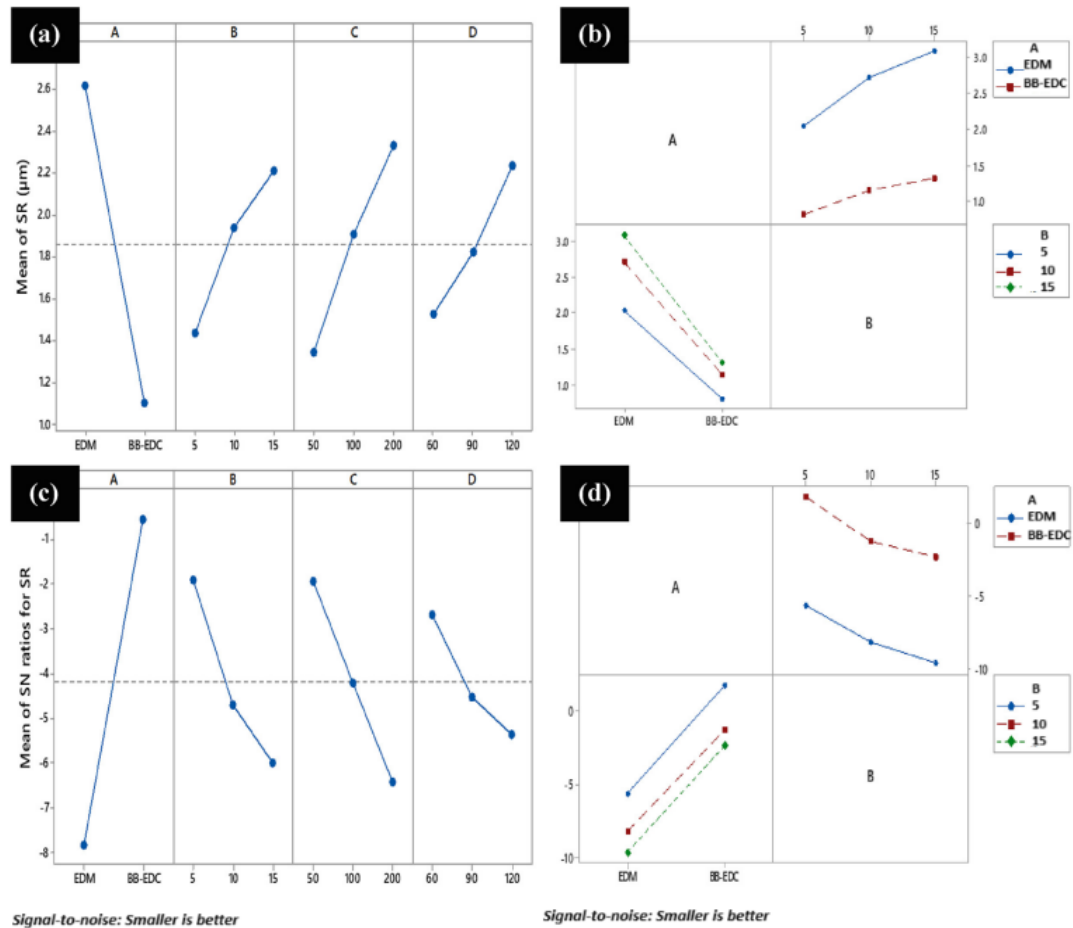


Figure 5.1 Mean and S/N ratio plot of surface roughness in BB-EDC process

From the Figure 5.1 (a) and (c), it is evident that surface roughness increases with higher peak current. As peak current rises, the thermal energy also increases, melting more workpiece material and resulting in the formation of deep & wide craters, which leads to higher SR. Also, a bigger peak current leads to a thicker recast layer with more

surface irregularities. Figure 5.1 (b) and (d) show the interaction plot between peak current and the type of process. It is found that surface roughness increases with any peak current value in both processes, but it is having significantly smaller value in the BB-EDC process. Similarly, an increase in surface roughness is noted with longer pulse durations due to the higher thermal energy generated, which increases surface roughness. Surface roughness also rises with increased supply voltage because the discharge energy power increases, generating more thermal energy in the machining zone. This enlarges the crater radius and removes more material from the workpiece surface.

Additionally, the unflushed molten debris redeposited as a recast layer, forming high ridges that increased the SR value of the material. Conversely, the SR significantly decreased when the sample machining was done using the BB-EDC. After the BB-EDC, the average SR was found approximately $1.10\text{ }\mu\text{m}$, less than half of the roughness on the EDM surface ($2.61\text{ }\mu\text{m}$). This improvement is due to the BB-EDC process, where hard ceramic balls (ZrO_2) apply compressive force on the machined surface, removing surface irregularities and cracks while compressing the high ridges of redeposited material. Consequently, the resulting surface is flat and free from surface defects. A similar observation was there in previous research studies (Lin et al., 2001; Yan et al., 2000, 2002). The mechanism for flattening the machined surface is illustrated in Figure 5.2.

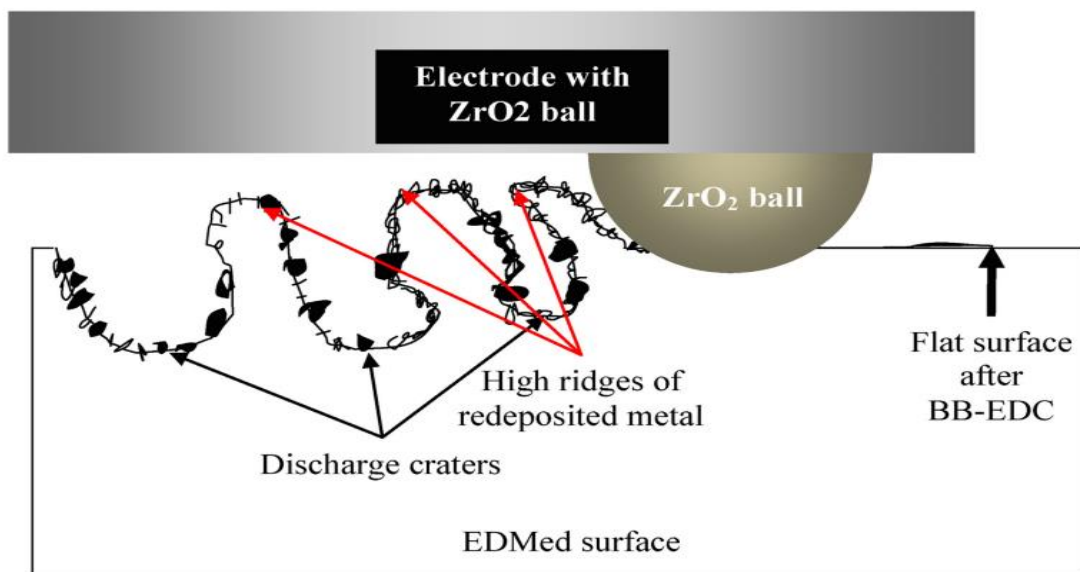


Figure 5.2 Diagram of the surface integrity enhancement mechanism in the BB-EDC process.

ANOVA was completed to fetch the effects of input parameters on surface roughness with statistical significance. Table 5.2 presents the ANOVA results for surface roughness, indicating that all process parameters significantly contribute to surface roughness. The type of process contributes the most (59.28%), followed by peak current (10.76%), pulse duration (16.89%), supply voltage (8.73%), and residual error have (4.37%) contribution in affecting the surface roughness. The contribution of all chosen process parameters is significant at a confidence level of 95%, indicated by their respective P-values < 0.05 .

Table 5.2 Surface Roughness Analysis (ANOVA)

Source	DF	Seq-SS	Adj-SS	Adj-MS	F	P	% Contribution
A	1	10.298	10.298	10.298	135.78	0.000	59.28
B	2	1.869	1.869	0.934	12.33	0.002	10.76
C	2	2.934	2.934	1.467	19.35	0.000	16.89
D	2	1.511	1.511	0.755	9.96	0.004	8.70
Residual	10	0.758	0.758	0.075			4.37
Total	17	17.373					

From Table 5.3 Surface roughness response table, it is evident that the type of process holds the highest significance among all the process parameters.

Table 5.3 Surface roughness response table.

Level	A	B	C	D
1	2.618	1.433	1.344	1.528
2	1.105	1.939	1.91	1.822
3	–	2.211	2.33	2.234
Delta	1.513	0.778	0.986	0.707
Rank	1	3	2	4

The optimal conditions for the Surface roughness response are presented in Table 5.4. In the subsequent section of this chapter, the surface hardness was evaluated at the optimized settings.

Table 5.4 The optimal condition for each output response characteristic with confirmation testing

Response (units)	Input process parameters				Predicted	Measured	Percentage error (%)
	A	B	C	D			
Surface roughness (μm)	BB- EDC	5	50	60	0.55	0.57	3.5

5.2 Impact of EDM and BB-EDC Process Parameters on Surface Roughness

The figure provides the effect of EDM and BB-EDC process parameters on SR. Figure 5.3 (a) clarifies the impact of peak current on surface roughness, showing that both processes exhibit similar characteristics. The plot indicates that the average SR increases with the rise in peak current (from 5 to 15 A). At a high peak current of 15 A, the surface roughness for the EDM and BB-EDC processes was approximately 2.6 μm and 1.3 μm , respectively.

The SR was found higher with the EDM process. Conversely, the BB-EDC process achieved a minimum surface roughness of approximately 0.57 μm . This improvement is due to the BB-EDC process, where hard ceramic balls (ZrO_2) exert compressive force on the machined surface, eliminating surface irregularities and cracks, and compressing the high ridges of redeposited material. Consequently, the resulting surface was smooth and free of defects. A similar observation was reported in previous research studies (Lin et al., 2001; Yan et al., 2000, 2002).

Figure 5.3 (b) provides the effect of pulse duration on SR.

The plot reveals that both processes exhibit similar characteristics, with the average surface roughness increasing as pulse duration extends from 50 to 200 μs . This is because longer pulse durations generate more thermal energy between the tool and workpiece, melting more material and forming deeper and wider craters, which leads to higher surface roughness. Additionally, since pulse duration is directly related to thermal energy, more heat is generated, further increasing surface roughness. The BB-EDC process, however, resulted in lower surface roughness.

Figure 5.3 (c) clearly showing the effect of supply voltage on SR. The plot reveals that both processes exhibit similar trends, with the average surface roughness rising as the supply voltage increases from 60 to 120 V. Additionally, surface roughness increases with higher supply voltage.

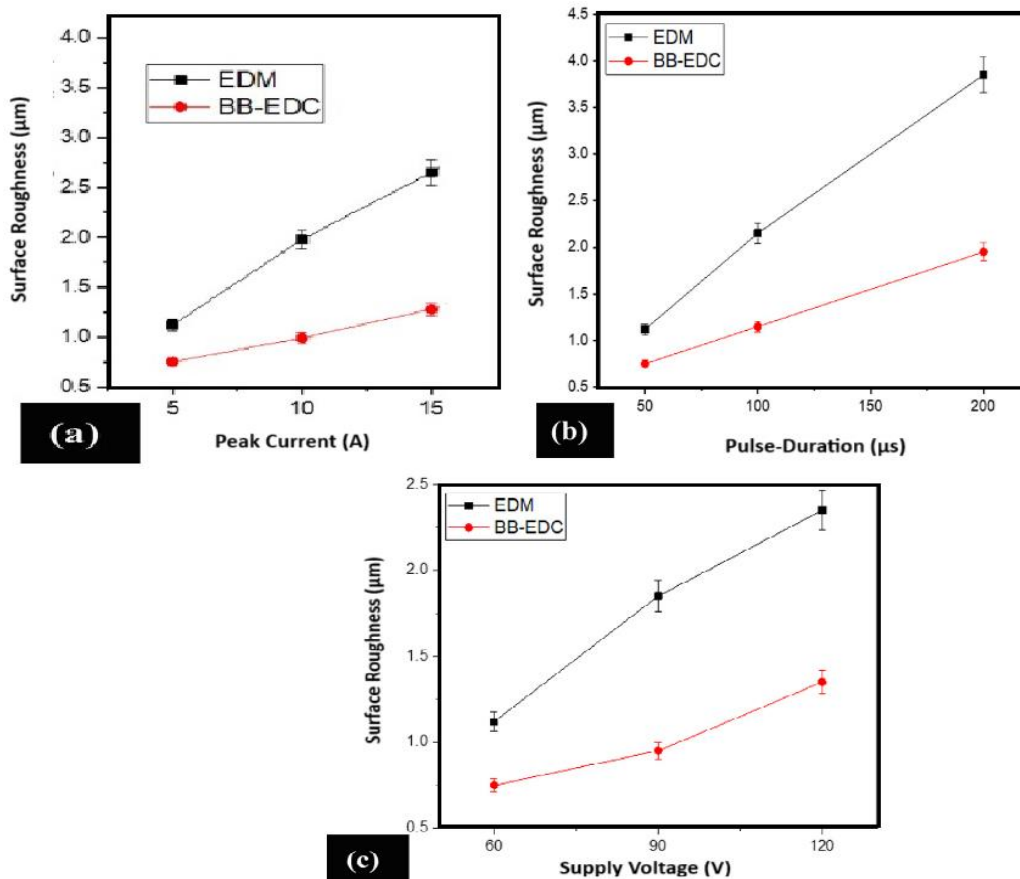


Figure 5.3 Effect of EDM and BB-EDC process parameters on SR: (a) peak current, (b) pulse duration, and (c) supply voltage

This occurs because an increase in supply voltage boosts the discharge energy power, consequently elevating the thermal energy generated within the machining zone. This, in turn, enlarges the crater radius and removes more material from the workpiece surface.

5.3 Impact of EDM and BB-EDC Process Parameters on Micro Hardness

Figure 5.4 reveals the impact of EDM and BB-EDC process parameters on the Surface

hardness of β -Ti alloy specimens. Figure 5.4 (a) shows the effect of peak current on surface micro-hardness. The plot indicates that the average micro-hardness initially increases from 5 to 15 A for both processes. At a peak current of 15 A, the microhardness is higher for the BB-EDC process (848 HV) compared to the EDM surface (545 HV). In the EDM process, surface hardness increases due to the escalation in peak current, which augments the thermal energy produced by the electrical spark, resulting a deeper heat-affected zone. Concurrently, the presence of distilled water induces sudden quenching, resulting in the recrystallization of the grain structure and the development of finer grains in the heat-affected zone, thereby enhancing the mechanical properties of the surface. Conversely, in the BB-EDC process, hard ceramic balls (ZrO_2) apply compressive force to the machined surface, facilitating recrystallization and strain-hardening, which further improves the mechanical properties (Uddin et al., 2021).

Figure 5.4 (b) displays the impact of pulse duration on surface micro-hardness. The average micro-hardness initially increases from 50 to 100 μs for both processes, followed by a decrease with further increases in pulse duration. At a pulse duration of 100 μs , the microhardness is higher for the BB-EDC compared to the EDM. In case of EDM-only, a pulse duration of 100 μs provides sufficient and optimal thermal energy between the tool and workpiece.

This results in a narrower and shallower heat-affected zone, promoting grain recrystallization in a confined area, thus enhancing surface properties. However, if the pulse duration exceeds 100 μs , the prolonged thermal energy exposure allows for deeper penetration inside workpiece surface, resulting wider and deeper heat-affected zone and subsequently reducing surface hardness.

Conversely, the BB-EDC process achieves notably high surface hardness comparison to EDM.

Figure 5.4 (c) depicts the influence of supply voltage on surface micro-hardness. The plot indicates a decrease in the average micro-hardness for both processes with increasing supply voltage. This decline can be attributed to the heightened discharge power accompanying the rise in supply voltage. Consequently, the thermal energy generated within the machining zone increases, leading to an expansion in the depth and radius of the heat-affected zone, thereby reducing surface hardness.

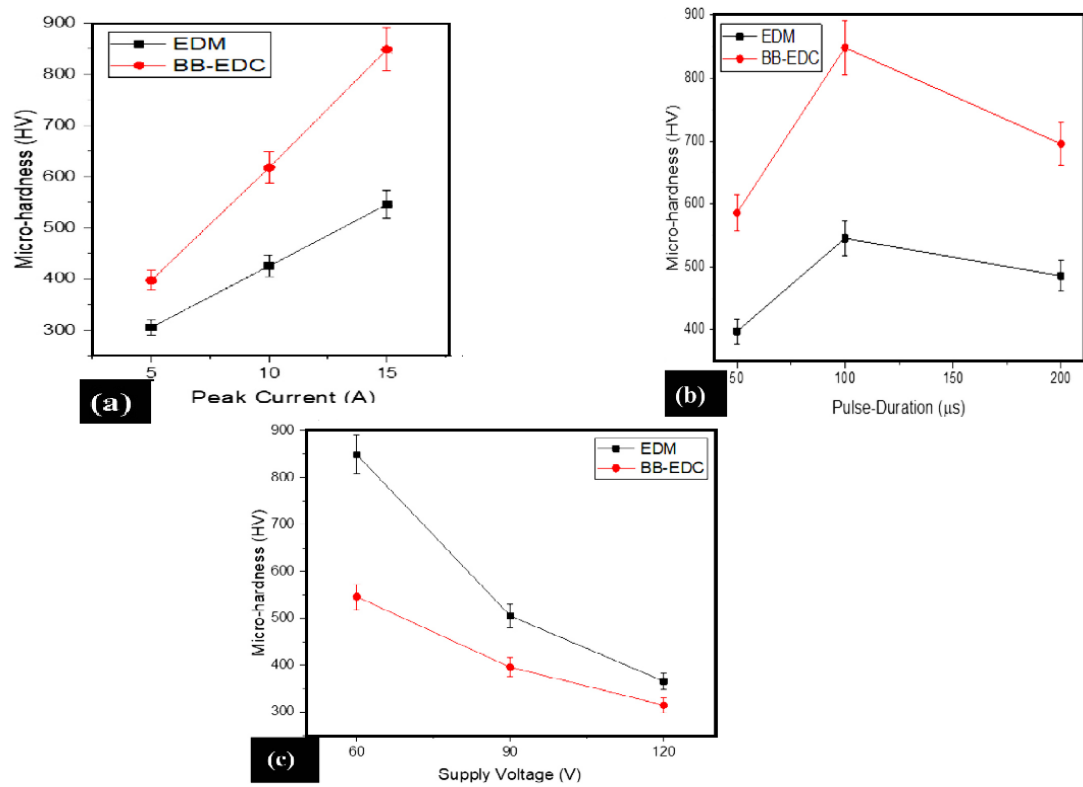


Figure 5.4 Effect of EDM and BB-EDC process parameters on surface micro-hardness: (a) peak current, (b) pulse duration, and (c) supply voltage

5.4 Surface Morphology and Elemental Composition

The surface morphology and elemental composition were analyzed using FE-SEM and EDS techniques. Figure 5.5 shows the FE-SEM images and EDS spectrum for EDM and BB-EDC treated surfaces. The images reveal that the EDM surface exhibits a high degree of craters and elevated ridges of redeposited molten metal, as seen in Figure 5.5 (a). These surface defects degrade surface integrity and reduce the lifespan of implants in the host location. It has been reported that the longevity of implants depends on their fatigue and corrosion performance, which, in turn, is influenced by surface roughness (Lin et al., 2001; Yan et al., 2000).

The EDM surface contained oxygen (O) along with other base-material elements, indicating that no additional contamination occurred during the process, as shown in Figure 5.5 (b).

Conversely, the BB-EDC treated surface exhibited a flat surface with few craters, as depicted in Figure 5.5 (c). The resulting surface after the BB-EDC process exhibited high strength and hardness, further improving the fatigue performance of the implant.

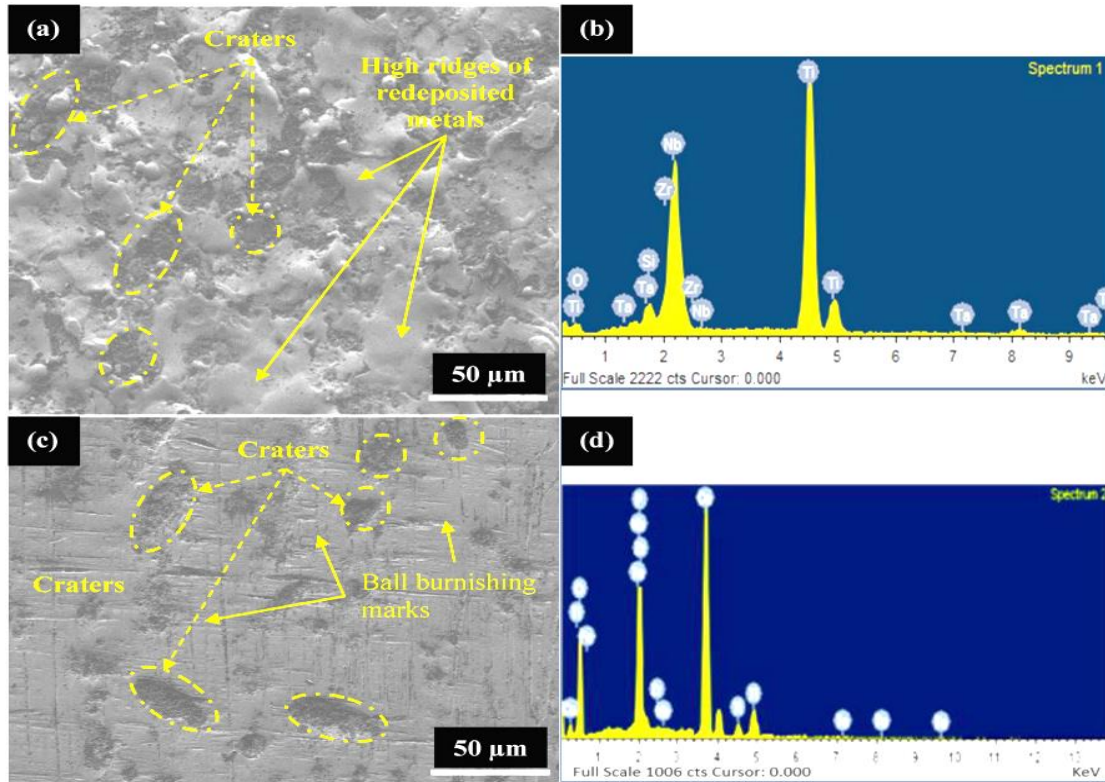


Figure 5.5 FE-SEM and EDS Spectrum: (a-b) EDM surface, (c-d) BB-EDC treated surface.

5.5 FE-SEM and Cross-Section Morphology of EDM and BB-EDC Treated Surface

The top and cross-sectional surface morphology were analyzed using the FE-SEM technique. Figure 5.6 displays the top and cross-sectional views of surfaces treated with EDM and BB-EDC. The images show that the EDM surface has a high degree of craters and prominent ridges of redeposited molten metal, as illustrated in Figure 5.6 (a). These surface defects compromise surface integrity and reduce the lifespan of implants at the host location. It has been reported that the lifespan of implants depends on their fatigue and corrosion performance, which in turn is influenced by surface roughness (Lin et al., 2001; Prakash et al., 2015; Prakash, Kansal, Pabla, & Puri, 2016b; Yan et al., 2000, 2002).

Figure 5.6 (b) showing the cross-section of our EDM modified surface, where a recast layer with a thickness of 25–30 μm was measured. The surface displays irregularities due to high ridges.

In contrast, the BB-EDC treated surface features a flat surface with few craters, as given in Figure 5.6 (c). The high ridges of redeposited metal have been flattened by the hard ZrO_2 balls, which exert a high compressive force on the workpiece. This rearranges the material on the machined surface.

Figure 5.6 (d) shows the cross-section of the BB-EDC modified surface, where the recast layer thickness ranges from 10 to 15 μm . The cross-section reveals a flat and smooth surface, free from irregularities. The surface produced by the BB-EDC process exhibits better strength.

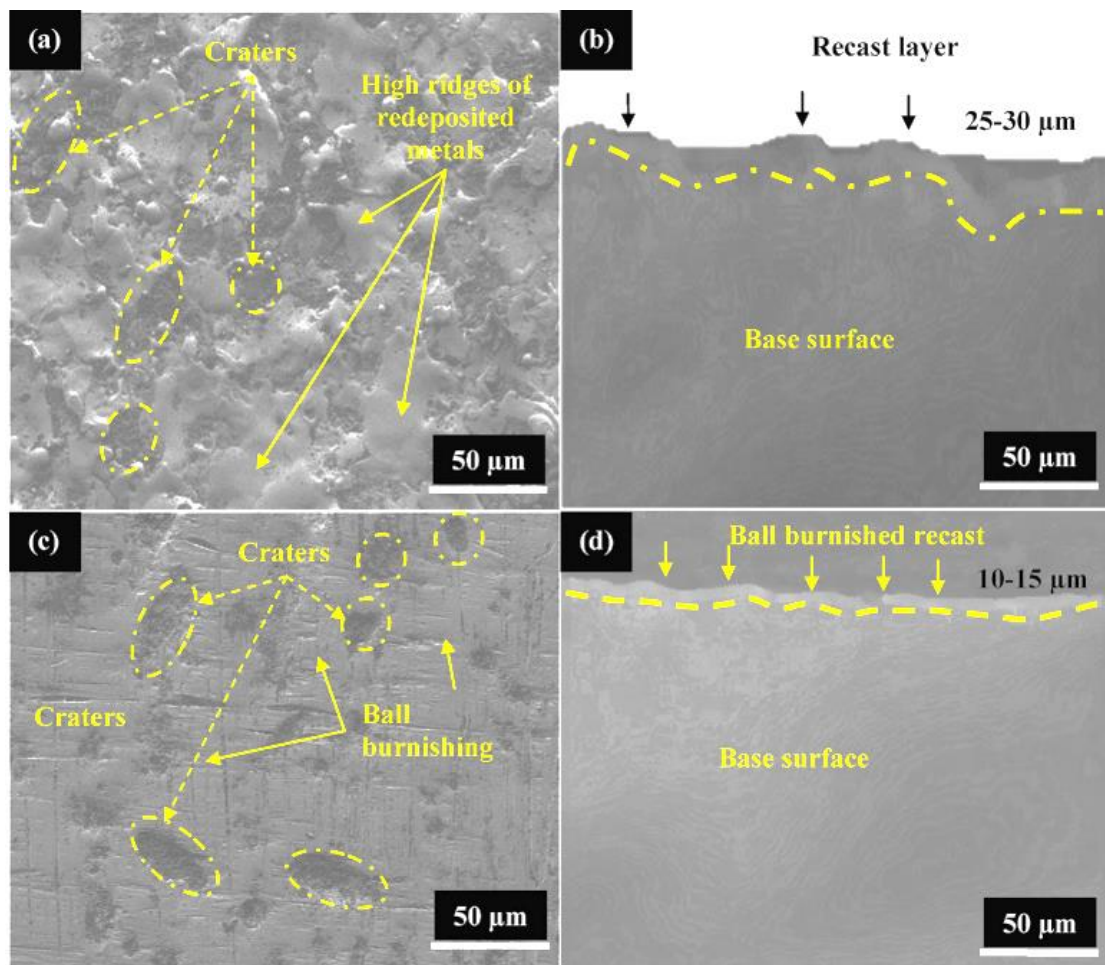


Figure 5.6 FE-SEM and cross-section morphology of: (a-b) EDM surface, (c-d) BB-EDC treated surface

5.6 Influence of Peak Current on the Surface Roughness, Cladded Thickness and Hardness

Another aspect of the experimental investigation involves understanding the characteristics of the synthesized TiO_2 loaded HAP-cladded layer using the EDC and BB-EDC processes, with a focus on parameters such as roughness, thickness, and hardness under the influence of peak current. Figure 5.7 (A, B, C) illustrates the impact of significant peak current on the average SR and depth of the HAP layer, as given in Figure 5.7 (A and B).

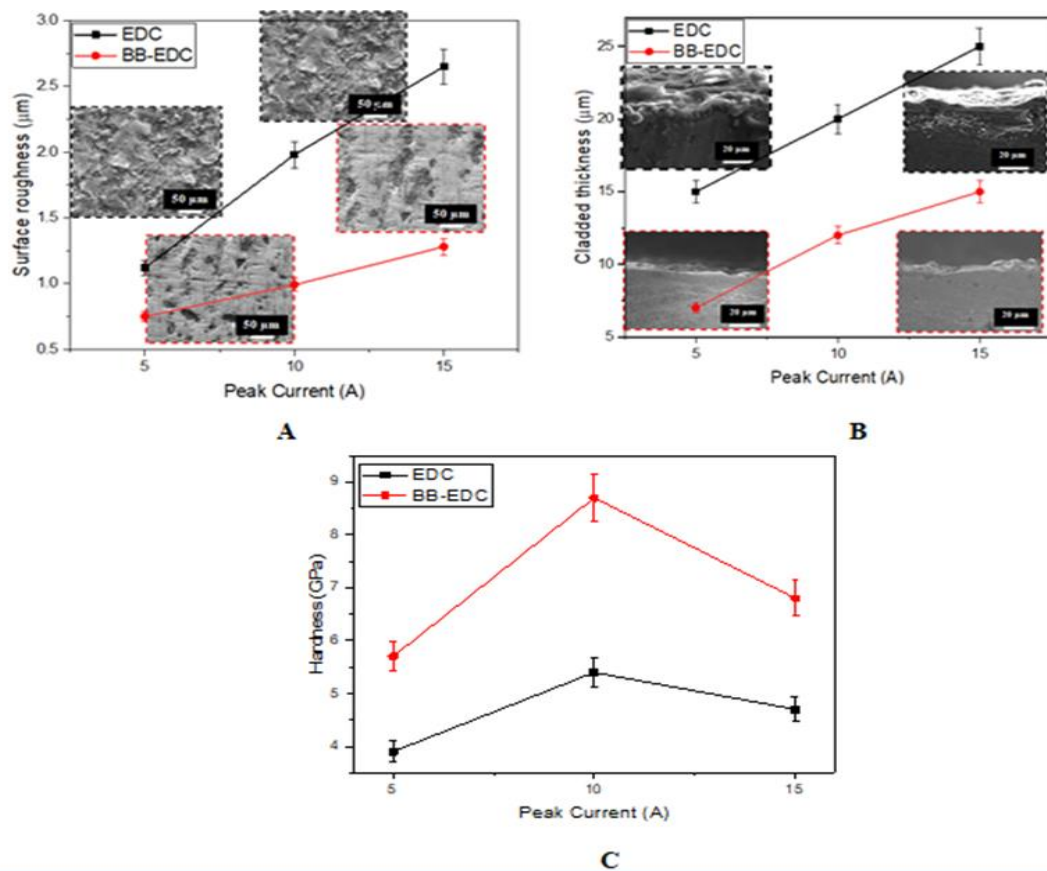


Figure 5.7 Influence of peak current on the surface roughness of HAP-layer clad through EDC and BB-EDC processes.

It is clear that, both roughness and layer thickness attained higher average values with rise in peak current in each of the EDC and BB-EDC process. The substantial thermal energy generated in the processing zone impacts the underlying surface, causing material removal and the occurrence of large craters or pits on the surface. However, SR and HAP-cladded layer thickness results out of BB-EDC process is lower compared

to the EDC. From the plot, it can be observed that the average micro-hardness in both processes initially increases from 5 to 10 A and then starts to decrease as the peak current increases further.

At 10 A peak current, the microhardness in the BB-EDC process (8.6 GPa) is higher compared to the EDM surface (5.4 GPa). In the EDM process, the surface hardness increases because the higher peak current generates more thermal energy, resulting in a deeper heat-affected zone.

Additionally, the presence of distilled water causes rapid quenching, which recrystallizes and develops a fine grain structure in the heat-affected zone, thereby improving the mechanical properties of the surface. In contrast, in the BB-EDC process, hard ceramic balls (ZrO_2) apply compressive force to the machined surface, leading to recrystallization and strain hardening, which enhance the mechanical properties.

5.7 Cross-Section Morphology and EDS Spectrum of HAP-Layer at a Peak Current of 10 A

Figure 5.8 illustrates the determination of both the thickness and elemental composition of the HAP-clad functionalized layer, after processing through the EDC and BB-EDC techniques.

Figure 5.8 (a) presents the thickness of the Hap-clad layer obtained via the EDC procedure, measuring approximately 20-25 μm . Figure 5.8 (b) demonstrates the uniformity of the Hap-clad layer across the cross-section, as well as the EDX spectrum after obtaining EDC-modified surface, revealing peaks corresponding to calcium (Ca), phosphorus (P), and oxygen (O), alongside the elemental constituents of the base materials (Ti, Nb, Ta, Zr). Figure 5.8 (c) displays the depth of the synthesized layer achieved through the BB EDC method, ranging from 5-10 μm . The BB-EDC process effectively eliminated irregularities, resulting in a flat and dense surface after the application of the Hap-clad layer. Figure 5.8 (d) presents the EDX spectrum of the BB-EDC technique, showcasing associated peaks of calcium (Ca), phosphorus (P), and oxygen (O), as well as the base materials (Ti, Nb, Ta, Zr), thus confirming the uniform cladding of HAP onto the enhanced layer.

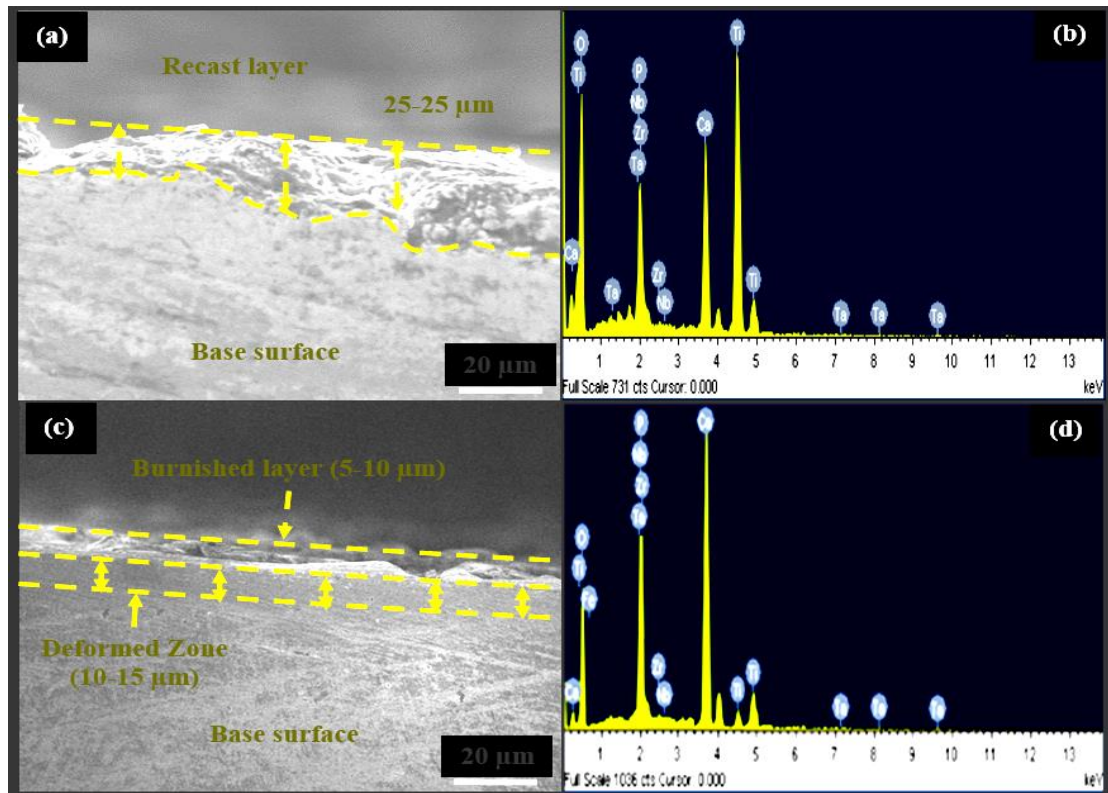


Figure 5.8 The cross-sectional shape and EDS spectra of the HAP-layer clad at a peak current of 10A utilizing the (a-b) EDC and (c-d) BB-EDC procedures

5.8 Functionalized HAP-Clad Layer on β -phase Ti-Alloy at a Peak Current of 10A

In another experimental investigation, as depicted in Figure 5.9, the EDC process demonstrates superior outcomes compared to the EDM-modified surface, with a relatively lower ridge of re-deposited metal observed.

Figure 5.9 (a) shows the induced surface characteristics after EDC, due to electrical discharges, such as the discharge craters (yellow dotted circles), globules (black solid arrows), and ridges of redeposited metal (red solid arrows). Figure 5.9 (b) clearly shows the inhibition of deep and wide crater formation after EDC. due to the presence of the Hap-clad layer and the re-deposited material under high magnification (500 $\times$). Figure 5.9 (c) showcases a plane surface clad with HAp obtained in BB-EDC-treated samples. Lastly, Figure 5.9 (d) exhibits a comparatively smoother surface after BB-EDC, along with some burnishing marks, and confirms the filling of earlier observed craters with

the functionalized Hap-clad layer, as validated under high magnification (500×) microscopy.

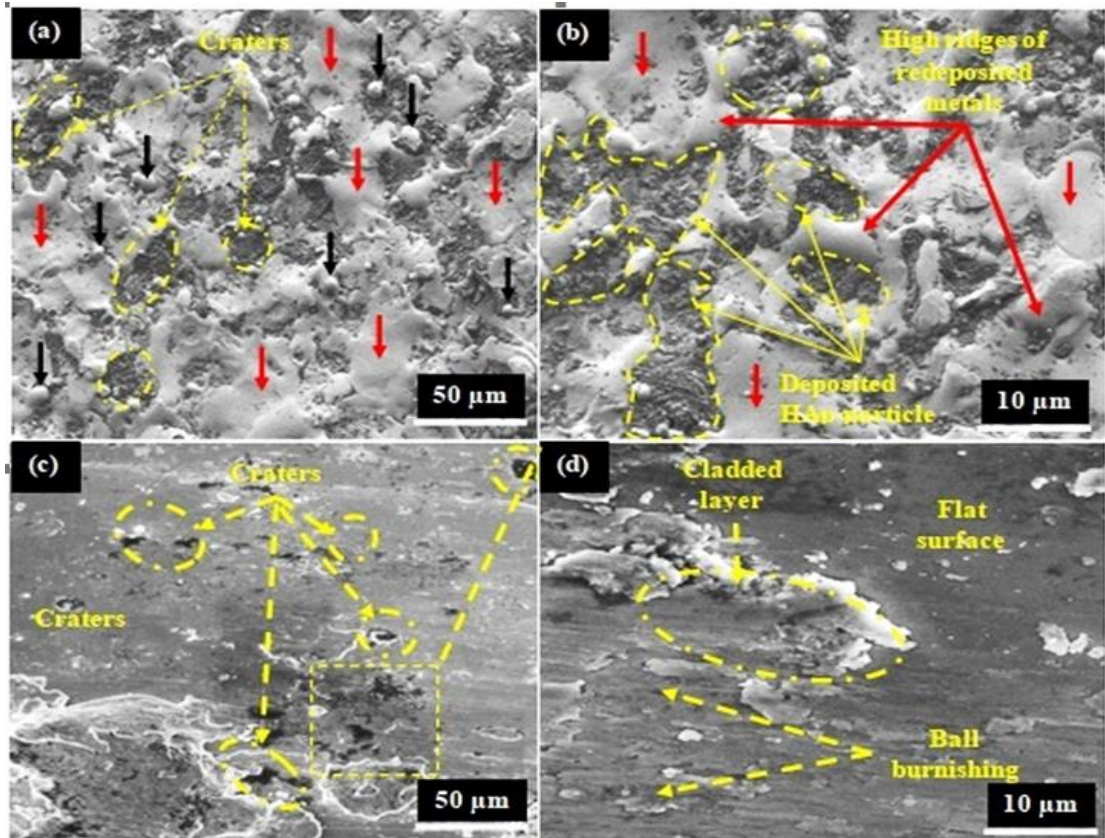


Figure 5.9 Top-surface morphology of EDC (a-b) and BB-EDC (c-d) modifications at a peak current of 10A

5.9 Porosity analysis

Porosity on the BB-EDC improves the mechanical anchorage between the bone and the implant, which promotes better blood circulation, ensuring proper nutrient delivery to cells, which further enhances osseointegration.

Figure 5.10 (a), the morphological characteristics of the material, as revealed by FE-SEM analysis, demonstrate that the powder particles used are extremely small and microscopic, in the range from 0.110 to 0.200 μm , while maintaining high sterility. The cladded layer is comprised of a blended mixture of TiO_2 (30 wt%) and HAP (70 wt%). The slurry needed for cladding is prepared by mixing the blended mixture of TiO_2 and HAP with distilled water at a concentration of 30 g/l. Furthermore, as per Figure 5.10 (b), the existence of calcium (Ca), phosphorus (P), and oxygen (O) elements, alongside

other base material elements, serves as confirmation of the reinforcement of hydroxyapatite (HAP) throughout the process.

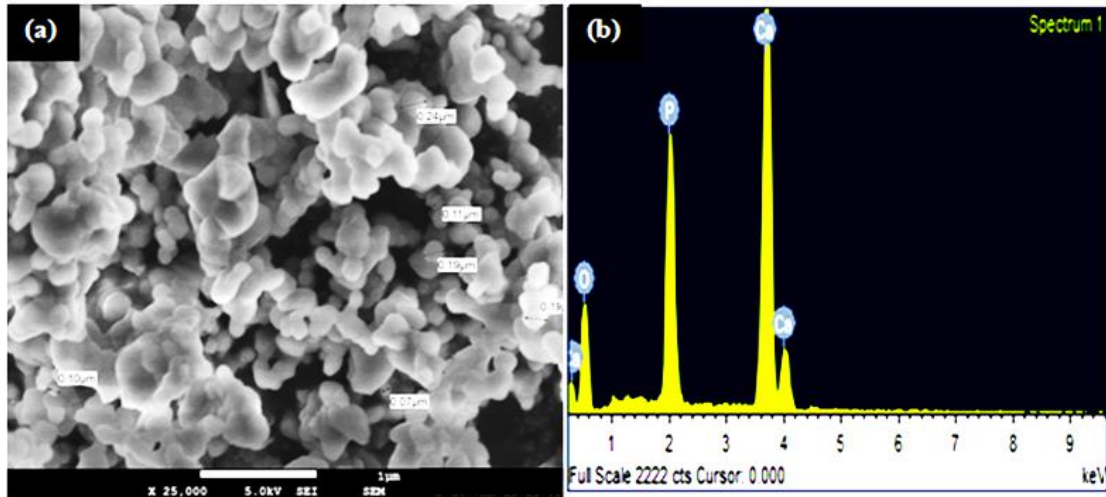


Figure 5.10 FE-SEM morphology with EDS spectrum of HAP powder particles, focusing on (a) their shape and size, and (b) their elemental composition

5.10 Corrosion resistance

Figure 5.11 provides the Tafel polarization curve for the untreated, EDC, and BB-EDC based samples.

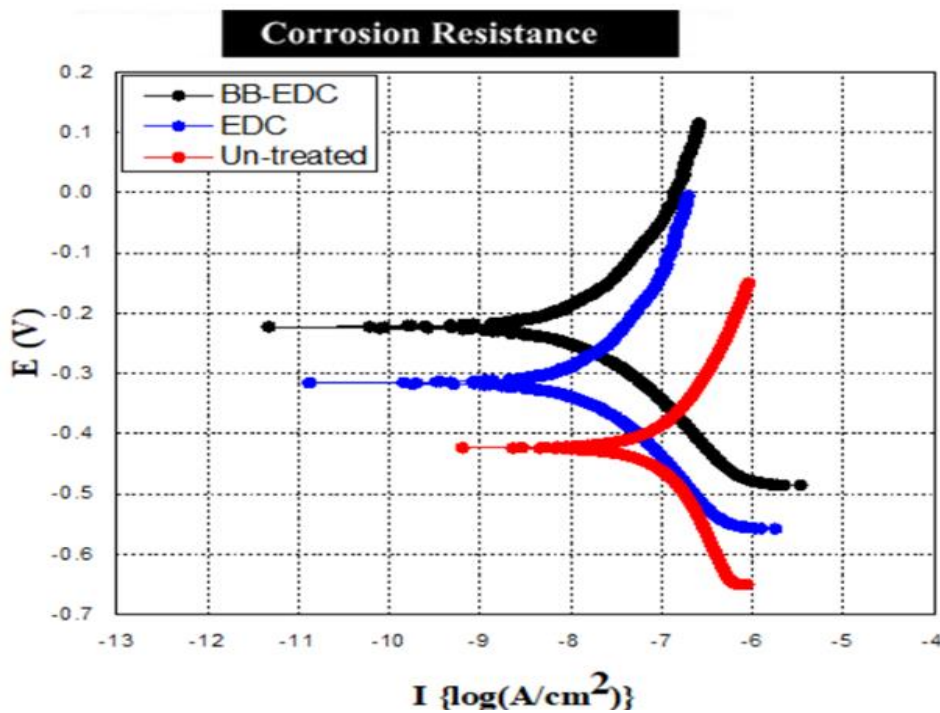


Figure 5.11 Corrosion resistance performance on un-treated, EDC and BB-EDC

The Tafel polarization curve reveals that an untreated sample of β -TNTZ alloy shows opposite results, with a high I_{cor} and a low E_{cor} , indicating poor corrosion resistance. This is due to the presence of a very thin layer of TiO_2 on the surface of the bare β -Ti alloy, resulting in low surface hardness. Consequently, untreated specimen fail in corrosive environments and promote pitting corrosion.

The Tafel polarization curve of BB-EDC modified specimens shifted to the left, indicating a lower I_{corr} ($-8.18 \mu\text{A}/\text{cm}^2$) and a lower E_{corr} (-222 mV), compared to untreated specimens.

The enhanced corrosion resistance of BB-EDC specimens is attributed to the formation of a homogeneously functional TiO_2 -loaded HAP-cladded layer and a fine-grain microstructure near the top surface, which acts as a barrier to corrosion and promotes apatite growth.

5.11 Wear Resistance

The wear rate after EDC decreases due to the loading of TiO_2 in to the HAP, which tends to enhance the properties, like high hardness and adhesion. Ball burnishing based BB-EDC, further enhances these properties by improving the surface roughness and hardness. As in Figure 5.12, compared to the untreated surface, which had a volume removal rate of 2.8 mm^3 , the volume decreased to 0.61 mm^3 after EDM treatment and further reduction was observed up to 0.089 mm^3 , after BB-EDC.

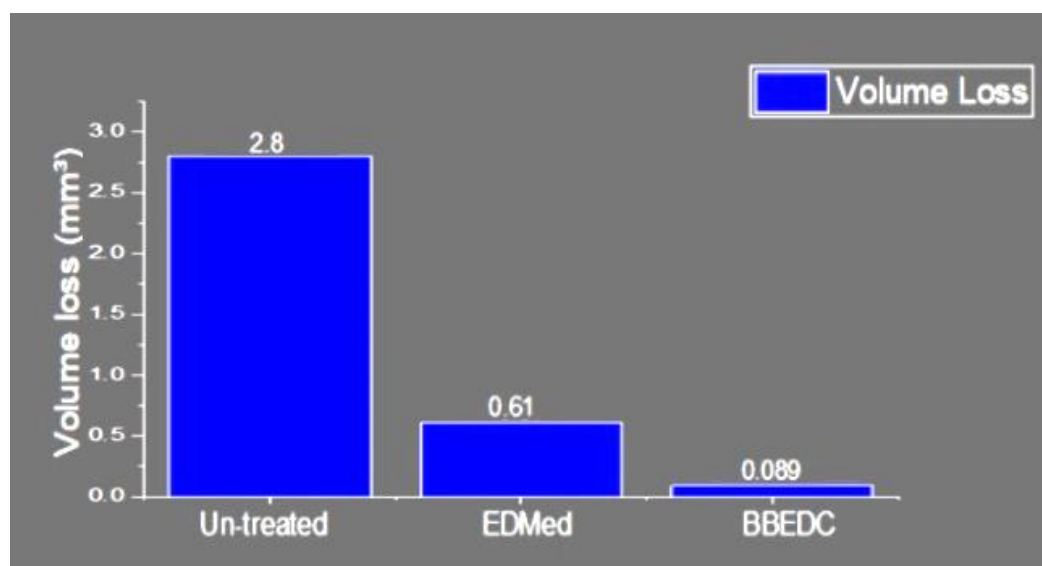


Figure 5.12 Volume loss shown by untreated, EDM and BB-EDC samples of β -Ti alloy at 5N load

In Figure 5.13, the wear rate reduces from 0.00003 mm³/N m in the untreated sample surface to 0.000025 mm³/N.m after EDM treatment, and further drops to 0.000011 mm³/N.m after BB-EDC. Similarly, the coefficient of friction (C.O.F), which was found 0.5 in case of the untreated, reduced to 0.39 after the EDM and further reduced to 0.35 after BB-EDC. So, both EDM and BB-EDC reduces wear rate and the results are improved more drastically in BB-EDC.

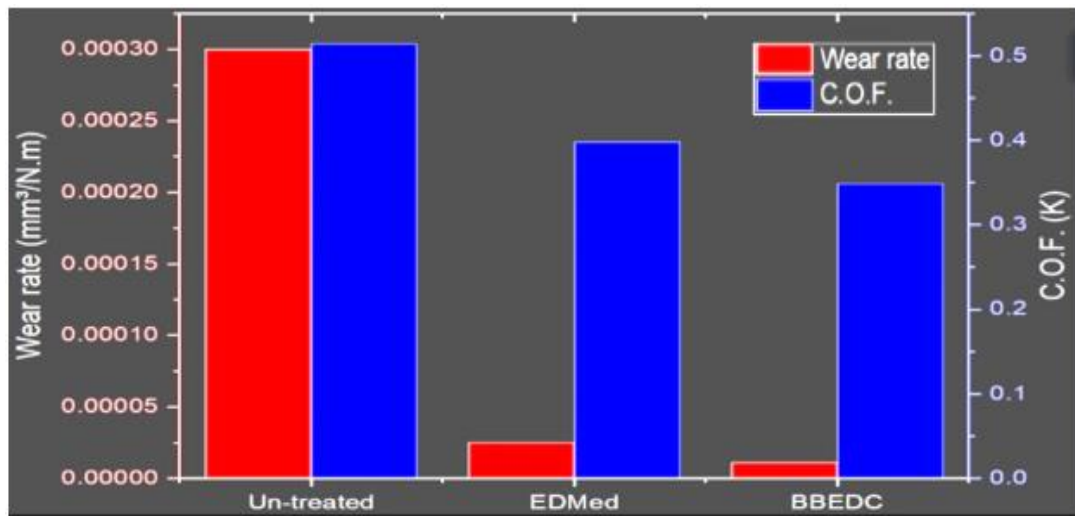


Figure 5.13 Wear rate and C.O.F. of untreated, EDM and BB-EDC specimens of β -Ti alloy at 5N load

5.11.1 Fractography of Worn Surface

The Fractography of the various worn surface of untreated, EDM-treated and BB-EDC-treated β -Ti alloy has been carried out using FE-SEM. Resulting grooves and plastically deformed or smeared surface, can give idea about volume loss.

For untreated surface as in Figure 5.14, the wear occurred mostly due to plastic deformation or ploughing, while sliding against EN-31 based hardened counter surface disc. The observed big and wide grooves across the worn surface of untreated specimen, leads to serious damage of untreated specimen.

Similarly, as shown in the Figure 5.15, behaviour of EDM treated worn sample illustrated that the volume loss was mainly the cause of brittle fracture, that results in to craters, plastics deformations and ploughing marks across the surface.

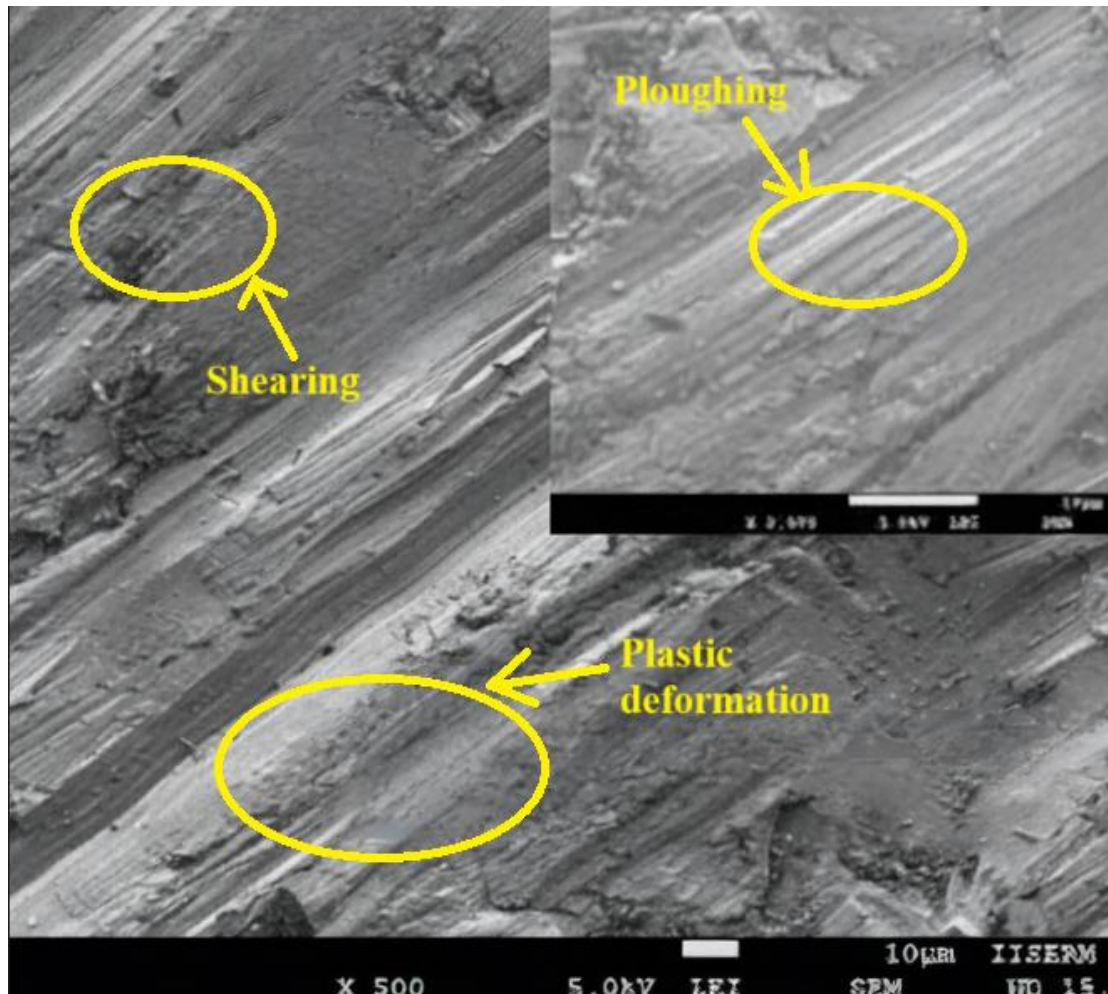


Figure 5.14 Fractography of worn surface of untreated β -Ti implant after wear test

Also as per Figure 5.16, the BB-EDC surface due to reinforcement of TiO₂ with HAP is almost free from defects and only ploughing marks are observed due to sliding effect, there is no such mark related to plastic deformation, thus describing BB-EDC as the optimum surface, after the wear test.

5.12 FATIGUE STRENGTH

The S-N curve in Figure 5.17 compares the fatigue strength of untreated, EDC, and BB-EDC-treated β -Ti alloy. The curve indicates that the fatigue performance of the β -Ti alloy is highly improved by the BB-EDC treatment. The surface hardness of the untreated β -Ti alloy was 285 Hv, which led to crack initiation and failure under high cyclic fatigue loading, with a fatigue limit of 385 MPa. In contrast, BB-EDC modified specimens endured a higher number of cycles before failure under stress amplitudes

between 515 and 580 MPa, with a fatigue limit of approximately 490 MPa, surpassing the untreated specimens.

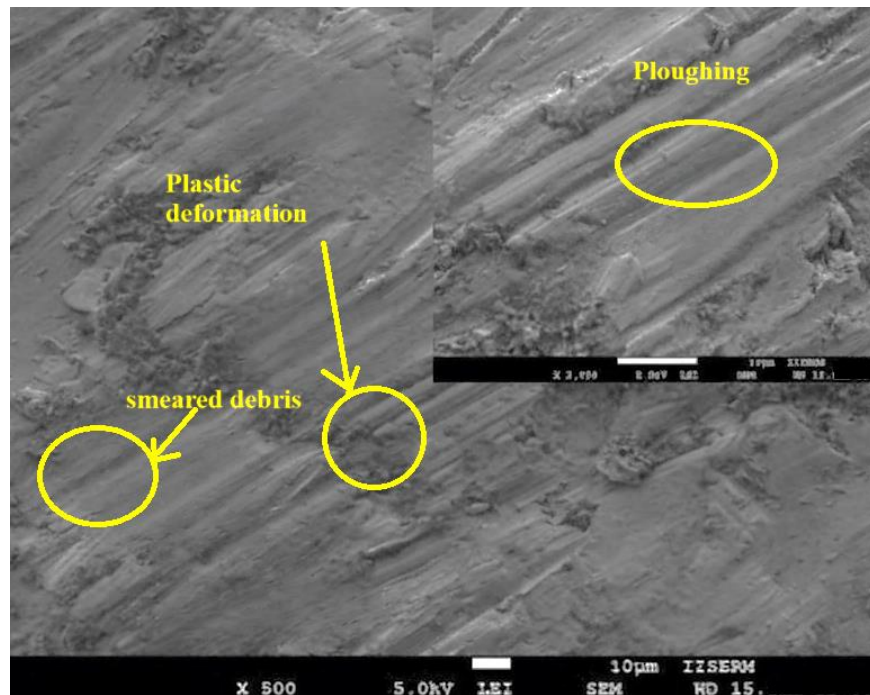


Figure 5.15 Fractography of worn surface of EDM-treated β Ti implant after wear test

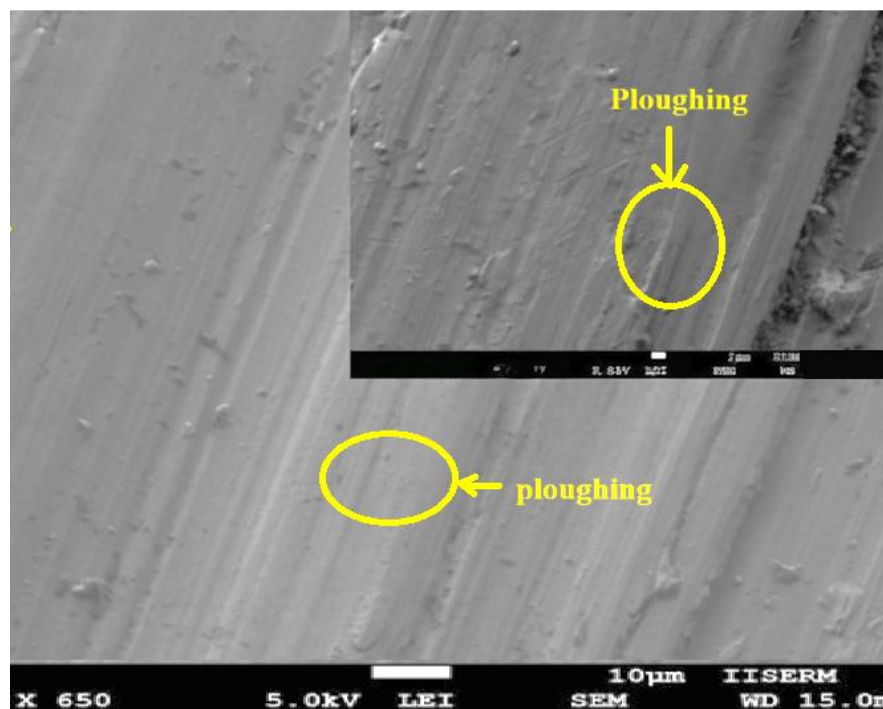


Figure 5.16 Fractography of worn surface of BB-EDC treated Ti implant after wear test

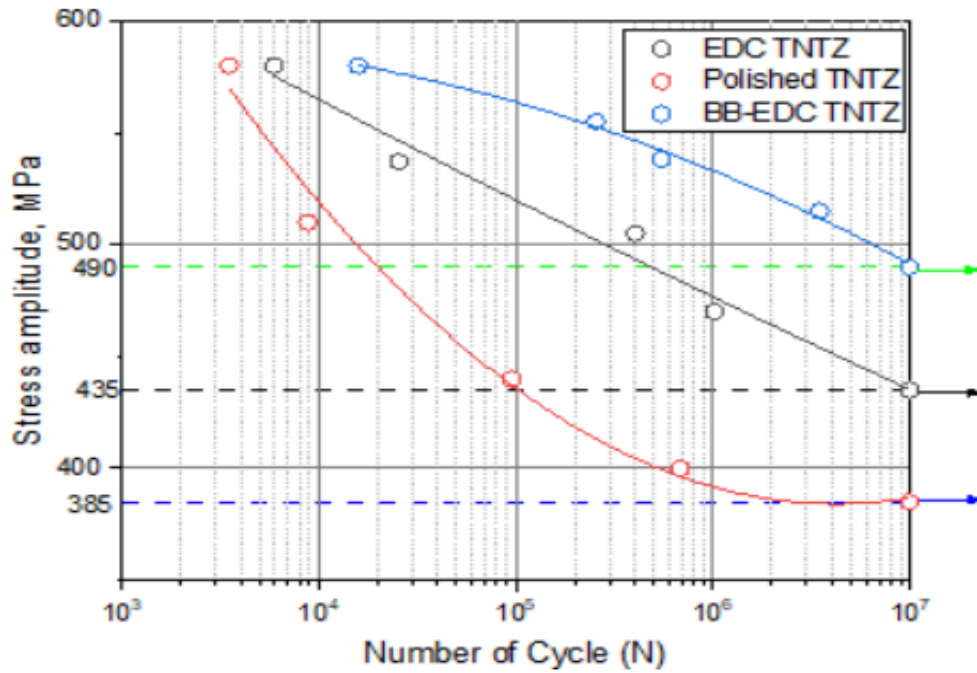


Figure 5.17 Biomechanical Fatigue on Un-treated, EDC and BB-EDC

The fatigue endurance strength of the EDC-treated specimen was observed at 435 MPa. Overall, the BB-EDC processed samples demonstrated superior.

5.13 Fractography

Figure 5.18 shows the failed specimens fractography under high cycle fatigue stress for three test batches: Untreated, EDC-treated, and BB-EDC samples. FE-SEM fractography study helps understand how surface modification methods affect fracture initiation and propagation.

In the picture, most samples had fracture initiation sites (dotted red circles), citation marks (dotted yellow arrows), and thick ridges. Figure 5.18 (a) shows the untreated surface fractography at 385 MPa fatigue limit at 500× magnification. Untreated specimens had many fracture initiations, ductile dimples, poor surface hardness (285 HV), and triggered crack nucleation and propagation, failing under high cyclic load. Figure 5.18 (b) shows substantial crack propagation on the EDC-treated surface, as shown by solid yellow arrows indicating fractures and fracture patterns at 500× magnification. Striation markings, chevron-like patterns, and secondary crack initiation are seen at 2500× magnification. The hardness of EDC surface is high (355 HV) as compared to untreated surface and fatigue limit corresponds to EDC is 435 MPa. Consequently, higher fatigue performance is observed in EDC-modified specimens

compared to untreated, due to fewer defects and the presence of an HAP-cladded layer providing high surface hardness, which transforms the microstructure and reduces crack propagation.

As depicted in Figure 5.18 (c) the BB-EDC modified specimens exhibit a fatigue limit value of 490 MPa, which is the highest among both the EDC-treated and untreated specimens. Additionally, the hardness value of the BB-EDC modified specimens is also found the highest (848 MPa).

This enhanced performance can be attributed to factors such as fine grain growth, increased surface hardness, and residual stresses induced by the burnishing balls within the dense TiO₂-loaded HAP-cladded layer, which provided significant resistance to fatigue failure and crack formation. Moreover, the fractography of the BB-EDC processed samples differs substantially from that of the untreated and EDC-modified specimens. At a high magnification of 2500 $\times$, striation marks and a few crack initiation sites are visible.

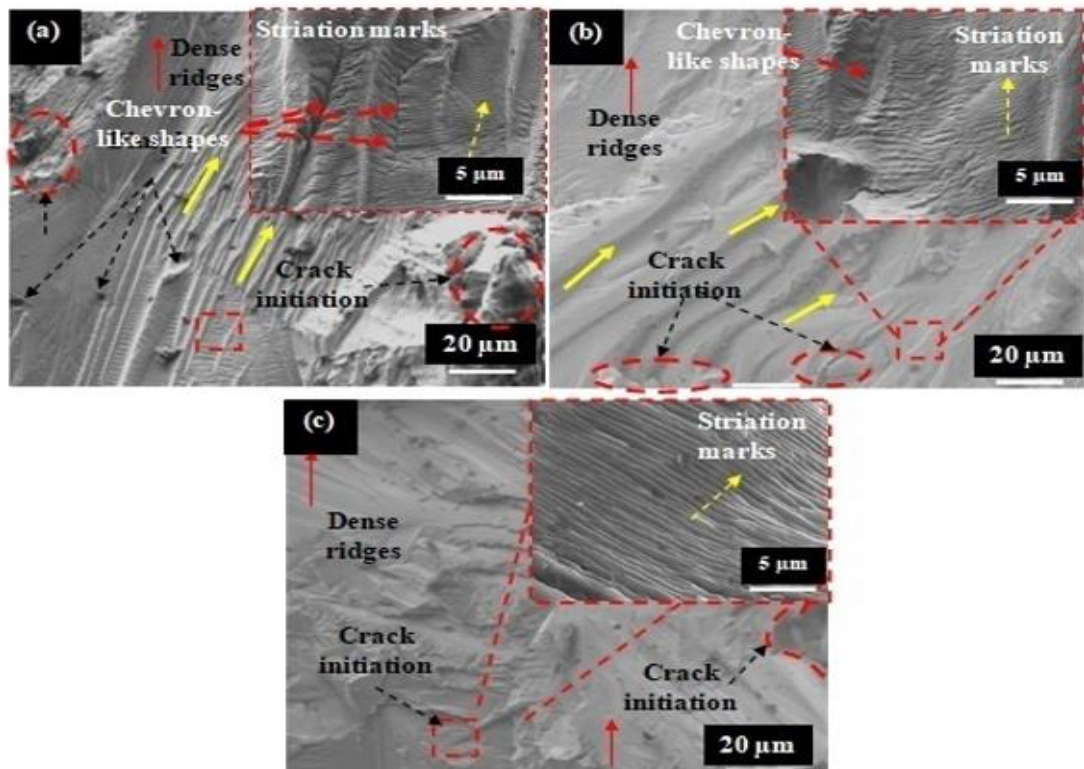


Figure 5.18 Fractography of fractured sample (a) un-treated, (b) EDC treated, and (c) BB-EDC treated (At magnification of 500 $\times$ and 2500 $\times$)

This enhanced performance can be attributed to factors such as fine grain growth, increased surface hardness, and residual stresses induced by the burnishing balls within the dense TiO₂-loaded HAP-cladded layer, which provided significant resistance to fatigue failure and crack formation. Moreover, the fractography of the BB-EDC processed samples differs substantially from that of the untreated and EDC-modified specimens. At a high magnification of 2500×, striation marks and a few crack initiation sites are visible.

5.14 In-Vitro bioactivity

In vitro bioactivity of test specimens, such as bare, EDC, and BB-EDC treated β -Ti alloy samples, is depicted in Figure 5.19 and Figure 5.20. All samples showed a rise in cell count and related optical density (a measure of cell viability and proliferation) with longer incubation times following the MTT experiment. Observations of the number of cells and their density across the untreated, EDC, and BB-EDC processed samples, on the first and third days revealed changes, as shown in Figure 5.19 and Figure 5.20 respectively.

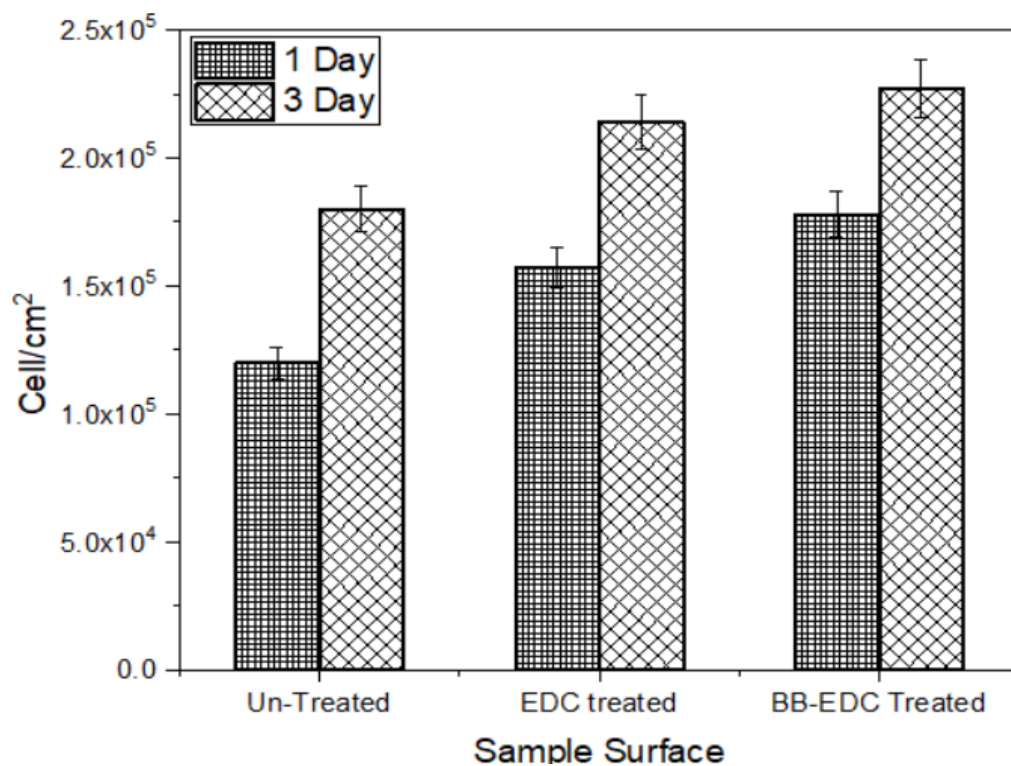


Figure 5.19 Cell count on Un-treated, EDC, and BB-EDC treated β -Ti alloy

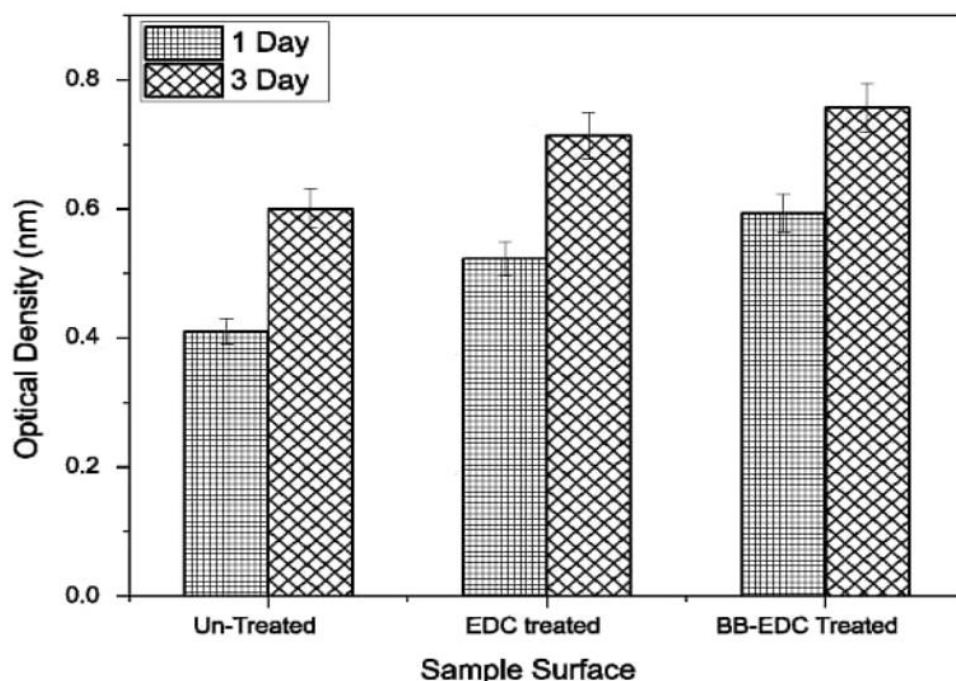


Figure 5.20 Cell viability and proliferation on untreated, EDC-treated, and BB-EDC-treated β -Ti alloy

It was noted that cell growth and the incubation period increased, as evidenced by the rise in cell count and optical density. The cell number growth and optical density on the untreated surface were significantly lower compared to the BB-EDC modified surface. This is because the untreated surface was plain, smooth, and lacked surface features. The biocompatibility of the untreated β -Ti alloy specimens is solely due to the elemental composition of the base material (Prakash & Uddin, 2017).

The BB-EDC modified surface significantly outperformed in the growth and proliferation of osteoblastic cells. This is due to the cladded layer containing a homogeneously distributed TiO_2 -loaded HAP layer within sub-micron scale craters, which provides excellent surface energy, promoting uniform cell growth and rapid proliferation across the entire surface. In another observation as per Figure 5.21, is showing the behavior of the cells attached to all the samples, along with fluorescent images and FE-SEM images after the third day. The results confirmed the biocompatibility and absence of cytotoxicity in all samples. Figure 5.21 (a-b) presents fluorescent images and FE-SEM images of the number of cells attached to the untreated surface after the third day, where the untreated surface was found having lower number of cells attached. Figure 5.21 (c-d) clears that the EDC-treated surface exhibits higher

cell growth, spread and attachment, confirming the presence of HAP layer for better osseointegration. Figure 5.21(e-f) notifies the excellent cell growth toward the BB-EDC samples, using FE-SEM and fluorescent images after 3rd day.

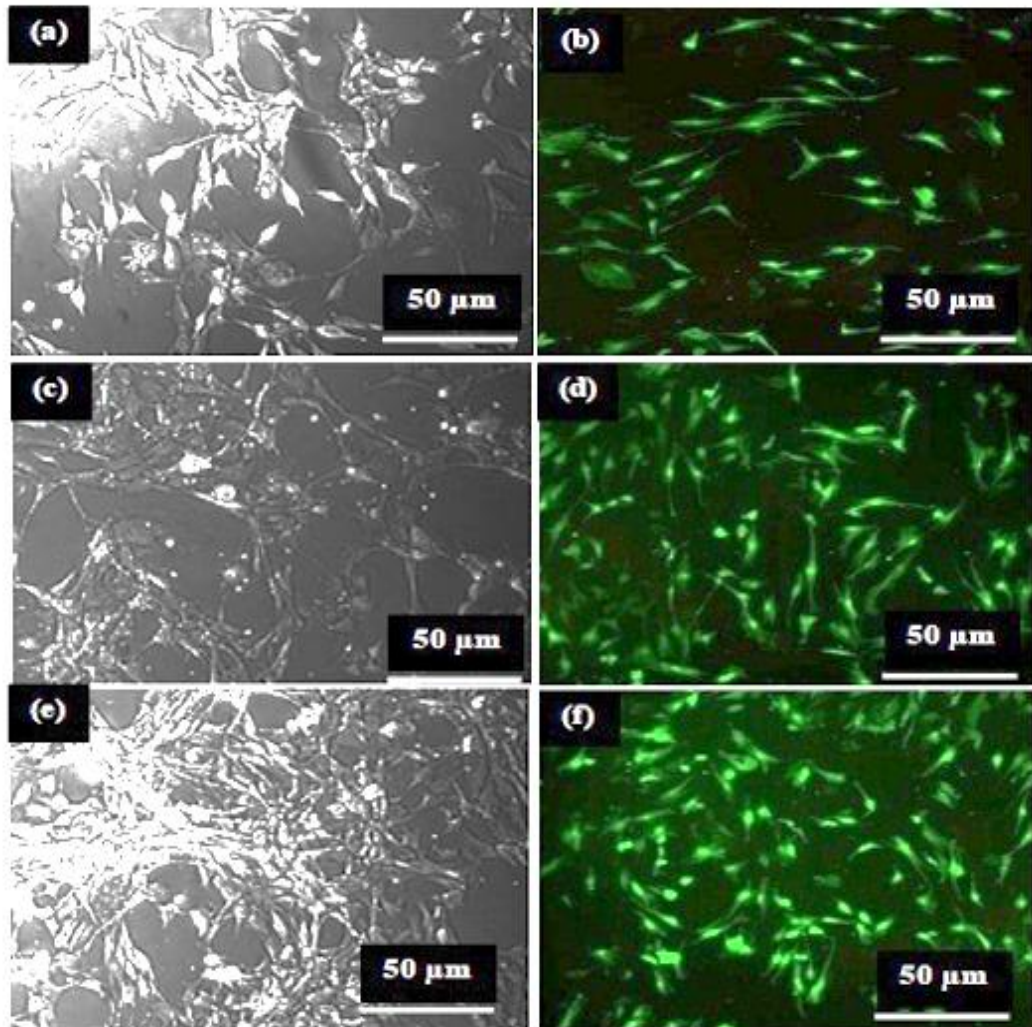


Figure 5.21 Cell adhesion and fluorescent images after 3 days of culturing: (a-b) Untreated, (c-d) EDC treated, and (e-f) BB-EDC treated.

5.15 Economy of the Process and Its Impact on Environment

The Ball Burnishing Assisted Electrical Discharge Cladding (BB-EDC) process presents a highly efficient and cost-effective method for surface modification of β -Ti alloys intended for orthopedic implants. Unlike conventional surface treatment techniques such as Chemical Vapor Deposition (CVD), Physical Vapor Deposition (PVD), and laser surface engineering—which often require expensive vacuum systems,

high-temperature environments, and sophisticated machinery—BB-EDC utilizes readily available EDM infrastructure with minor modifications, significantly reducing capital and operational expenditure.

Economic Advantages:

1. **Low Equipment Cost:** The BB-EDC setup integrates a rotary tool and slurry delivery system into a conventional EDM machine, eliminating the need for complex, high-cost equipment.
2. **Reduced Material Wastage:** As BB-EDC is a non-contact process, material removal is minimal and localized. The powder-based slurry ensures controlled deposition of cladding material with minimal waste.
3. **Dual Functionality:** The process combines deposition (EDC) and mechanical finishing (burnishing) in a single step, reducing the number of processing stages and overall machining time.
4. **Minimal Tool Wear:** The use of reverse polarity and controlled spark energy prolongs the life of the electrode, reducing consumable costs.
5. **Energy Efficiency:** Operating at moderate electrical parameters (e.g., 5–15 A peak current, 60–120 V supply), BB-EDC consumes relatively less energy compared to high-energy beam processes, making it more cost-efficient for industrial adoption.

Environmental Impact:

1. **Eco-Friendly Dielectric:** Unlike traditional EDM fluids based on hydrocarbon oils, the BB-EDC process employs distilled water as the dielectric medium.
2. **Reduced Emissions:** The slurry-based system with TiO₂ and HAP powders does not generate toxic fumes or particulate emissions, ensuring a cleaner work environment.
3. **Recyclable Waste:** The used slurry can be filtered and reused multiple times, minimizing the generation of chemical waste and promoting sustainable material use.
4. **Minimal Heat Affected Zone (HAZ):** The controlled thermal input reduces the size and severity of HAZ and eliminates the need for post-processing treatments that consume additional energy and chemicals.

CONCLUSION AND SCOPE FOR FUTURE

The current research work examines the potential and efficacy of electrical discharge cladding in hybridization with ball burnishing to modify the β -Ti implant alloy.

6.1 Conclusion

A novel HAP-Coating, achieved through Ball-Burnishing assisted electrical discharge cladding, has been successfully applied to the β -Ti surface for potential integration into orthopaedic implants. After thorough analysis of the research presented on β -Ti alloy machining utilizing BB-EDC, the main conclusions are outlined below:

1. The study confirms the effectiveness of Ball Burnishing-Assisted Electrical Discharge Cladding (BB-EDC) as a potent surface engineering method for β -phase titanium alloys. It offers substantial improvements over conventional Electrical Discharge Machining (EDM) by eliminating major surface defects and delivering a uniform, reinforced surface finish suitable for orthopedic biomedical implants.
2. The surface roughness (SR) of BB-EDC-treated β -Ti alloy was significantly reduced to 1.16 μm from the 2.61 μm observed in EDM-treated surfaces. The reduction in roughness directly correlates to fewer micro-cracks and smoother surfaces, thereby enhancing corrosion and fatigue performance.
3. The surface hardness improved drastically with BB-EDC. The microhardness of the BB-EDC surface reached 848 HV, compared to 355 HV for EDM and 285 HV for untreated surfaces, which is attributed to strain hardening induced by the burnishing balls and the uniform cladding of TiO_2 -reinforced HAP layers.
4. The recast layer thickness after BB-EDC treatment ranged from 10 to 15 μm , and cross-sectional analysis revealed a flat and uniform layer without visible surface irregularities or porosity. This layer exhibited excellent integrity and contributes to overall structural strength.

5. Corrosion resistance testing, based on Tafel polarization, revealed a substantial enhancement in BB-EDC specimens, with I_{corr} reduced to $-8.18 \mu A/cm^2$ and E_{cor} shifted to $-222 mV$, indicating the formation of a stable and homogeneous TiO_2 -loaded HAP passive layer. These values outperform both untreated and EDM-treated counterparts, confirming superior resistance to corrosion.
6. The wear rate decreased drastically from $0.00003 mm^3/N \cdot m$ in untreated specimens to $0.000011 mm^3/N \cdot m$ in BB-EDC-treated ones. The coefficient of friction (COF) was also minimized to 0.35, compared to 0.5 for untreated and 0.39 for EDM surfaces. Fractography showed that BB-EDC minimized volume loss due to brittle fracture or plastic deformation, as only ploughing marks were visible, indicating a transition to milder wear mechanisms.
7. In fatigue testing, BB-EDC specimens achieved a fatigue limit of 490 MPa, outperforming both untreated (385 MPa) and EDM-treated (435 MPa) samples. Fractographic analysis confirmed reduced crack initiation and propagation, with striation patterns and fewer initiation sites observed at higher magnification ($2500\times$), which is attributed to finer grains and residual compressive stresses induced during the BB-EDC process.
8. The in-vitro biocompatibility of BB-EDC surfaces was notably superior. Cell proliferation studies using the MG-63 osteoblast line showed higher cell adhesion and viability after 3 days of incubation compared to untreated and EDM-treated surfaces. The TiO_2 -reinforced HAP layer with submicron porosity offered excellent surface energy for cell attachment and growth, thus validating BB-EDC as a bioactive and cytocompatible surface.
9. BB-EDC-treated surfaces demonstrated optimized surface chemistry and morphology, as verified via FE-SEM and EDS analysis. Uniform distribution of Ca, P, and Ti elements in the cladded layer confirmed successful deposition of the HAP coating. The structural composition and microstructure analysis revealed a dense, homogenous, and defect-free top layer favorable for biomedical integration.
10. From a process optimization perspective, BB-EDC also exhibited better process control over parameters such as peak current, pulse duration, and voltage. It was observed that increased current and voltage during EDM resulted in deeper

craters and higher roughness, while BB-EDC maintained uniformity and superior surface characteristics even under varied input conditions.

6.2 Scope for Future Research Work

- Ball-Burnishing Assisted Electrical Discharge Cladding (BB-EDC) as a surface engineering technique has shown promising results and thus offering it promising for various engineering applications beyond the scope of this study.
- Moving forward, a more comprehensive investigation is needed to understand the influence of non-electrical factors, such as workpiece and electrode rotation, on the surface modification of β -Ti alloy. This will contribute to a deeper understanding of the BB-EDC process and its optimization for diverse industrial applications.
- With BB-EDC treated β -Ti implant, finite element modeling (F.E.M.) can be done to predict the distribution of stress and bone in-growth through various loading situations.
- Further refinement of BB-EDC can be done using advanced robotic setups, enabling highly repeatable and reliable surface modifications for precision medical parts. This approach would further optimize the manufacturing of β -Ti implants with consistent quality.
- Furthermore, while the bioactivity and biocompatibility of BB-EDC treated β -Ti alloy implants have been evaluated through in-vitro studies involving cell culture, further research is essential to assess bone ingrowth effects in vivo through animal studies. This step is crucial for validating the efficacy and safety of BB-EDC treated implants in real-world scenarios and clinical applications.

References

- Abdullah, A., Shabgard, M. R., Ivanov, A., & Shervanyi-Tabar, M. T. (2009). Effect of ultrasonic-assisted EDM on the surface integrity of cemented tungsten carbide (WC-Co). *The International Journal of Advanced Manufacturing Technology*, 41(3–4), 268–280. <https://doi.org/10.1007/s00170-008-1476-7>
- Albrektsson, T., Dahlin, C., Jemt, T., Sennerby, L., Turri, A., & Wennerberg, A. (2014). Is Marginal Bone Loss around Oral Implants the Result of a Provoked Foreign Body Reaction? *Clinical Implant Dentistry and Related Research*, 16(2), 155–165. <https://doi.org/10.1111/cid.12142>
- Alipour, S., Nour, S., Attari, S. M., Mohajeri, M., Kianersi, S., Taromian, F., Khalkhali, M., Aninwene, G. E., & Tayebi, L. (2022). A review on *in vitro* / *in vivo* response of additively manufactured Ti–6Al–4V alloy. *Journal of Materials Chemistry B*, 10(46), 9479–9534. <https://doi.org/10.1039/D2TB01616H>
- Amani, H., Arzaghi, H., Bayandori, M., Dezfuli, A. S., Pazoki-Toroudi, H., Shafiee, A., & Moradi, L. (2019). Controlling Cell Behavior through the Design of Biomaterial Surfaces: A Focus on Surface Modification Techniques. *Advanced Materials Interfaces*, 6(13), 1900572. <https://doi.org/10.1002/admi.201900572>
- Anselme, K., Davidson, P., Popa, A. M., Giazgon, M., Liley, M., & Ploux, L. (2010). The interaction of cells and bacteria with surfaces structured at the nanometre scale. *Acta Biomaterialia*, 6(10), 3824–3846. <https://doi.org/10.1016/j.actbio.2010.04.001>
- Asri, R. I. M., Harun, W. S. W., Samykano, M., Lah, N. A. C., Ghani, S. A. C., Tarlochan, F., & Raza, M. R. (2017). Corrosion and surface modification on biocompatible metals: A review. *Materials Science and Engineering: C*, 77, 1261–1274. <https://doi.org/10.1016/j.msec.2017.04.102>
- Att, W., Tsukimura, N., Suzuki, T., & Ogawa, T. (2007). Effect of supramicron roughness characteristics produced by 1- and 2-step acid etching on the osseointegration capability of titanium. *The International Journal of Oral & Maxillofacial Implants*, 22(5), 719–728.
- Bahl, S., Das, S., Suwas, S., & Chatterjee, K. (2018). Engineering the next-generation tin containing β titanium alloys with high strength and low modulus for orthopedic

applications. *Journal of the Mechanical Behavior of Biomedical Materials*, 78, 124–133. <https://doi.org/10.1016/j.jmbbm.2017.11.014>

Bao Ha, T. L., Minh, T., Nguyen, D., & Minh, D. (2013). Naturally Derived Biomaterials: Preparation and Application. In J. A. Andrades (Ed.), *Regenerative Medicine and Tissue Engineering*. InTech. <https://doi.org/10.5772/55668>

Baroi, B. K., Jagadish, & Patowari, P. K. (2022). Parametric study of electric discharge machining of titanium grade 2 alloy in distilled water. *Materials Today: Proceedings*, 59, 1584–1590. <https://doi.org/10.1016/j.matpr.2022.02.275>

Bartolo, P., Kruth, J.-P., Silva, J., Levy, G., Malshe, A., Rajurkar, K., Mitsuishi, M., Ciurana, J., & Leu, M. (2012). Biomedical production of implants by additive electro-chemical and physical processes. *CIRP Annals*, 61(2), 635–655. <https://doi.org/10.1016/j.cirp.2012.05.005>

Batchelor, A. W., & Chandrasekaran, M. (2004). *Service Characteristics of Biomedical Materials and Implants* (Vol. 3). IMPERIAL COLLEGE PRESS. <https://doi.org/10.1142/p343>

Bauer, S., Schmuki, P., Von Der Mark, K., & Park, J. (2013). Engineering biocompatible implant surfaces. *Progress in Materials Science*, 58(3), 261–326. <https://doi.org/10.1016/j.pmatsci.2012.09.001>

Bheemappa, S., Gurumurthy, H., Badami, V. V., & Hegde, P. R. (2023). Tribological behavior of polymeric systems in lubricated surfaces or conditions. In *Tribology of Polymers, Polymer Composites, and Polymer Nanocomposites* (pp. 357–399). Elsevier. <https://doi.org/10.1016/B978-0-323-90748-4.00005-4>

Bojorquez, B., Marloth, R. T., & Es-Said, O. S. (2002). Formation of a crater in the workpiece on an electrical discharge machine. *Engineering Failure Analysis*, 9(1), 93–97. [https://doi.org/10.1016/S1350-6307\(00\)00028-5](https://doi.org/10.1016/S1350-6307(00)00028-5)

Boyan, B. D., Bonewald, L. F., Paschalis, E. P., Lohmann, C. H., Rosser, J., Cochran, D. L., Dean, D. D., Schwartz, Z., & Boskey, A. L. (2002). Osteoblast-Mediated Mineral Deposition in Culture is Dependent on Surface Microtopography. *Calcified Tissue International*, 71(6), 519–529. <https://doi.org/10.1007/s00223-001-1114-y>

Bruschi, M., Steinmüller-Nethl, D., Goriwoda, W., & Rasse, M. (2015). Composition and Modifications of Dental Implant Surfaces. *Journal of Oral Implants*, 2015, 1–14. <https://doi.org/10.1155/2015/527426>

- Bucciotti, F., Mazzocchi, M., & Bellosi, A. (2010). Perspectives of the Si₃N₄-TiN ceramic composite as a biomaterial and manufacturing of complex-shaped implantable devices by electrical discharge machining (EDM). *Journal of Applied Biomaterials & Biomechanics: JABB*, 8(1), 28–32.
- Buser, D., Broggini, N., Wieland, M., Schenk, R. K., Denzer, A. J., Cochran, D. L., Hoffmann, B., Lussi, A., & Steinemann, S. G. (2004). Enhanced Bone Apposition to a Chemically Modified SLA Titanium Surface. *Journal of Dental Research*, 83(7), 529–533. <https://doi.org/10.1177/154405910408300704>
- Chang-bin, T., Dao-xin, L., Zhan, W., & Yang, G. (2011). Electro-spark alloying using graphite electrode on titanium alloy surface for biomedical applications. *Applied Surface Science*, 257(15), 6364–6371. <https://doi.org/10.1016/j.apsusc.2011.01.120>
- Chen, S. L., Hsieh, S. F., Lin, H. C., Lin, M. H., & Huang, J. S. (2008). Electrical discharge machining of a NiAlFe ternary shape memory alloy. *Journal of Alloys and Compounds*, 464(1–2), 446–451. <https://doi.org/10.1016/j.jallcom.2007.10.012>
- Chen, S.-L., Lin, M.-H., Chen, C.-C., & Ou, K.-L. (2008). Effect of electro-discharging on formation of biocompatible layer on implant surface. *Journal of Alloys and Compounds*, 456(1–2), 413–418. <https://doi.org/10.1016/j.jallcom.2007.02.055>
- Chen, S.-L., Lin, M.-H., Huang, G.-X., & Wang, C.-C. (2014). Research of the recast layer on implant surface modified by micro-current electrical discharge machining using deionized water mixed with titanium powder as dielectric solvent. *Applied Surface Science*, 311, 47–53. <https://doi.org/10.1016/j.apsusc.2014.04.204>
- Curtis, A. S. G., Gadegaard, N., Dalby, M. J., Riehle, M. O., Wilkinson, C. D. W., & Aitchison, G. (2004). Cells React to Nanoscale Order and Symmetry in Their Surroundings. *IEEE Transactions on Nanobioscience*, 3(1), 61–65. <https://doi.org/10.1109/TNB.2004.824276>
- Daneshmand, S., Kahrizi, E. F., Neyestanak, A. A. L., & Ghahi, M. M. (2013). Experimental Investigations into Electro Discharge Machining of NiTi Shape Memory Alloys using Rotational Tool. *International Journal of Electrochemical Science*, 8(6), 7484–7497. [https://doi.org/10.1016/S1452-3981\(23\)12820-3](https://doi.org/10.1016/S1452-3981(23)12820-3)

Das, S., Paul, S., & Doloi, B. (2021). A gap-active electrical discharge machining (GA-EDM) to rectify the textural defects of the processed surface. *Journal of Manufacturing Processes*, 64, 594–605. <https://doi.org/10.1016/j.jmapro.2021.01.053>

Devgan, S., & Sidhu, S. S. (2020). Surface modification of β -type titanium with multi-walled CNTs/ μ -HAp powder mixed Electro Discharge Treatment process. *Materials Chemistry and Physics*, 239, 122005. <https://doi.org/10.1016/j.matchemphys.2019.122005>

Duraccio, D., Mussano, F., & Faga, M. G. (2015). Biomaterials for dental implants: Current and future trends. *Journal of Materials Science*, 50(14), 4779–4812. <https://doi.org/10.1007/s10853-015-9056-3>

Ekmekci, N., & Efe, Y. (2022). The effect of nano and micro hydroxyapatite powder additives on surface integrity in electrical discharge machining of Ti6Al4V alloy. *Surface and Coatings Technology*, 445, 128708. <https://doi.org/10.1016/j.surfcoat.2022.128708>

Ekmekci, N., & Ekmekci, B. (2016). Electrical Discharge Machining of Ti6Al4V in Hydroxyapatite Powder Mixed Dielectric Liquid. *Materials and Manufacturing Processes*, 31(13), 1663–1670. <https://doi.org/10.1080/10426914.2015.1090591>

ElHawary, H., Baradaran, A., Abi-Rafah, J., Vorstenbosch, J., Xu, L., & Efanov, J. I. (2021). Bone Healing and Inflammation: Principles of Fracture and Repair. *Seminars in Plastic Surgery*, 35(03), 198–203. <https://doi.org/10.1055/s-0041-1732334>

Elias, C., Oshida, Y., Lima, J., & Muller, C. (2008). Relationship between surface properties (roughness, wettability and morphology) of titanium and dental implant removal torque. *Journal of the Mechanical Behavior of Biomedical Materials*, 1(3), 234–242. <https://doi.org/10.1016/j.jmbbm.2007.12.002>

Eliaz, N. (2019). Corrosion of Metallic Biomaterials: A Review. *Materials*, 12(3), 407. <https://doi.org/10.3390/ma12030407>

Feller, L., Jadwat, Y., Khammissa, R. A. G., Meyerov, R., Schechter, I., & Lemmer, J. (2015). Cellular Responses Evoked by Different Surface Characteristics of Intraosseous Titanium Implants. *BioMed Research International*, 2015, 1–8. <https://doi.org/10.1155/2015/171945>

Furutania, K., Saneto, A., Takezawa, H., Mohri, N., & Miyake, H. (2001). Accretion of titanium carbide by electrical discharge machining with powder suspended in

working fluid. *Precision Engineering*, 25(2), 138–144. [https://doi.org/10.1016/S0141-6359\(00\)00068-4](https://doi.org/10.1016/S0141-6359(00)00068-4)

Geetha, M., Singh, A. K., Asokamani, R., & Gogia, A. K. (2009). Ti based biomaterials, the ultimate choice for orthopaedic implants – A review. *Progress in Materials Science*, 54(3), 397–425. <https://doi.org/10.1016/j.pmatsci.2008.06.004>

Gentleman, M. M., & Gentleman, E. (2014). The role of surface free energy in osteoblast–biomaterial interactions. *International Materials Reviews*, 59(8), 417–429. <https://doi.org/10.1179/1743280414Y.0000000038>

Gittens, R. A., McLachlan, T., Olivares-Navarrete, R., Cai, Y., Berner, S., Tannenbaum, R., Schwartz, Z., Sandhage, K. H., & Boyan, B. D. (2011). The effects of combined micron-/submicron-scale surface roughness and nanoscale features on cell proliferation and differentiation. *Biomaterials*, 32(13), 3395–3403. <https://doi.org/10.1016/j.biomaterials.2011.01.029>

Gittens, R. A., Olivares-Navarrete, R., Schwartz, Z., & Boyan, B. D. (2014). Implant osseointegration and the role of microroughness and nanostructures: Lessons for spine implants. *Acta Biomaterialia*, 10(8), 3363–3371. <https://doi.org/10.1016/j.actbio.2014.03.037>

Gohil, V., & Puri, Y. M. (2018). Statistical analysis of material removal rate and surface roughness in electrical discharge turning of titanium alloy (Ti-6Al-4V). *Proceedings of the Institution of Mechanical Engineers, Part B: Journal of Engineering Manufacture*, 232(9), 1603–1614. <https://doi.org/10.1177/0954405416673104>

Harcuba, P., Bačáková, L., Stráský, J., Bačáková, M., Novotná, K., & Janeček, M. (2012). Surface treatment by electric discharge machining of Ti-6Al-4V alloy for potential application in orthopaedics. *Journal of the Mechanical Behavior of Biomedical Materials*, 7, 96–105. <https://doi.org/10.1016/j.jmbbm.2011.07.001>

Hasegawa, M., Saruta, J., Hirota, M., Taniyama, T., Sugita, Y., Kubo, K., Ishijima, M., Ikeda, T., Maeda, H., & Ogawa, T. (2020). A Newly Created Meso-, Micro-, and Nano-Scale Rough Titanium Surface Promotes Bone-Implant Integration. *International Journal of Molecular Sciences*, 21(3), 783. <https://doi.org/10.3390/ijms21030783>

Havlikova, J., Strasky, J., Vandrovcova, M., Hrcuba, P., Mhaede, M., Janecek, M., & Bacakova, L. (2014). Innovative surface modification of Ti-6Al-4V alloy with a positive effect on osteoblast proliferation and fatigue performance. *Materials Science and Engineering: C*, 39, 371–379. <https://doi.org/10.1016/j.msec.2014.03.024>

Henao, J., Poblano-Salas, C., Monsalve, M., Corona-Castuera, J., & Barceinas-Sanchez, O. (2019). Bio-active glass coatings manufactured by thermal spray: A status report. *Journal of Materials Research and Technology*, 8(5), 4965–4984. <https://doi.org/10.1016/j.jmrt.2019.07.011>

Ho, K. H., & Newman, S. T. (2003). State of the art electrical discharge machining (EDM). *International Journal of Machine Tools and Manufacture*, 43(13), 1287–1300. [https://doi.org/10.1016/S0890-6955\(03\)00162-7](https://doi.org/10.1016/S0890-6955(03)00162-7)

Hong, J.-Y., Ko, S.-Y., Lee, W., Chang, Y.-Y., Kim, S.-H., & Yun, J.-H. (2020). Enhancement of Bone Ingrowth into a Porous Titanium Structure to Improve Osseointegration of Dental Implants: A Pilot Study in the Canine Model. *Materials*, 13(14), 3061. <https://doi.org/10.3390/ma13143061>

Huang, H., Zhang, H., Zhou, L., & Zheng, H. Y. (2003). Ultrasonic vibration assisted electro-discharge machining of microholes in Nitinol. *Journal of Micromechanics and Microengineering*, 13(5), 693–700. <https://doi.org/10.1088/0960-1317/13/5/322>

Indira, K., Kamachi Mudali, U., & Rajendran, N. (2013). Corrosion behavior of electrochemically assembled nanoporous titania for biomedical applications. *Ceramics International*, 39(2), 959–967. <https://doi.org/10.1016/j.ceramint.2012.07.013>

Jain, S., & Parashar, V. (2021). Critical review on the impact of EDM process on biomedical materials. *Materials and Manufacturing Processes*, 36(15), 1701–1724. <https://doi.org/10.1080/10426914.2021.1942907>

Jimi, E., Hirata, S., Osawa, K., Terashita, M., Kitamura, C., & Fukushima, H. (2012). The Current and Future Therapies of Bone Regeneration to Repair Bone Defects. *International Journal of Dentistry*, 2012, 1–7. <https://doi.org/10.1155/2012/148261>

Kansal, H. K., Singh, S., & Kumar, P. (2005). Parametric optimization of powder mixed electrical discharge machining by response surface methodology. *Journal of Materials Processing Technology*, 169(3), 427–436. <https://doi.org/10.1016/j.jmatprotec.2005.03.028>

- Kieswetter, K., Schwartz, Z., Dean, D. D., & Boyan, B. D. (1996). The Role of Implant Surface Characteristics in the Healing of Bone. *Critical Reviews in Oral Biology & Medicine*, 7(4), 329–345. <https://doi.org/10.1177/10454411960070040301>
- Kolli, R., & Devaraj, A. (2018). A Review of Metastable Beta Titanium Alloys. *Metals*, 8(7), 506. <https://doi.org/10.3390/met8070506>
- Kopova, I., Stráský, J., Harcuba, P., Landa, M., Janeček, M., & Bačáková, L. (2016). Newly developed Ti–Nb–Zr–Ta–Si–Fe biomedical beta titanium alloys with increased strength and enhanced biocompatibility. *Materials Science and Engineering: C*, 60, 230–238. <https://doi.org/10.1016/j.msec.2015.11.043>
- Kubo, K., Tsukimura, N., Iwasa, F., Ueno, T., Saruwatari, L., Aita, H., Chiou, W.-A., & Ogawa, T. (2009). Cellular behavior on TiO₂ nanonodular structures in a micro-to-nanoscale hierarchy model. *Biomaterials*, 30(29), 5319–5329. <https://doi.org/10.1016/j.biomaterials.2009.06.021>
- Kumar, S., Singh, R., Singh, T. P., & Sethi, B. L. (2009). Surface modification by electrical discharge machining: A review. *Journal of Materials Processing Technology*, 209(8), 3675–3687. <https://doi.org/10.1016/j.jmatprotec.2008.09.032>
- Kunieda, M., Lauwers, B., Rajurkar, K. P., & Schumacher, B. M. (2005). Advancing EDM through Fundamental Insight into the Process. *CIRP Annals*, 54(2), 64–87. [https://doi.org/10.1016/S0007-8506\(07\)60020-1](https://doi.org/10.1016/S0007-8506(07)60020-1)
- Lee, W.-F., Yang, T.-S., Wu, Y.-C., & Peng, P.-W. (2013). Nanoporous Biocompatible Layer on Ti–6Al–4V Alloys Enhanced Osteoblast-like Cell Response. *Journal of Experimental & Clinical Medicine*, 5(3), 92–96. <https://doi.org/10.1016/j.jecm.2013.04.002>
- Li, Y., Yang, C., Zhao, H., Qu, S., Li, X., & Li, Y. (2014). New Developments of Ti-Based Alloys for Biomedical Applications. *Materials*, 7(3), 1709–1800. <https://doi.org/10.3390/ma7031709>
- Liew, P. J., Yap, C. Y., Wang, J., Zhou, T., & Yan, J. (2020). Surface modification and functionalization by electrical discharge coating: A comprehensive review. *International Journal of Extreme Manufacturing*, 2(1), 012004. <https://doi.org/10.1088/2631-7990/ab7332>
- Lin, Y. C., Yan, B. H., & Huang, F. Y. (2001). Surface Improvement Using a Combination of Electrical Discharge Machining with Ball Burnish Machining Based

on the Taguchi Method. *The International Journal of Advanced Manufacturing Technology*, 18(9), 673–682. <https://doi.org/10.1007/s001700170028>

Lipski, A. M., Pino, C. J., Haselton, F. R., Chen, I.-W., & Shastri, V. P. (2008). The effect of silica nanoparticle-modified surfaces on cell morphology, cytoskeletal organization and function. *Biomaterials*, 29(28), 3836–3846. <https://doi.org/10.1016/j.biomaterials.2008.06.002>

Liu, J. F., & Guo, Y. B. (2015). Process Capability of Wire-EDM of NiTi Shape Memory Alloy at Main Cut and Trim Cut Modes. *Procedia Manufacturing*, 1, 904–914. <https://doi.org/10.1016/j.promfg.2015.09.083>

Liu, Y., Ji, R., Zhang, Y., & Zhang, H. (2010). Investigation of emulsion for die sinking EDM. *The International Journal of Advanced Manufacturing Technology*, 47(1–4), 403–409. <https://doi.org/10.1007/s00170-009-2209-2>

Long, M., & Rack, H. J. (1998). Titanium alloys in total joint replacement—A materials science perspective. *Biomaterials*, 19(18), 1621–1639. [https://doi.org/10.1016/S0142-9612\(97\)00146-4](https://doi.org/10.1016/S0142-9612(97)00146-4)

Luo, Y. F. (1997). The dependence of interspace discharge transitivity upon the gap debris in precision electrodischarge machining. *Journal of Materials Processing Technology*, 68(2), 121–131. [https://doi.org/10.1016/S0924-0136\(96\)00019-2](https://doi.org/10.1016/S0924-0136(96)00019-2)

Manjaiah, M., Narendranath, S., Basavarajappa, S., & Gaitonde, V. N. (2014). Wire electric discharge machining characteristics of titanium nickel shape memory alloy. *Transactions of Nonferrous Metals Society of China*, 24(10), 3201–3209. [https://doi.org/10.1016/S1003-6326\(14\)63461-0](https://doi.org/10.1016/S1003-6326(14)63461-0)

Martin, J. Y., Schwartz, Z., Hummert, T. W., Schraub, D. M., Simpson, J., Lankford, J., Dean, D. D., Cochran, D. L., & Boyan, B. D. (1995). Effect of titanium surface roughness on proliferation, differentiation, and protein synthesis of human osteoblast-like cells (MG63). *Journal of Biomedical Materials Research*, 29(3), 389–401. <https://doi.org/10.1002/jbm.820290314>

McGeough, J. A. (1988). *Advanced methods of machining*. Chapman and Hall.

Mendonça, G., Mendonça, D. B. S., Aragão, F. J. L., & Cooper, L. F. (2010). The combination of micron and nanotopography by H₂SO₄/H₂O₂ treatment and its effects on osteoblast-specific gene expression of hMSCs. *Journal of Biomedical Materials Research Part A*, 94A(1), 169–179. <https://doi.org/10.1002/jbm.a.32701>

Mohammadi, H., Muhamad, N., Sulong, A. B., & Ahmadipour, M. (2021). Recent advances on biofunctionalization of metallic substrate using ceramic coating: How far are we from clinically stable implant? *Journal of the Taiwan Institute of Chemical Engineers*, 118, 254–270. <https://doi.org/10.1016/j.jtice.2021.01.013>

Mohammed, M. T., Khan, Z. A., & Siddiquee, A. N. (2014). *Beta Titanium Alloys: The Lowest Elastic Modulus For Biomedical Applications: A Review*. <https://doi.org/10.5281/ZENODO.1094481>

Mohd Abbas, N., Solomon, D. G., & Fuad Bahari, Md. (2007). A review on current research trends in electrical discharge machining (EDM). *International Journal of Machine Tools and Manufacture*, 47(7–8), 1214–1228. <https://doi.org/10.1016/j.ijmachtools.2006.08.026>

Munir, K., Lin, J., Wright, P. F. A., Ozan, S., Li, Y., & Wen, C. (2022). Mechanical, corrosion, nanotribological, and biocompatibility properties of equal channel angular pressed Ti-28Nb-35.4Zr alloys for biomedical applications. *Acta Biomaterialia*, 149, 387–398. <https://doi.org/10.1016/j.actbio.2022.07.005>

Nag, S., & Banerjee, R. (2012). Fundamentals of Medical Implant Materials. In R. J. Narayan (Ed.), *Materials for Medical Devices* (pp. 6–17). ASM International. <https://doi.org/10.31399/asm.hb.v23.a0005682>

Ndaliman, M. B., Khan, A. A., & Ali, M. Y. (2013). Influence of electrical discharge machining process parameters on surface micro-hardness of titanium alloy. *Proceedings of the Institution of Mechanical Engineers, Part B: Journal of Engineering Manufacture*, 227(3), 460–464. <https://doi.org/10.1177/0954405412470443>

Niinomi, M. (2003). Recent research and development in titanium alloys for biomedical applications and healthcare goods. *Science and Technology of Advanced Materials*, 4(5), 445–454. <https://doi.org/10.1016/j.stam.2003.09.002>

Niinomi, M. (2007). Fatigue characteristics of metallic biomaterials. *International Journal of Fatigue*, 29(6), 992–1000. <https://doi.org/10.1016/j.ijfatigue.2006.09.021>

Niinomi, M., Nakai, M., & Hieda, J. (2012). Development of new metallic alloys for biomedical applications. *Acta Biomaterialia*, 8(11), 3888–3903. <https://doi.org/10.1016/j.actbio.2012.06.037>

- Nobles, K. P., Janorkar, A. V., & Williamson, R. S. (2021). Surface modifications to enhance osseointegration—Resulting material properties and biological responses. *Journal of Biomedical Materials Research Part B: Applied Biomaterials*, 109(11), 1909–1923. <https://doi.org/10.1002/jbm.b.34835>
- Okazaki, Y. (2001). A New Ti–15Zr–4Nb–4Ta alloy for medical applications. *Current Opinion in Solid State and Materials Science*, 5(1), 45–53. [https://doi.org/10.1016/S1359-0286\(00\)00025-5](https://doi.org/10.1016/S1359-0286(00)00025-5)
- P., K., Mohanty, S., & Das, A. K. (2022). Sustainable surface modification of Ti-alloy using powder mixed in bio-dielectrics through micro-electrical discharge coating process. *Journal of Cleaner Production*, 362, 132375. <https://doi.org/10.1016/j.jclepro.2022.132375>
- Parida, P., Behera, A., & Chandra Mishra, S. (2012). Classification of Biomaterials used in Medicine. *International Journal of Advances in Applied Sciences*, 1(3), 125–129. <https://doi.org/10.11591/ijaas.v1i3.882>
- Peng, P.-W., Ou, K.-L., Lin, H.-C., Pan, Y.-N., & Wang, C.-H. (2010). Effect of electrical-discharging on formation of nanoporous biocompatible layer on titanium. *Journal of Alloys and Compounds*, 492(1–2), 625–630. <https://doi.org/10.1016/j.jallcom.2009.11.197>
- Prakash, C., Kansal, H. K., Pabla, B., Puri, S., & Aggarwal, A. (2016). Electric discharge machining – A potential choice for surface modification of metallic implants for orthopedic applications: A review. *Proceedings of the Institution of Mechanical Engineers, Part B: Journal of Engineering Manufacture*, 230(2), 331–353. <https://doi.org/10.1177/0954405415579113>
- Prakash, C., Kansal, H. K., Pabla, B. S., & Puri, S. (2015). Processing and Characterization of Novel Biomimetic Nanoporous Bioceramic Surface on β -Ti Implant by Powder Mixed Electric Discharge Machining. *Journal of Materials Engineering and Performance*, 24(9), 3622–3633. <https://doi.org/10.1007/s11665-015-1619-6>
- Prakash, C., Kansal, H. K., Pabla, B. S., & Puri, S. (2016a). Effect of Surface Nanoporosities Fabricated by Powder Mixed Electric Discharge Machining on Bone-Implant Interface: An Experimental and Finite Element Study. *Nanoscience and Nanotechnology Letters*, 8(10), 815–826. <https://doi.org/10.1166/nnl.2016.2255>

- Prakash, C., Kansal, H. K., Pabla, B. S., & Puri, S. (2016b). Powder Mixed Electric Discharge Machining: An Innovative Surface Modification Technique to Enhance Fatigue Performance and Bioactivity of β -Ti Implant for Orthopedics Application. *Journal of Computing and Information Science in Engineering*, 16(4), 041006. <https://doi.org/10.1115/1.4033901>
- Prakash, C., Kansal, H. K., Pabla, B. S., & Puri, S. (2017). Experimental investigations in powder mixed electric discharge machining of Ti–35Nb–7Ta–5Zr β -titanium alloy. *Materials and Manufacturing Processes*, 32(3), 274–285. <https://doi.org/10.1080/10426914.2016.1198018>
- Prakash, C., Kotecha, K., Pramanik, A., Basak, A., & Shankar, S. (2022). Synthesis of functionalized HAp-surface on β -Ti alloy using ball-burnishing assisted EDC process for biomedical application. *Archives of Civil and Mechanical Engineering*, 22(4). <https://doi.org/10.1007/s43452-022-00510-9>
- Prakash, C., Singh, S., Singh, M., Verma, K., Chaudhary, B., & Singh, S. (2018). Multi-objective particle swarm optimization of EDM parameters to deposit HA-coating on biodegradable Mg-alloy. *Vacuum*, 158, 180–190. <https://doi.org/10.1016/j.vacuum.2018.09.050>
- Prakash, C., & Uddin, M. S. (2017). Surface modification of β -phase Ti implant by hydroxyapatite mixed electric discharge machining to enhance the corrosion resistance and in-vitro bioactivity. *Surface and Coatings Technology*, 326, 134–145. <https://doi.org/10.1016/j.surfcoat.2017.07.040>
- Rack, H. J., & Qazi, J. I. (2006). Titanium alloys for biomedical applications. *Materials Science and Engineering: C*, 26(8), 1269–1277. <https://doi.org/10.1016/j.msec.2005.08.032>
- Riehle, M. O., Dalby, M. J., Johnstone, H., MacIntosh, A., & Affrossman, S. (2003). Cell behaviour of rat calvaria bone cells on surfaces with random nanometric features. *Materials Science and Engineering: C*, 23(3), 337–340. [https://doi.org/10.1016/S0928-4931\(02\)00282-5](https://doi.org/10.1016/S0928-4931(02)00282-5)
- Rupp, F., Scheideler, L., Olshanska, N., De Wild, M., Wieland, M., & Geis-Gerstorfer, J. (2006). Enhancing surface free energy and hydrophilicity through chemical modification of microstructured titanium implant surfaces. *Journal of*

Biomedical Materials Research Part A, 76A(2), 323–334.

<https://doi.org/10.1002/jbm.a.30518>

Sales, W. F., Oliveira, A. R. F., & Raslan, A. A. (2016). Titanium perovskite (CaTiO₃) formation in Ti6Al4V alloy using the electrical discharge machining process for biomedical applications. *Surface and Coatings Technology*, 307, 1011–1015. <https://doi.org/10.1016/j.surfcoat.2016.10.028>

Saruta, J., Sato, N., Ishijima, M., Okubo, T., Hirota, M., & Ogawa, T. (2019). Disproportionate Effect of Sub-Micron Topography on Osteoconductive Capability of Titanium. *International Journal of Molecular Sciences*, 20(16), 4027. <https://doi.org/10.3390/ijms20164027>

Schwartz, Z., & Boyan, B. D. (1994). Underlying mechanisms at the bone–biomaterial interface. *Journal of Cellular Biochemistry*, 56(3), 340–347. <https://doi.org/10.1002/jcb.240560310>

Setiawati, R., & Rahardjo, P. (2019). Bone Development and Growth. In H. Yang (Ed.), *Osteogenesis and Bone Regeneration*. IntechOpen. <https://doi.org/10.5772/intechopen.82452>

Sharma, D., Mohanty, S., & Das, A. K. (2020). Surface modification of titanium alloy using hBN powder mixed dielectric through micro-electric discharge machining. *Surface and Coatings Technology*, 381, 125157. <https://doi.org/10.1016/j.surfcoat.2019.125157>

Singh, V., Katyal, P., Kumar, K., & Kumar, R. (2022). Surface Integrity and Biological Response of Ti-Alloy Implants after Surface Modification. *Materials Today: Proceedings*, 56, 2451–2468. <https://doi.org/10.1016/j.matpr.2021.08.235>

Stich, T., Alagboso, F., Křenek, T., Kovářík, T., Alt, V., & Docheva, D. (2022). Implant-bone-interface: Reviewing the impact of titanium surface modifications on osteogenic processes in vitro and in vivo. *Bioengineering & Translational Medicine*, 7(1), e10239. <https://doi.org/10.1002/btm2.10239>

Sul, Y.-T., Johansson, C., Wennerberg, A., Cho, L.-R., Chang, B.-S., & Albrektsson, T. (2005). Optimum surface properties of oxidized implants for reinforcement of osseointegration: Surface chemistry, oxide thickness, porosity, roughness, and crystal structure. *The International Journal of Oral & Maxillofacial Implants*, 20(3), 349–359.

Takakubo, Y., Berce, A., Trebše, R., Tamaki, Y., Milošev, I., Al-Samadi, A., Tiainen, V.-M., Orton, & Konttinen, Y. T. (2013). Wear and corrosion in the loosening of total joint replacements (TJRs). In *Bio-Tribocorrosion in Biomaterials and Medical Implants* (pp. 74–110). Elsevier. <https://doi.org/10.1533/9780857098603.1.74>

Thakur, A., Rao, P. S., & Khan, Mohd. Y. (2021). Study and optimization of surface roughness parameter during electrical discharge machining of titanium alloy (Ti-6246). *Materials Today: Proceedings*, 44, 838–847. <https://doi.org/10.1016/j.matpr.2020.10.785>

Theisen, W., & Schuermann, A. (2004). Electro discharge machining of nickel–titanium shape memory alloys. *Materials Science and Engineering: A*, 378(1–2), 200–204. <https://doi.org/10.1016/j.msea.2003.09.115>

Tijo, D., Kumari, S., & Masanta, M. (2017). Hard and wear resistance TiC-composite coating on AISI 1020 steel using powder metallurgy tool by electro-discharge coating process. *Journal of the Brazilian Society of Mechanical Sciences and Engineering*, 39(11), 4719–4734. <https://doi.org/10.1007/s40430-017-0851-4>

Tyagi, R., Das, A. K., & Mandal, A. (2018). Electrical discharge coating using WS₂ and Cu powder mixture for solid lubrication and enhanced tribological performance. *Tribology International*, 120, 80–92. <https://doi.org/10.1016/j.triboint.2017.12.023>

Tyagi, R., Mandal, A., Das, A. K., Tripathi, A., Prakash, C., Campilho, R., & Saxena, K. K. (2022). Electrical Discharge Coating a Potential Surface Engineering Technique: A State of the Art. *Processes*, 10(10), 1971. <https://doi.org/10.3390/pr10101971>

Uddin, M., Hall, C., Santos, V., Visalakshan, R., Qian, G., & Vasilev, K. (2021). Synergistic effect of deep ball burnishing and HA coating on surface integrity, corrosion and immune response of biodegradable AZ31B Mg alloys. *Materials Science and Engineering: C*, 118, 111459. <https://doi.org/10.1016/j.msec.2020.111459>

Umar Farooq, M., Pervez Mughal, M., Ahmed, N., Ahmad Mufti, N., Al-Ahmari, A. M., & He, Y. (2020). On the Investigation of Surface Integrity of Ti6Al4V ELI Using Si-Mixed Electric Discharge Machining. *Materials*, 13(7), 1549. <https://doi.org/10.3390/ma13071549>

Umeda, H., Mano, T., Harada, K., Tarannum, F., & Ueyama, Y. (2017). Appearance of cell-adhesion factor in osteoblast proliferation and differentiation of apatite coating titanium by blast coating method. *Journal of Materials Science: Materials in Medicine*, 28(8), 112. <https://doi.org/10.1007/s10856-017-5913-8>

Uthayakumar, M., Raj, S. A., Ko, T. J., Kumaran, S. T., & Davim, J. P. (Eds.). (2019). *Handbook of Research on Green Engineering Techniques for Modern Manufacturing*: IGI Global. <https://doi.org/10.4018/978-1-5225-5445-5>

Velasco-Ortega, E., Jos, A., Cameán, A. M., Pato-Mourelo, J., & Segura-Egea, J. J. (2010). In vitro evaluation of cytotoxicity and genotoxicity of a commercial titanium alloy for dental implantology. *Mutation Research/Genetic Toxicology and Environmental Mutagenesis*, 702(1), 17–23. <https://doi.org/10.1016/j.mrgentox.2010.06.013>

Walma, D. A. C., & Yamada, K. M. (2020). The extracellular matrix in development. *Development*, 147(10), dev175596. <https://doi.org/10.1242/dev.175596>

Wandra, R., Prakash, C., & Singh, S. (2022). Investigation on surface roughness and hardness of β -Ti alloy by ball burnishing assisted electrical discharge cladding for bio-medical applications. *Materials Today: Proceedings*, 50, 848–854. <https://doi.org/10.1016/j.matpr.2021.06.075>

Ward, B. C., & Webster, T. J. (2007). Increased functions of osteoblasts on nanophase metals. *Materials Science and Engineering: C*, 27(3), 575–578. <https://doi.org/10.1016/j.msec.2006.05.032>

Williams, D. (2012). Concepts in biocompatibility: New biomaterials, new paradigms and new testing regimes. In *Biocompatibility and Performance of Medical Devices* (pp. 3–17). Elsevier. <https://doi.org/10.1533/9780857096456.1.1>

Xu, M., Li, C., Kurniawan, R., Park, G., Chen, J., & Ko, T. J. (2022). Study on surface integrity of titanium alloy machined by electrical discharge-assisted milling. *Journal of Materials Processing Technology*, 299, 117334. <https://doi.org/10.1016/j.jmatprotec.2021.117334>

Yan, B. H., Lin, Y. C., & Huang, F. Y. (2002). Surface modification of Al–Zn–Mg alloy by combined electrical discharge machining with ball burnish machining. *International Journal of Machine Tools and Manufacture*, 42(8), 925–934. [https://doi.org/10.1016/S0890-6955\(02\)00026-3](https://doi.org/10.1016/S0890-6955(02)00026-3)

- Yan, B. H., Wang, C. C., Chow, H. M., & Lin, Y. C. (2000). Feasibility study of rotary electrical discharge machining with ball burnishing for Al₂O₃/6061Al composite. *International Journal of Machine Tools and Manufacture*, 40(10), 1403–1421. [https://doi.org/10.1016/S0890-6955\(00\)00005-5](https://doi.org/10.1016/S0890-6955(00)00005-5)
- Yang, T.-S., Huang, M.-S., Wang, M.-S., Lin, M.-H., Tsai, M.-Y., & Wang Wang, P.-Y. (2013). Effect of Electrical Discharging on Formation of Nanoporous Biocompatible Layer on Ti-6Al-4V Alloys. *Implant Dentistry*, 22(4), 374–379. <https://doi.org/10.1097/ID.0b013e31829a170a>
- Zhao, G., Schwartz, Z., Wieland, M., Rupp, F., Geis-Gerstorfer, J., Cochran, D. L., & Boyan, B. D. (2005). High surface energy enhances cell response to titanium substrate microstructure. *Journal of Biomedical Materials Research Part A*, 74A(1), 49–58. <https://doi.org/10.1002/jbm.a.30320>
- Zhao, G., Zinger, O., Schwartz, Z., Wieland, M., Landolt, D., & Boyan, B. D. (2006). Osteoblast-like cells are sensitive to submicron-scale surface structure. *Clinical Oral Implants Research*, 17(3), 258–264. <https://doi.org/10.1111/j.1600-0501.2005.01195.x>
- Zou, R., Yu, Z., Li, W., Guo, M., & Li, J. (2016). Influence of porous structure on the machining performance of micro EDM. *Journal of Materials Processing Technology*, 232, 43–51. <https://doi.org/10.1016/j.jmatprotec.2016.01.027>

List of publications

Published:

Journal

1. Prakash, C., Wandra, R., Singh, S., Pramanik, A., Basak, A., Aggarwal, A., & Yadaiah, N. (2021). Synthesis of functionalized TiO₂-loaded HAp-coating by ball-burnishing assisted electric discharge cladding process. *Materials Letters*, 301, 130282

Conference Proceedings

2. Wandra, R., Prakash, C., & Singh, S. (2022). Investigation on surface roughness and hardness of β -Ti alloy by ball burnishing assisted electrical discharge cladding for bio-medical applications. *Materials Today: Proceedings*, 50, 848-854.
3. Wandra, R., Prakash, C., & Singh, S. (2022). Experimental investigation and optimization of surface roughness of β -Phase titanium alloy by ball burnishing assisted electrical discharge cladding for implant applications. *Materials Today: Proceedings*, 48, 975-980.