Role of Surface Modification on Bacterial Adhesion of Bio-Implant Materials

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Role of Surface Modification on Bacterial Adhesion of Bio-Implant Materials

Machining, Characterization, and Applications

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Preface

Orthopaedic joint implants such as hip, knee, shoulder, ankle, and elbow prosthesis are used for hard tissue replacement. These are load-bearing joints subjected to high levels of mechanical stress, fatigue, and wear in the normal daily activities of humans. The artificial implants used for these applications should possess structural integrity as well as surface compatibility with the surrounding biological environment for prolonged survivability without complications. Despite the adoption of advanced technology in manufacturing new implants and in surgical and medical management procedures, there are still a large number of implants that are subject to infection. The major causes for implants failure are a fracture, prosthetic dislocation, loosening, excessive wear rate at mating surfaces and its associated debris (although this cause has been much reduced by the use of cross-linked polyethylene), and pre-surgical contamination/infection (i.e. bacterial adhesion). In this book, a focus has been made on the implant failure due to bacterial infections, which is critically important from a clinical perspective when surfaces are made with a finishing process. Bacterial infections on implants are usually initiated through adhesion of bacteria to the implant surface by means of physiochemical interaction between the implant surface and bacteria, that is van der Waals forces, electrostatic forces, and gravitational forces. The reversible bacterial adhesion is followed by colonization and formation of biofilm that forms a layer of bacteria binding irreversibly to the implant surface, which is later difficult to remove from implants. The bacterial adhesion on the implant surface depends on the (1) properties of the implant material such as surface topography, surface roughness, surface chemistry, and surface energy; (2) properties of the bacteria such as surface charge, surface hydrophobicity/hydrophilicity, and appendages; and (3) properties of the surrounding environment such as type of antibiotics and associated flow condition, temperature, pH value, a period of exposure, chemical treatment, and

bacterial concentration. In particular, surface roughness and topography are the most influencing parameters because studies show that surface roughness increase the surface area of the material and also the depression, grooves, pits, scratches, and crevices in the rough surfaces influences the bacterial adhesion and acts as favourable sites for colonization and biofilm formation.

The near-net-shape of the bio-implants with the required surface finish is achieved by various finishing steps after the primary machining process. After the pre-fabrication, implants are subjected to the grinding process at the end. Final finishing of these surfaces is performed using polishing integrated with multi-axis CNC machines, vibratory abrasive polishing, free abrasive polishing, fixed abrasive polishing, belt polishing, and so on, which have all been comprehensively used in the past few decades to remove the finer irregularities and these techniques along with some of the advanced surface modification techniques are explained in this book.

Most of the implants have a complex and partly freeform surface that raises the difficulty in finishing at one step, and also one should have the knowledge of frequently changing contact conditions while performing finishing operation. There is a need for a technique that can be used to finish the complex surfaces of orthopaedic implants without changing its contact condition and without disturbing the dimensional accuracy. Abrasive flow finishing is one such process, in which flexible abrasive media is used to finish the complex surfaces. Previous studies have demonstrated that this process is capable of finishing complex internal and external surfaces and also freeform surfaces with controlled surface topography without disturbing the dimensional accuracy. Although there have been studies on the finishing of freeform surfaces using this process, understanding the biocompatibility of surfaces finished with this process is still far from the complete study.

In the present research work, to overcome the challenges associated with conventional finishing processes to finish complex internal and external surfaces along with free form surfaces, a unidirectional abrasive flow finishing (UAFF) process is developed in which abrasive media flows only in one direction to achieve the uniform lay patterns on the work surface in one direction with the required finish, which finds wide application in lubrication, biomedical, and surface engineering. The experimental setup mainly consists of the hydraulic power pack, a

hydraulic cylinder, an abrasive media cylinder, direction control valves, and pressure gauges.

In the present work, a viscoelastic polymer-based abrasive media is developed using a two-roll mill process. The selected constituents are viscoelastic polymer: silicone rubber (38%), plasticizer: silicone oil (12%) reinforced with suitable abrasive particles: SiC (50%). The developed abrasive media is subjected to different characterization studies such as the morphology using a high-resolution scanning electron microscope (HRSEM) – Inspect F50, thermal properties using thermo gravimetric analysis (TGA) – SDT Q600 instrument, and differential scanning calorimetry (DSC) – Model Q2000 V24 instrument. The tensile property of the abrasive media is measured using the Universal testing machine –ZwickRoell. Further, rheological properties such as viscosity, shear stress, storage modulus, loss modulus, loss tangent, and complex viscosity are ascertained using the Anton Paar Physica MCR 301 rheometer instrument. The rheological properties are studied at different temperatures: 25°C, 35°C, 45°C, and 55°C to study the behaviour of abrasive media under different temperature condition.

The experimentation has been carried out at various levels to investigate the performance of the developed unidirectional abrasive flow finishing setup. Machinability study has been carried out on different engineering materials having different hardness values such as aluminium, brass, copper, and mild steel machined with varying number of cycles. The objective of this experiment is to finish the biomaterials for different cycles and to study the effect of the UAFF process on the surface roughness, surface morphology, bearing area curve (BAC), and wettability of the biomaterial surfaces. The selected process parameters are number of cycles: 3, 6, 9, and pressure: 40 bar, 50 bar, and 60 bar. The obtained surface roughness and BAC are measured using optical profilometer – Wyko NT1100 instrument, and morphology of the surface is studied using an HRSEM image. Further, the wettability of the finished surface is studied by measuring the contact angle (θ) for three different liquids – water, formamide, and diiodomethane by sessile drop technique using contact angle Goniometer – GBX-Digidrop MCAT instrument. The measured contact angles are used to ascertain the surface free energy using the van-Oss-Chaudhury-Good equation. Further, response surface methodology (RSM) is used to optimize the input parameters such as a number of cycles and pressure to obtain the desired output responses, namely, average surface roughness (R_a), and material removed (MR).

Bacterial adhesion study has been carried out on machined biomaterials to investigate the influence of process parameters on surface roughness and bacterial adhesion. The developed setup has been used to finish the biomaterials – stainless steel (SS316L) and titanium alloy (Ti-6Al-4V) with two different abrasive media (220 mesh and 400 mesh) and a varying number of cycles (3 and 6). The purpose of this study is to evaluate the effect of this finishing process on enhancing surface characteristics such as surface roughness, surface morphology, and its role in initial bacterial adhesion. Optical profilometer and an HRSEM are used to examine the surface roughness and surface morphology of the finished surfaces. Further, for the bacterial adhesion study, both *Escherichia coli* (*E. coli*) and *Staphylococcus aureus* (*S. aureus*) are selected because these bacterial strains are commonly found on orthopaedic implant-related infections.

The procedure followed during the experiments and experimental findings is detailed in the book under various chapters with different headings along with an introduction chapter, which covers the different surface modification techniques.

Structure of the Book

Chapter 1 covers introduction to bio-implants, classifications of various implants, different materials used to manufacture implants, the reason for implant failures, failure due to bacterial adhesion and biofilm formation, and a broad literature survey.

A wide variety of established methods developed for implant surface modification such as mechanical, chemical, and thermal processes are summarized in **Chapter 2**. Further, this chapter mainly focuses on implant surface roughness and topographical features that influence implant failures. After obtaining a panoramic view on the topic through the literature survey, the objective and scope of the book are carried out, followed by the motivation.

Chapter 3 introduces the readers to different types of finishing processes, the importance of finishing processes, surface topography, and surface morphology. Different variants of abrasive flow machining, material removal mechanism, and process parameters are discussed in detail. A literature survey on various aspects of the abrasive flow machining, motivation, objectives, and methodology of the research work is presented in this chapter.

Chapter 4 contains the development and fabrication details of the experimental setup carried out at different stages. It also includes the information on materials and various fixtures developed in the present work to carry out the experimentation. This chapter also consists of a selection of materials and preparation steps followed in the development of abrasive media. This section also highlights the different characterization studies used to study the properties of the abrasive media and results.

Chapter 5 presents the results and analysis of the experimentation carried out in the present research work. Machinability study on different work materials has been carried out. Further, surface roughness and

morphology of the abrasive flow finished work surfaces on wettability, and bacterial adhesion is presented at the end of the chapter.

Chapter 6 includes conclusions drawn based on the experimental studies carried out in the present research work, and the scope for future work is highlighted. The contribution of the work is also listed in this chapter.

Abbreviations

AA	Aluminium alloy
AFF	Abrasive flow finishing
AFM	Abrasive flow machining
AISI	American Iron and Steel Institute
Al	Aluminium
Al₂O₃	Aluminium oxide
ANN	Artificial Neural Network
ANOVA	Analysis of variance
AOA NJRR	Australian Orthopedic Association National Joint Replacement Registry
ASTM	American Society for Testing and Materials
ATCC	American Type Culture Collection
B₄C	Boron carbide
BAC	Bearing area curve
CCD	Central composite design
C-D	Convergent-Divergent
CFAAFM	Centrifugal force assisted abrasive flow machining
CFD	Computational fluid dynamics
CLUAFF	Closed loop unidirectional abrasive flow finishing
CNC	Computer numerical control
Cu	Copper
DBG-AFF	Drill bit guided-abrasive flow finishing
DCV	Directional control valve
DDS	Data dependent systems
DI	Deionized water
DSC	Differential scanning calorimetry
<i>E. coli</i>	<i>Escherichia coli</i>
ECAFM	Electro-chemical aided abrasive flow machining
EDAX	Energy dispersive X-ray analysis

EDM	Electrical discharge machining
GA	Genetic algorithms
GMDH	Group method of data handling
GP	Gel point
GRA	Grey relational analysis
HB	Hardness Brinell
HRSEM	High resolution scanning electron microscope
ID	Internal diameter
LB	Lysogeny broth
MAFM	Magneto abrasive flow machining
MEMS	Micro-electro-mechanical-systems
MR	Material removed
MRAFF	Magneto rheological abrasive flow finishing
MRR	Material removal rate
MS	Mild steel
MWD	Molecular weight distribution
OD	Outer diameter
PII	Prosthetic implant infection
R-AFF	Rotational-abrasive flow finishing
R-MRAFF	Rotational magneto rheological abrasive flow finishing
RMS	Root mean square
RSM	Response surface methodology
<i>S. aureus</i>	<i>Staphylococcus aureus</i>
S/N	Signal-to-noise ratio
SEM	Scanning electron microscope
SiC	Silicon carbide
SR	Silicone rubber
SS	Stainless steel
TGA	Thermo gravimetric analysis
Ti	Titanium
UAAF	Ultrasonic assisted abrasive flow machining
UAFF	Unidirectional abrasive flow finishing
USA	United States of America
UV	Ultraviolet
V	Vanadium

Notations

$\dot{\gamma}$	Shear rate, 1/s
γ^-	Base component, mN/m
$\theta^\perp$	Contact angle – Perpendicular direction, deg
$\theta^\parallel$	Contact angle – Parallel direction, deg
γ^+	Acid component, mN/m
γ^{LW}	Lifshitz–van der Waals or dispersive component, mN/m
γ^{TOT}	Total surface tension of the liquid, mN/m
ω	Angular velocity, rad/s
δ	Oscillation phase angle, deg
τ	Shear stress, Pa
η	Viscosity, Pa.s
$\Delta\theta$	Change in contact angle, deg
η'	Dynamic viscosity (In-phase), Pa.s
η''	Dynamic viscosity (Out-of-phase), Pa.s
η^*	Complex viscosity, Pa.s
ϕ_0	Angular amplitude of oscillation
ρ_w, ρ_p, ρ_o	Density of abrasive particles, density of polymer and density of oil, kg/m ³
θ_D	Contact angle – Diiodmethane, deg
θ_F	Contact angle – Formamide, deg
ρ_m	Density of abrasive media, kg/m ³
ΔR_a	Change in average surface roughness, μm
η_t	Total efficiency
θ_w	Contact angle – Water, deg
$A1$	Area of peak, μm^2
$A2$	Area of valley, μm^2
A_a	Area of abrasive cylinder, m ²
A_f	Area of hydraulic cylinder piston end (forward direction), m

A_r	Area of hydraulic cylinder rod end (return direction), m
D	Diiodomethane
d_a	Diameter of abrasive media cylinder, m
d_w	Maximum internal diameter of the workpiece, m
F	Formamide
f	Frequency, Hz
F_a	Axial force, N
F_r	Radial force, N
G'	Storage modulus, Pa
G''	Loss modulus, Pa
h	Thickness of the sample or gap between two plates, mm
l_s	Stroke length, m
M	Number of peaks in the profile
M	Torque, N m
M_0	Torque amplitude, Nm
M_{r1}	Peak material component
M_{r2}	Valley material component
P	Power, kW
P_c	Peak count
p_h	Pressure at hydraulic cylinder, bar
P_{max}	Maximum operating pressure, bar
Q_{max}	Maximum media flow rate, kg/m ³
r	Radius of the plate, mm
R_a	Average surface roughness, μm
R_k	Core roughness depth, μm
R_{max}	Maximum roughness depth, μm
R_p	Maximum height of peaks, μm
R_{pk}	Reduced peak height, μm
R_q	Root-mean-square roughness, μm
R_t	Maximum height of the profile, μm
R_v	Maximum depth of valleys, μm
R_{vk}	Reduced valley depth, μm
R_z	Ten-point height, μm
S	Mean spacing of adjacent local peaks, μm
t	Time, s
V	Volume of working cylinder, m ³
v_a	Volume of the abrasive cylinder, m ³
v_{max}	Maximum media flow velocity, m/s
W	Water

w_a w_p w_o	Weight percentage of abrasive particles, wt% of polymer and wt% of oil
W_c	Weight of the constituent, kg
w_c	Weight percentage of each constituent
w_m	Total weight of abrasive media, kg
Δ_q	RMS slope of the profile
λ_q	RMS wave length, μm

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Introduction to Bio-Implants

1.1 INTRODUCTION

Bio-implants are artificial medical devices used to replace the missing biological system and the damaged biological system, and also to support or enhance the existing biological structure. These implants can be of many types, such as load-bearing implants used in orthopaedic applications, dental implants used in restoring the functionality and appearance of natural teeth, implants used in cardiovascular system, and many more. Metallic biomaterials are widely used among other materials, especially in load-bearing implants and internal fixation devices, due to their superior characteristics such as high tensile strength, high yield strength, resistance to fatigue loading, resistance to creep, high corrosion resistance, and biocompatibility. Most popular metallic biomaterials in use today are stainless steel, cobalt-chromium alloys, titanium, and titanium alloys. These materials are used in cardiovascular, orthopaedic, dentistry, craniofacial, and otorhinology implants. The implants are usually manufactured using milling, casting, forging, compression moulding, powder metallurgy, and rapid prototype techniques; further features are generated using drilling, electrical discharge machining (EDM), electron beam machining (EBM), laser beam machining (LBM), and ultrasonic machining (USM).

According to statistics in the United States, around 1.32 million implant-related infections are registered every year and amount to devastating

consequences that expose patients to high morbidity and mortality. The implant infections are mainly associated with bacterial adhesion and biofilm formation. Bacterial adhesion and biofilm formation, in turn, are influenced by several micro-environmental factors. Some of such micro-environmental factors listed in the literature are mass transport, surface charge, surface conditioning, hydrophobicity, surface topography, and surface roughness. This chapter emphasises on the different types of implants, materials classifications, and bacteria-related infections under various sections.

1.2 CLASSIFICATIONS OF BIO-IMPLANTS

Bio-implants are prostheses devices used to regularize physiological functions. They are made up of biosynthetic materials like collagen, and tissue-engineered products like artificial skin or tissues. Most bio-engineered products like cardiac pacemakers and orthopaedic artificial implants are also covered under bio-implants because they are implanted entirely in the patient's body. Bio-implants are mostly classified at a broader level such as orthopaedic, cardiovascular, dental, ophthalmic, and neurostimulation implants; these are listed in Table 1.1.

1.3 MATERIALS USED IN IMPLANTS

The materials used in the implants should be a biocompatible, corrosion resistance, and wear resistance; they should have excellent mechanical properties and better Osseo-integration; and should not create any effect on biological system/tissue (Mahajan and Sindhu, 2018). All the available materials on earth cannot be used as biomaterials. Researchers have developed plenty of materials that can be used as biomaterials and are continually working towards developing new biocompatible materials. Some of the factors affecting implant biomaterial are *chemical factors*: these include three basic types of corrosion: general, pitting, and crevice; *surface specific factors*: the events at the bone-implant interface can be divided into the behaviour of the implant material, the host response; *electrical factors*: physiochemical methods, morphologic methods, and biochemical methods; *mechanical factor*: modulus of elasticity, tensile or compressive forces, and elongation and metallurgical aspects. The biomaterials are classified under metals, ceramics, and polymers. Some of the major materials used in the implants are listed as follows:

TABLE 1.1 Broad classifications of bio-implants (Global Trends & Forecasts till 2017)

Implant system	Different types under each implants system
Cardiovascular Implants	<ul style="list-style-type: none"> • Pacing devices • Implantable cardiac pacemakers • Stents and related implants <ol style="list-style-type: none"> 1. Coronary stents 2. Peripheral stents 3. Stent-related implants • Structural cardiac implants • Heart valves • Tissue heart valves • Mechanical heart valves • Implantable heart monitors
Spinal Implants	<ul style="list-style-type: none"> • Thoracolumbar implants • Intervertebral spacers • Machined allograft spacers • Cervical implants • Implantable spinal stimulators
Orthopaedics and Trauma	<ul style="list-style-type: none"> • Reconstructive joint replacements • Knee replacement implants • Hip replacement implants • Extremities <ol style="list-style-type: none"> 1. Shoulder implants 2. Elbow replacements 3. Ankle implants 4. Other joint replacements fusion products

(Continued)

TABLE 1.1 (Cont.)

Implant system	Different types under each implants system
Dental Implants	<ul style="list-style-type: none">• Orthobiologics• Trauma implants• Sports medicine• Dental screw• Abutment
	<ul style="list-style-type: none">• Crown and bone graft
	<ul style="list-style-type: none">• Intraocular lens
	<ul style="list-style-type: none">• Glaucoma and other lenses
	<ul style="list-style-type: none">• Cortical stimulators
Neurostimulators Implants	<ul style="list-style-type: none">• Deep brain stimulators• Sacral nerve stimulators• Spinal cord stimulators• Vagus nerve stimulators

Metals: Titanium and its alloys are the metals of choice for dental and orthopaedic implants. Some of the commonly used metals are stainless steel, titanium and titanium alloys, gold, cobalt-chromium alloys, zirconium, niobium, tantalum, and so on.

Ceramics: Ceramics are non-organic, non-metallic, and non-polymeric materials manufactured by compacting and sintering at elevated temperatures. The entire implant may be made of ceramic, or this may be applied as a coat to a metallic core. Some of the ceramics biomaterials are aluminium oxide, zirconium oxide, hydroxyapatite, tricalcium phosphate, tetracalcium phosphate, calcium pyrophosphate, fluorapatite, brushite, carbon glass, pyrocarbon, and bioglass.

Polymers: Polymers are most commonly used for liners and small implants. These materials can be used alone or mixed with some other ceramics. Commonly used polymers are polyethylene, polyamide, ultra-high molecular weight polyethylene (UHMW-PE), polymethyl methacrylate,

polypropylene (PP), polytetrafluoro-ethylene, silicone rubber (SR), and polyurethane (Sykaras et al., 2000).

1.4 IMPLANT FAILURE – INFECTIONS

Orthopaedic joint implants such as hip, knee, shoulder, ankle, and elbow prosthesis are used for hard tissue replacement. These are load-bearing joints subjected to a high level of mechanical stress, fatigue, and wear in the normal daily activity of humans. The artificial implants used for these applications should possess structural integrity as well as surface compatibility with the surrounding biological environment for prolonged survivability without complications and damage (Choudhury et al., 2017; Lysaght and O'Loughlin, 2000; Navarro et al., 2008; Ribeiro et al., 2012). Most of the dental implants also fail due to poor primary stability, bacterial infection, manufacturing defect, and improper selection of surgical protocol. Despite the adoption of advanced technology in manufacturing these new implants and in surgical and medical management procedures, the number of revisions of total knee and hip arthroplasties keeps on increasing over the period (Moriarty et al., 2016).

A significant cause for the implant failure is a fracture, prosthetic dislocation, loosening, excessive wear rate at mating surfaces and its associated debris, and pre-surgical contamination/infection (i.e., bacterial adhesion). American Joint Replacement Registry annual report (2018) collected between 2012 and 2017 showed that 8.2% of 47,378 hip arthroplasties and 7.9% of 40,488 knee arthroplasties are due to the infection and inflammatory reactions. The National Joint Registry for England, Wales, Northern Ireland, and the Isle of Man surgical data of 2018 report shows that the overall 0.72% of hip replacement, 0.93% of knee replacement, and 6% of shoulder replacement results in implants-related infections. Similarly, it was reported in the Australian Orthopaedic Association National Joint Replacement Registry (AOA NJRR) annual report (2018) that, overall, less than 1% of revision of knee and hip arthroplasties accounts due to infection.

1.5 BACTERIAL ADHESION AND BIOFILM FORMATION

Bacterial infections on implants are usually initiated through the adhesion of bacteria to the implant surface by means of physiochemical interaction between the implant surface and bacteria, that is van der Waals forces, electrostatic forces, and gravitational forces (Arciola

et al., 2015; Koseki et al., 2014). The reversible bacterial adhesion is followed by colonization and formation of biofilm that forms a layer of bacteria binding irreversibly to the implant surface that is later difficult to remove from the implants (Chan et al., 2017; Koseki et al., 2014). These adhered bacteria further form a colony of the bacteria that later on forms a thick layer of the dense mass of the bacteria called biofilms.

The bacterial adhesion on the implant surface depends on the

1. Properties of the implant material such as surface topography, surface roughness, surface chemistry, and surface energy;
2. Properties of the bacteria such as surface charge, surface hydrophobicity (contact angle, $\theta > 90^\circ$)/hydrophilicity (contact angle, $\theta < 90^\circ$); and appendages; and
3. Properties of the surrounding environment, such as the type of antibiotics and its associated flow condition, temperature, pH value, a period of exposure, chemical treatment, and bacterial concentration. Figure 1.1 shows the schematic diagram of bacterial adhesion and the effects of implant material properties.

FIGURE 1.1 Schematic diagram of bacterial adhesion and the effect of implant material properties (Song et al., 2015).

1.6 EFFECT OF SURFACE ROUGHNESS ON IMPLANTS INFECTION AND WETTABILITY

In particular, surface roughness and topography are the most influencing parameters because studies show that surface roughness increases the surface area of the material, in turn, increases bacterial adhesion. Here, the focus has been made on the relation between the surface roughness and the bacterial adhesion, which is critically important from a clinical perspective when the work surfaces are finished with the finishing process. The depression, grooves, pits, scratches, and crevices in the rough surfaces influences the bacterial adhesion and acts as favourable sites for colonization and biofilm formation (An and Friedman, 2000; Barbour et al., 2007; Cox et al., 2017; Hocevar et al., 2014; Ribeiro et al., 2012; Wassmann et al., 2017; Yoda et al., 2014). These topographical features depend on the type of machining, finishing, coating, and surface treatment procedure followed. The near-net-shape of the bio-implants with the required surface finish is achieved by various finishing steps after the primary machining process. After the pre-fabrication, implants are subjected to the grinding process at the end. Final finishing of these surfaces is performed using either polishing integrated with multi-axis CNC machines or vibratory abrasive polishing, free abrasive polishing, fixed abrasive polishing, belt polishing, and so on. These processes have all been comprehensively used in the past decades (Bohinc et al., 2016; Cox et al., 2017; Kang and Fang, 2018; Turger et al., 2013) to remove the finer irregularities. It was estimated that polishing processes typically accounted for 10% to 15% of the total manufacturing cost and detailed review on different manufacturing and finishing processes used in implant manufacturing are detailed in a study conducted (Petare and Jain, 2018).

It is understood that, despite the adoption of advanced technology in manufacturing the implants and in surgical and medical management procedures, there are still a large number of implants that are subject to failure due to the surface finish achieved on the implant surfaces. The major causes for implants failure are a fracture, prosthetic dislocation, loosening, excessive wear rate at mating surfaces and its associated debris, and pre-surgical contamination/infection (i.e., bacterial adhesion). Surface roughness is one of the factors that influence the bacterial adhesion on implants and acts as favourable sites for colonization and biofilm formation leading to prosthetic implant infections (PIIs). A further effect of surface roughness on the wettability of surfaces is explained subsequently.

Wettability of the surface is strongly influenced by surface energy, surface roughness, and liquid surface tension. Among these, the surface roughness of any work surfaces is the most critical parameter considered in many applications and processes, where the wetting and the adhesion play a significant role. Some of the critical applications are lubrication, implants, pharmaceutical manufacturing, oil recovery, pharmacy, adhesive, coating, semiconductor, spray quenching, textiles, paper printing, polymers, dental, optics, medical field, preservation of the building, biology, cosmetics, high performance microfluidic devices, self-cleaning surfaces, and biomedical products (Guo et al., 2014; Gui et al., 2018).

Wettability is the ability of the solid surfaces to be wetted when it is in contact with any liquids by reducing liquid surface tension such that the liquid spreads over the surface and wets it. It can be described quantitatively by measuring the tangential contact angle between a liquid droplet and a solid surface-air interface. If the contact angle (θ) is less than 90° ($\theta < 90^\circ$ = hydrophilic surface), it shows that the surface has high wettability characteristics; and if it is greater than 90° ($\theta > 90^\circ$ = hydrophobic surface), it shows low wettability (Law and Zhao, 2016). Figure 1.2 shows the schematic nature of contact angle for different work surfaces (Nuraje et al., 2013).

There are many well-developed processes to modify surfaces, such as the following:

- 1. Non-deforming methods – polishing, machining, acid etching, and anodizing.
- 2. Deforming methods – sandblasting, shot peening, and surface mechanical attrition treatment.

FIGURE 1.2 Nature of contact angle for different work surfaces.

The above methods are presently in use to control the roughness and wettability of metal surfaces. All these treatments control the roughness and wettability of surfaces but alter the strength of materials due to the formation of residual stresses and fine grains (Arifvianto et al., 2012).

Wenzel (1936) was the first to describe the wetting concept in a solid substrate having rough surfaces using a thermodynamic argument. The tilting plate principle was used to measure the contact angle, and he stated that the wetting properties of a solid substrate are directly proportional to the roughness of the surface wetted. Similarly, Cassie and Baxter (1944) extended the research on the analysis of wetting of porous surfaces. It is reported that when a liquid wets the porous surfaces, air pockets are formed between surface and liquid, and the interface becomes a composite surface. Based on the earlier-mentioned models, many researchers have studied the wetting of rough surfaces experimentally and analytically. Kubiak et al. (2011) established a mechanism of wettability versus surface roughness of different engineering surfaces. In the study conducted by Kubiak et al. (2011), several engineering materials such as aluminium alloy: AA7064; titanium alloy: Ti-6Al-4V; steel: AISI 8630; copper alloy, and ceramic and poly-methylmethacrylate surfaces are polished with a different process, and four different roughness values are obtained. Further, these surfaces are subjected to contact angle measurement, and conclusions are drawn based on the experimentation. It is observed that roughness has a strong influence on the wettability of engineering surfaces, and the same has been found on different materials also. Fischer et al. (2014) carried out the wettability study on the steel surfaces and reported that the directional spreading and the wettability of the surfaces are strongly influenced by anisotropic surface properties, and droplet takes place along the grooves, which leads to droplet elongation in one direction.

Li et al. (2015) experimentally identified the wetting of irregularly micro-structured surfaces. The silicon surfaces are machined with rubbing, grinding, and polishing with a 0.22 μm –3.58 μm depth, and wetting properties were investigated. This study concluded that the contact angle increases with an increase in the roughness factor and the aspect ratio on the irregular micro-structured surfaces. Also, anisotropic wetting properties were observed on regularly micro-grooved surfaces but not on the irregularly micro-grooved surface. Liang et al. (2015) conducted the wettability experiments on the aluminium alloy – AA6061 and nylon GF30 polished with ten different grit sandpapers to identify anisotropic

wetting characteristics versus roughness on machined surfaces. It was concluded that roughness has a significant influence on anisotropic wetting on machined surfaces. As the roughness decreases, anisotropic wetting reduces. In the direction perpendicular to lay energy, a barrier exists, which is the main reason for anisotropic wetting. Gui et al. (2018) carried out the wettability experiments on titanium-coated textured surfaces. This study helps in understanding the anisotropic wettability of the surfaces and its applications in different areas, including orthopaedic titanium implants.

From the above literature, it is concluded that the wetting of the surface finds wide applications in many areas. Surface roughness has a great influence on the wettability of the surfaces; especially roughness range and topographical features (i.e., lay direction, width, depth, etc.) decide the anisotropic wetting behaviour of the surfaces. Further study in this direction on work surfaces finished with a different technique and how these surfaces affect the wetting is essential to understand.

1.7 SUMMARY

This chapter gives a glimpse of different types of implants and materials used in the manufacturing of implants, and also implants' failures are explained in detail. Bacterial infections on the implants are explained along with the biofilm formation on the implants. Further, broad literature on the effect of surface roughness on the bacterial infection and wettability of the implants materials surfaces are elaborated in this chapter.

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