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Advances in sustainable grinding of different types of the titanium biomaterials for medical applications: A review

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ABSTRACT

This review discusses various grades of titanium biomaterials and their sustainable grindability for application in the medical field. Titanium biomaterials are most commonly utilized for medical applications due to their exceptional characteristics such as high corrosion resistance and biocompatibility. The presented review looks at the principal requirements of titanium for medical applications, such as some good mechanical properties, biocompatibility, corrosion, wear resistance properties, and processability that facilitate the successful implantation of implants. It discusses the various types of titanium alloys that are commercially available and, more specifically, used for medical applications. It highlights the properties of different grades of titanium alloys and further narrows down its primary focus on applications, advantages, and shortcomings of commercially available titanium biomaterials. Machining titanium alloys is a difficult task due to their inherent properties such as low thermal conductivity and chemical reactivity at high temperatures and usually results in changes in metallurgy and surface integrity at the machined surface. Conventional machining, which has been the main machining method, has some limitations related to environmental hazards, cutting fluid costs, and operator health issues that have necessitated the development of sustainable machining. The emphasis in this review has been placed on sustainable grinding techniques such as MQL machining, cryogenic machining, nano-particle MQL machining, high-pressure machining, and solid lubrication machining used to grind titanium alloys and their benefits and limitations. Finally, the review will highlight some of the potential areas for future research and trends on different cooling and lubrication methods in the sustainable grinding of titanium alloys for medical applications. It is believed that this review will be of great benefit to the industries involved in manufacturing titanium-based medical implants.

Introduction

For centuries, human beings have used biomedical devices to correct, augment, replace, or restore lost or malfunctioning biological structures [1–4]. As a person grows older and the number of ailments increases, there is a growing demand for implants to replace hard tissues [1,5]. Biomaterials are materials that are used to repair or replaced damaged, compromised, or degenerated body parts [6] without causing any negative side effects or damage to the body parts [7]. They have had a huge impact on the enhancement of the quality of life for many patients around the globe [1] through functional restoration engineering of different body tissues [6].

Modern implantable biomaterials have improved mechanical, biological, and chemical properties and can be tailored for a specific medical application and a specific patient. They can be made of metallic alloys, composites, polymers, or ceramics, but the majority are made of metallic alloys, which are used to replace damaged hard tissues [8,9]. Modern metallic biomaterials used for making medical implants include cobalt alloys, titanium and its alloys, magnesium, tantalum or niobium alloys, medical-grade stainless steel, and gold alloys [4]. In most medical applications, titanium alloys are becoming the material of choice [6] and comprise 70–80% of the materials commonly used in making orthopaedic implants [10].

Due to their resistance to high-temperature effects and stress-

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resistance properties, titanium alloys were originally developed in the 1940s for use in aeronautical applications [11]. Because of their unique properties, they have also found applications in marine engineering, chemical processing, medical industries, pharmaceutical manufacturing, and many other industries. They are amongst the most useful for biomedical applications due to their excellent mechanical, chemical, and physical properties, which allow for successful implantation [12]. They are very vital in the manufacturing of medical implants [5] due to their favourable features such as higher biocompatibility, better bio-corrosion and wear resistance in the body fluids [11,13], a high strength-density ratio [14,15], high mechanical strength and low Young's modulus compared to other biomaterials [1].

Different varieties of medical implants have been developed from Ti and its alloys. Some of them include cardiovascular devices (such as pacemakers and mechanical heart valves), orthopaedic implants (such as hip implants), dental implants, stents in blood vessels, external prostheses, and operational devices (plates, pins, and hip joints) [3,16,17]. Dental devices such as dental implants, orthodontic brackets and wires, endodontic files, and prosthetic appliances have been manufactured using Ti alloys [18].

Although some fundamental information about biomaterials is known, the understanding of the behaviour of most titanium biomaterials is still limited. Therefore, the review aims to discuss different types of titanium alloys currently used in biomedical applications and their properties in relation to their performance. Despite the extensive knowledge about the grinding of titanium alloys, the potential for sustainable grinding has not yet been fully explored for most titanium alloys. Hence, the main motivation of the review is to present an updated overview of the sustainable grinding of titanium alloys used in medical device manufacturing.

This review starts with an introduction and description of the requirements and considerations of titanium alloys for medical applications. Secondly, the general characteristics of the currently used titanium alloys are described. The third part describes the sustainable grinding of titanium alloys, where different cooling and lubrication methods are highlighted. The advantages, limitations, and research gaps of these different sustainable techniques for grinding titanium biomaterials are highlighted. Finally, a conclusion on the main aspects presented previously and future recommendations are discussed.

Considerations for selecting Ti-based alloys for medical applications

A biomaterial's properties for body implant applications must be assessed to determine its suitability to avoid rejection in the human body. However, the location of the implant and the medical history of the patient may change the properties of an ideal biomaterial [1]. Some of the principal requirements stated below are discussed in detail in the subsections:

(i) Biocompatibility
(ii) Mechanical properties
(iii) Corrosion resistance
(iv) Wear resistance
(v) Osteointegration
(vi) Processability

Biocompatibility

The capacity of a biomedical device to perform its function adequately over time without generating any unfavourable local or systemic effects at the site of implantation or on nearby tissues or organs is referred to as biocompatibility [3,19,20]. It is determined by the implant material's properties and mechanical design [21]. Overall biocompatibilities, including tissue-, cyto-, and haemo-compatibilities, are often evaluated using cell markers, histological sections, and metabolite measurements. The implant materials should be chemically inert, biocompatible with the human body, and should not produce toxic and unwanted immunogenicity. They should also not produce allergic reactions once implanted [22]. The biocompatibility of any implant material depends on how the human body reacts to it when it is implanted, and this reaction determines the success of the implantation process [19]. The problems associated with biocompatibility are thrombosis and the development of fibrous tissues around biomaterials implanted in the body [23].

Because no biomaterial is fully inert, some bodily reactions to implants must be measured and factored into the implant's material selection, design, and use [24]. The human body's reaction to the implant and the implant's deterioration in the body after implantation are two major factors that influence the implant's biocompatibility [23,25]. There are four types of implants based on the body's response to the implant: bio-tolerant, bioinert, bioactive, and bio-reabsorbable materials [3,25,26]. Once a bio-tolerant material has been implanted in the body, there is the development of a thin fibrous tissue interface between it and the body tissue, i.e. there is indirect contact between bone and the implant during bone formation (distant osteogenesis) [23]. Polymethyl methacrylate (PMMA) and Co-Cr-Mo alloys are examples of bio-tolerant materials. Titanium and its alloys, partly stabilized zirconia, 316 L stainless steel, and aluminium oxide are bioinert materials that often integrate well into the bone. They have limited interaction with their surroundings. For example, during bone formation, bone and implant are in direct contact (contact osteogenesis) [23].

Bioactive materials interact with the body tissues and bones, causing direct chemical and/or biological connections between them and the bone tissues (bonding osteogenesis) [23]. Synthetic hydroxyapatite, glass ceramics, and bio-glass fall into this category [25]. Materials that dissolve slowly once implanted in the human body and are slowly replaced by developing human tissue and bones are referred to as bio-resorbable materials. Tricalcium phosphate, calcium carbonate, calcium oxide, gypsum, and polylactic–polyglycolic acid copolymers are common bioresorbable materials that have been used in the manufacture of implants during the last three decades [26].

Titanium is biocompatible because of its inertness and resistance to corrosion by all bodily fluids and tissues [1,11]. Titanium implants are immune to the body's rejection and have a high rate of physical attachment to the host tissues and bones [1]. Ti alloys are non-toxic, hence they are useful in making surgical implants and prosthetic devices [1]. Titanium and its alloys have greater biocompatibility because, in any oxidative environment, a thin and adhesive passive native TiO_2 layer forms spontaneously on their surfaces [5]. The presence of these spontaneous thin films, it was argued, plays a crucial role in Ti and its alloys' outstanding biocompatibility and corrosion resistance [23]. The titanium also has low electrical conductivity, which makes it biocompatible [27].

Mechanical properties

Biomaterials should have mechanical biocompatibility to ensure long-term implantation. Mechanical bio-compatibility generally refers to mechanical properties appropriate for the proper functioning of the implant once implanted in the body [1]. The mechanical properties of the biomaterials are important when selecting an appropriate implant for a specific application, and it is expected that those properties are almost similar to the mechanical properties of natural bone [23]. The mechanical properties determine how a biomaterial will respond to different related force and load conditions [2,28]. Regarding mechanical behaviours, the primary classes of clinical biomaterials include metals, ceramics, polymers, and composites of these biomaterials [2]. Standardized mechanical tests using standard specimens subjected to specific load conditions are used to determine the mechanical properties of the biomaterials, taking into consideration international standards such as ASTM (ASTM E8–04, ASTM E9, ASTM E92–17, ASTM 143, ASTM F1264, ASTM 1044, ASTM F2193, ASTM E2546–15) [2,29].

The mechanical properties considered when selecting a biomaterial for medical applications include tensile strength, hardness, Young's modulus, fatigue strength, and elongation [1,23,25]. These characteristics are influenced by the human body's components [30]. They depend on the microstructure formed during phase transformations, chemical compositions and element types, and processing and transformation techniques [2,30]. The degradation of biomaterials has been shown in many studies to have a substantial impact on the mechanical characteristics of biomaterials as well as biological functions due to interactions between body cells and biomaterials [2].

The hardness of the biomaterial is one of the vital engineering requirements considered in choosing implant materials. Machining processes such as grinding can improve the implant's hardness due to strain hardening on the surface. Hardness has a relationship with tribological behaviour, strength, toughness, and modulus of elasticity of the implant material [31]. It is a factor in determining the suitability of biomaterials for clinical use. In tribological behaviour, as hardness increases, the wear mechanism in implant materials decreases. It is a consideration that the hardness value of implant materials like Ti-6Al-4 V should be almost similar to the hardness value of bone (614–736 MPa) when selecting it for biomedical application [14].

The stiffness of the biomaterial is a material property measured by Young's modulus [26]. A human bone's Young's modulus varies from 4 to 30 GPa, depending on the type of bone and the measurement direction [25]. Because artificial implant materials are stiffer than human bone, the necessary stress is not transferred to the surrounding bone. This promotes the resorption of the bone around the implant, causing it to loosen. The "stress shielding effect" causes the death of natural bone cells due to the biomechanical incompatibility of the implants [25]. It is thus recommended that an implant material should have a low Young's modulus comparable to or closer to the natural human bone to avoid the loosening of implants. In comparison to other implant materials, Ti alloys, particularly β-titanium alloys, have a low Young's modulus of elasticity (50–120 GPa) [32]. This low modulus of elasticity of Ti alloy is advantageous biochemically because it generates a smaller stress shielding effect [1]. A smaller stress shield promotes faster and healthier bone regeneration and eliminates the need for revision surgery [33].

The implant should also have high mechanical strength (i.e., tensile and fatigue strength) in order to withstand the applied load and avoid fracture [34]. The implant's long-term success when subjected to repetitive cycle stress is determined by the material's fatigue strength. Fatigue strength depends on the composition of the alloy, thermo-mechanical processes, surface processes, finishing processes, and heat treatments. To have a longer service period in the body and avoid revision surgery, it is recommended that an implant material have high strength, such as tensile strength (1000 MPa).

Titanium-based alloys have sufficient fatigue strength to prevent implant fatigue and fractures [25] due to cyclic stresses. The density of titanium-based alloys is around 4700 kg.m^{-3}[5]. This is a good feature as this will reduce the weight of the implant, and this makes Ti alloy present a behaviour mechanic similar to human bones [1].

Corrosion resistance

The corrosion behaviour of biomaterials, especially metallic ones, is of significant significance because ions released from the implant by bodily fluids can cause biocompatibility complications such as thrombosis [32]. Corrosion is the decomposition of a material into its constituent atoms as a result of chemical reactions between the material and its surroundings [28]. It is a critical consideration for a biomaterial intended for use as an implant in the human body since implant biocompatibility is highly dependant on it [6,35]. The higher the rate of corrosion, the more toxic ions are produced in the body, and consequently, the more adverse effects lead to implant incompatibility with human bones and tissues [28].

Corrosion affects the service and life of the implant made of metal and its alloy implanted in the body [7]. The corrosion behaviour of materials used for implantation has a strong influence on foreign body reactions at the implantation site [36]. Many factors influence the corrosion resistance of any implant, including the material's grain size, surface roughness, chemical composition (including the degree of interstitial elements), texture, mechanical stresses, and manufacturing procedures [1,24,35,37]. The implant's excellent corrosion resistance inhibits the discharge of metallic ions into the body due to the corrosion mechanism [35]. The service life of an implant in the body is determined by its corrosion resistance. Corrosion of the implant appears to impair the material's final strength and fatigue life, leading to biomechanical failure of the implants if they corrode [37,38]. Biomedical devices can undergo different and distinct types of corrosion after implantation, such as uniform, pitting, fretting, crevice cracking, galvanic, stress cracking, and fatigue corrosion [28]. Corroded implants can release harmful metal ions, which have been linked to allergic reactions, carcinogenicity, local tissue toxicity, hypersensitivity, inflammation, and genotoxicity in humans [1,32].

Titanium and its alloys have higher corrosion resistance to human bodily fluids over time when compared to other metallic biomaterials [24]. They form a stable, protective oxide layer quickly due to their extremely high affinity for oxygen [39]. Surface modification of titanium alloys used in implant fabrication is a promising technique to overcome implant problems and improve the performance of implants [28]. Surface modification procedures include applying a thin uniform coating to the implant, forming a stable passivation oxide layer, ion beam processing, and surface texturing [28]. Surface modification of the titanium implant leads to the formation of a thin protective layer which improves the corrosion resistance.

Wear resistance

The wear behaviour of the biomaterials is of great concern because the discharge of wear debris from the implant may give rise to an adverse cellular response that leads to inflammation, pain, bone resorption, and the release of harmful enzymes [1]. Wear of the implants occurs due to articulation of the artificial joints and has an impact on the longevity of metallic implants in the body [34]. The wear resistance of an implant is amongst the major factors considered when selecting a biomaterial because the biocompatibility of implants depends greatly on it [35]. The wear mechanism limits the discharge of metallic ions into the body due to its high wear resistance [35]. Metallic ions can stimulate allergic reactions in the body while others are toxic, which can cause health issues and diseases [32]. Implant loosening and tissue responses are reduced when the implants have high wear resistance.

Titanium and its alloys have low resistance to wear, and this has limited their usage in the making of implants used in situations involving wear. Titanium has been shown to suffer considerable wear when brushed against itself or other materials [23]. This limitation has been mediated by adequate surface treatment of the titanium alloy using methods such as plasma nitriding and coatings such as physical vapour deposition (PVD) [7]. Coating of the implants with thin films such as titanium nitride coating (TiN) increases their wear resistance, corrosion resistance, and surface hardness and lengthens the service life of the implant [7].

Osseointegration

Osseointegration is the integration of an implant with bone cells and tissues after implantation [11,17]. It is the anchoring of the implant to the bone through the production of bone tissue at the bone-implant interface without troublesome fibrous tissue development. The implant's successful osseointegration ensures that it is safe and effective throughout its useful life [23,40]. The loosening of an implant can occur if the implant surface does not integrate adequately with the bone organs and other bodily tissues [23,40].

The surface integrity of any implant is critical to the implant's integration with the bodily tissues. Right surface roughness, surface chemistry, and surface topography ensure the proper development of good osseointegration [25]. Studies have demonstrated that chemical modifications of implant materials, such as surface coating of the implants, promote osseointegration [9,41,42]. The coated surfaces induce initial adsorption of selected proteins, such as vitronectin and fibronectin, promoting osteoblast adhesion and osseointegration [43,44]. The osteoblast-like cells may be stimulated to recognise the implant surface by the nano-topological titanium oxide layer, which can resemble the natural topography of native bone [42].

Surface roughness at the micro-scale of a biomaterial can improve its biological performance as it facilitates attachment of cells, proliferation, and differentiation of osteogenic cells [5,43]. It also determines the ground surface characteristics such as fatigue strength, service life, and the medical component's chemical stability [21,45,46]. The osseointegration parameters differ depending on the extent of surface roughness, such as macro-roughness, micro-roughness, and nano-roughness. While most research finds mean surface roughness (Ra) of 1–1.5 μm appropriate for bone formation, there is no agreement on the adequate roughness needed to obtain optimum osseointegration [47]. The production of toxic ions by implant materials can also affect osseointegration. Poor osseointegration may be detrimental to the regeneration of the local cells [48].

Titanium and its alloys have a high osseointegration rate, which implies they can efficiently fuse with the bone and, as a result, increase the implanted device's lifetime [5]. However, studies have shown that titanium alloys form a direct bond with the bone but not a chemical bond with bone tissue [23]. This limitation has been mediated by proper surface treatment of the implant using methods such as electrochemical deposition, physical vapour deposition, and dipping techniques [23]. Surface treatment of titanium alloy implants can help them perform several times better than their natural capacity.

Processability

The material characteristics of titanium biomaterials are largely determined by their microstructure. Thermo-mechanical processing and heat treatment operations influence the microstructures of titanium and its alloys [2,30]. Alpha and beta-titanium alloys have single microstructures [24]. Through manipulation of their microstructures, alpha-beta titanium alloys can be improved or their biomaterial properties customised for medical applications. Titanium can have equiaxed, transformed, Widmanstatten, and metastable beta microstructures depending on the heat treatment. As-worked structures have higher anisotropic tensile strengths but are less ductile. Recrystallized equiaxed structures possess lower strengths but are more ductile [24]. In summary, in selecting titanium alloys for the manufacturing of implants, the objective is to develop implants that can last a long time once implanted and ensure the patient lives a quality life. Hence, the considerations above formed the basis for the selection of titanium alloys.

General characteristics of titanium and its alloys

Structures, stabilizers, and types of titanium and its alloys

Titanium is a D-transition element and it can exist as commercially pure (CP) titanium and as an alloy [13]. CP titanium is 99.9% pure and is graded 1, 2, 3, or 4 based on the amounts of interstitial elements such as carbon, iron, oxygen, nitrogen, and hydrogen [49]. Each grade of CP titanium has different corrosion resistance, ductility, and strength [13]. CP titanium (unalloyed) exists in two crystallographic forms and they are a hexagonal closed-packed crystalline structure (HCP) and a body-centred cubic crystalline structure (BCC) [11,13]. It has an HCP structure at room temperature as well as up to 888 °C [11,50]. At a transformation temperature of 888 °C, CP titanium transforms from an HCP structure to a BCC structure that is stable until melting [5,6].

Titanium alloys with various properties result from the manipulation of the crystal structure through alloying of CP titanium with stabilizers and thermo-mechanical processes [11]. To optimize mechanical qualities, including tensile strength and wear resistance, alloying elements are added to CP titanium [44]. Based on their effects on the transformation temperature of titanium metal, alloying elements are classified as beta (β) stabilizers, alpha (α) stabilizers, and neutral stabilizers [51]. Oxygen, aluminium, germanium, carbon, nitrogen, and gallium, amongst other alpha stabilizers, dissolve efficiently in the α-phase, raising the transition temperature of titanium [50]. Aluminium is the principal alpha stabilizer in titanium alloys [39], and it strengthens the titanium by increasing its tensile strength, creep strength, and modulus of elasticity through solid solution strengthening mechanisms. The aluminium content used in alloying titanium is typically below 7% to avoid embrittlement [39].

Molybdenum, chromium, vanadium, and niobium are beta stabilizers that lower the titanium transition temperature by stabilizing the β-phase [13,50]. Molybdenum improves hardenability, making it an effective stabiliser in the design of β-type alloys [52]. It however reduce long-term, elevated-temperature strength. The addition of niobium to titanium is largely used to strengthen the alloy's high-temperature oxidation resistance. Neutral stabilizers do not affect transformation temperature and include zirconium, silicon, and tin [53]. Tin is highly soluble in both phases and is commonly used with aluminium as a solid solution strengthener to provide greater strength without embrittlement. With titanium, zirconium forms a continuous solid solution. It operates as a neutral element and is isomorphic with both the α and β phases [54]. It is utilized to boost strength in low and mid-temperature environments. The zirconium content used in alloying titanium should be below 6% to avoid a reduction in ductility and creep strength [39]. Other alloying elements, for example, nickel, iron, silicon, copper, and boron, are commonly added to titanium alloys to improve castability, mechanical strength, and chemical stability.

Based on their atomic crystal structures, titanium alloys are divided into four groups: alpha (α), near-alpha, alpha-beta (α + β), and beta (β) alloys [55,56]. The α-type Ti alloys have a principal component phase called the alpha phase, which has the crystal structure HCP [57]. The major alloying ingredients in the alpha alloy are iron and oxygen. They are more resistant to creep than beta alloys. They are used in cryogenic and high-temperature applications. They offer adequate strength, hardness, and weldability, but they are difficult to forge. A commercially produced alpha alloy is the single-phase -alloy Ti-Al-2.5Sn. Near-alpha alloys have a higher percentage of α-stabilizers than β-stabilizers. Ti-3Al-2.5 V, Ti-8Al-1Mo-1 V, and Ti-6Al-2Sn-4Zr-2Mo are some of the alloys in this group. These alloys are used to make components that operate at temperatures between 400 and 520 °C [11]. Alpha-beta alloy has α- and β-phases and the beta-phase may vary from ten to fifty per cent of the whole phase at room temperature. Ti-6Al-4 V, Ti-6Al-2Sn-4Zr-6Mo and Ti-6Al-2Sn are the most common alloys. These alloys' characteristics are influenced by heat treatment, which modifies the amounts and types of β-phase elements. Components made from α-β alloys can operate at temperatures ranging from 315 to 400 °C [39].

The beta alloys have β-stabilizers in their phases and include Ti–15V–3Cr–3Al–3Sn, Timetal 21S, Ti–3Al–8V–6Cr–4Mo 4Zr, and Ti–10V–2Fe–3Al alloys [14]. The beta phase, which has a BCC crystal structure, is the most common component phase in -type Ti alloys [57]. They can be forged easily across a larger range of forging temperatures [39]. They are easy to work with, have good stress corrosion resistance and can also be heat-treated to high strength. Because of their cold rolling characteristics, they are frequently utilized to produce sheets. Some β-alloys, such Ti–10V–2Fe–3Al, have outstanding fatigue properties, whilst others, like Ti–15V–3Cr–3Al–3Sn, have poor fatigue qualities when compared to their strengths.

Some grades/types of titanium biomaterials and some of their mechanical properties

There are different grades and types of titanium and alloys that exist. The designation of CP titanium is commonly known by ASTM grades [55]. Table 3.1 shows different types of titanium and its alloys utilized in medical applications, while Table 3.2 shows some of the mechanical properties of different grades of titanium and its alloys.

Titanium biomaterials in use, their applications, and limitations in biomedical applications

Titanium biomaterials are widely employed in medical applications due to their superior properties over other metallic biomaterials. Different types of titanium and their alloys used for medical applications are discussed in the following sections.

Commercially pure (CP) titanium

CP titanium available on the market can be categorised as grades 1, 2, 3, and 4 according to the ASTM B265 standard [63,64]. The grading determines the corrosion resistance, ductility, and strength of the CP titanium [13]. The lowest quantities of iron (Fe) and oxygen (O) in CP Titanium Grade 1 make it the most formable. Oxygen and iron content, as well as mechanical strength, increase progressively in grades 2, 3, and 4 [55]. In all four classes of CP titanium, corrosion resistance is almost similar. For applications requiring high corrosion resistance and good ductility, grade 2 is preferred compared to others [63]. CP titanium is used as an implant because it has more corrosion resistance and higher mechanical strength than Co-Cr and 316 L stainless steel [65]. Its tensile strength ranges from 240 to 550 MPa, its Young's modulus is 100 GPa, and its hardness is 70–100 HRB [24]. Due to its limited mechanical qualities, it is primarily employed in the production of dental implants.

Ti-6Al-4 V alloy

It is the most common titanium alloy and belongs to the category of α+β titanium alloys [38]. There are two commercially available grades of Ti-6Al-4V: grade 5 and grade 23 (ELI). Grade 5 was amongst the first generation of biomaterials to be used for making orthopaedic implants. In the Ti-6Al-4 V alloy, Al (5.5–6.1%) serves as an α-phase stabilizer and V (3.9–4.2%) serves as a β-phase stabilizer [5]. This alloy can be processed through mill-annealing, solution treatment, and ageing, and when used in its annealed state, its microstructure has a considerable impact on its mechanical properties [54].

Ti-6Al-4V-ELI (extra low interstitials) is an alloyed grade of titanium, also known as Ti grade 23 [66]. This alloy has fewer iron inclusions and more carefully regulated interstitial components of carbon and oxygen than standard Ti-6Al-4 V. It has a lower degree of interstitial impurities than standard Ti-6Al-4 V, which increases ductility and fracture toughness. It is found in the annealed condition as well as the beta annealed condition sometimes. Grade 23 has good corrosion resistance as well as great strength and toughness.

Table 3.1
Examples of Titanium biomaterials used for biomedical applications [14,58].

S/No	Materials	S/No	Materials
1	CP Ti	20	Ti-15Mo-3Nb
2	Ti-6Al-4 V (TAV)	21	Ti–35.3Nb–5.1Ta–7.1Zr
3	Ti-6Al-4 V ELI	22	Ti–15Mo–5Zr–3Al
4	Ti-Zr	23	Ti–29Nb–13Ta–4.6Zr
5	Ti-Ni	24	Ti–20Cr–0.2Si
6	Ti-13Nb-13Zr (TNZ)	25	Ti–13Cu–4.5Ni
7	Ti–6Al–7Nb (TAN)	26	Ti–25Pd–5Cr
8	Ti–15Zr– 4Nb–4Ta (Ti-15–4–4)	27	Ti-50Ta
9	Ti–5Al–2.5Fe	28	Ti-25Nb-11Sn
10	Ti–5Al–3Mo–4Zr	29	Ti-12Mo-5Zr
11	Ti–12Mo–6Zr–2Fe	30	Ti-39.3Nb-13.3Zr-10.7Ta
12	Ti–15Sn–4Nb–2Ta–0.2Pd	31	Ti-35Nb-4Sn
13	Ti–15Mo	32	Ti-29Nb-13Ta-2Sn
14	Ti–15Zr–4Nb–2Ta–0.2Pd	33	Ti-36Nb-2.2Ta-3.7Zr-0.3O
15	Ti–16Nb–10Hf	34	Ti-30Zr-3Cr-3Mo
16	Ti-25Nb-2Mo-4Sn	35	Ti-12Mo-3Nb
17	Ti-31.0Fe-9.0Sn	36	Ti-12Mo-5Ta
18	Ti-30Zr-3Cr-3Mo	37	Ti-29Nb-13Ta-4Mo
19	Ti-29Nb-13Ta-6Sn	38	Other medical Ti alloy(s)

Ti-6Al-4 V is a titanium alloy that is commonly used in biomedical applications such as orthopaedic implants [44]. Originally intended for aerospace applications, this alloy has found its way into the biomedical industry because of its excellent biocompatibility and corrosion resistance [54]. Its tensile strength is exceptional (it can reach up to 1100 MPa), it has a low specific density, low Young's modulus, high hardness of 36 HRC, a high fatigue strength, enhanced corrosion resistance, a high fracture toughness, and superior biocompatibility. It also has good weldability and creep resistance up to 300 °C [39]. These characteristics make this alloy an excellent choice for biomedical implants [25,54,67]. High-strength prosthetic implants [39], dental implants, cardiac implants, hip and knee prostheses, artificial knee joints and bone plates, etc. are made from Ti-6Al-4 V [38,68].

According to new research, implants consisting of Ti-6Al-4 V alloys have been related to long-term health difficulties like Alzheimer's disease, neuropathy, and osteomalacia [34]. The release of aluminium and vanadium ions from Ti-6Al-4 V implants is thought to be the cause of these complications [34]. Vanadium is a cytotoxic element [69], while neurotoxicity and neurodegenerative diseases have been associated with the presence of aluminium in titanium-aluminium-based alloys [70]. The modulus of the Ti-6Al-4 V alloy is higher than that of human bone tissue, and this is a matter of concern because bone resorption and implant failure can occur as a result of the stress-shielding effects induced by modulus mismatch [51].

Ti-6Al-7Nb (TAN)

In biomedical applications, Ti-6Al-7Nb (IMI-367) is a novel titanium alloy that is generally substituted for Ti-6Al-4 V alloy [71,72]. It was developed in the 1970s as an implant without vanadium to mitigate the cytotoxicity associated with vanadium [69]. This alloy is available in milled forms such as rods, bars, billets, extrusions, etc. [39] and is described by the ISO 5832–11 standard [59]. It is a ternary vanadium-free α+β titanium alloy used as an implant material and has improved biocompatibility, mechanical characteristics, and corrosion resistance [73].

Due to its microstructure, where Al stabilized the α-phase and Nb stabilized the β-phase, TAN alloy has a high strength that is practically identical to Ti-6Al-4 V [63,72]. It has a tensile strength of 862 MPa and an elongation of 10% less than the Ti-6Al-4 V alloy [73]. The niobium in TAN is non-toxic, and its Young's modulus is lower and closer to that of human bone tissue, reducing stress-shielding around the implant [71]. Its wear rate is comparable to that of CP-Ti but lower than that of Ti-6Al-4 V alloy [72]. Its resistance to corrosion and bio-tolerance is higher compared to Ti-6Al-4 V [50,63]. This alloy was developed as a wrought material for orthopaedic applications and has been investigated as a new alloy for whole hip prostheses [73]. It's a high-strength alloy with good biocompatibility, so it's employed as a surgical implant alloy [39] in dental implants and femoral stem prostheses [71].

Nickel-titanium (Ni-Ti) alloy

Nickel-titanium alloys, also known as nitinol, are shape memory alloys (SMA) that have been widely employed in a variety of applications, including biomedical, microelectromechanical systems, actuators, aerospace, and automotive devices [74]. Because of its remarkable features such as shape memory effect, super-elasticity, biocompatibility, and corrosion resistance, nitinol is the most extensively used SMA amongst other types of SMAs such as CuZnAl, and CuAlNi [75,76]. The shape memory effect is the process where an initially deformed low-temperature material recovers plastic strains through austenite-martensite phase transformation [75,77]. Due to its atomic

Table 3.2
Mechanical Properties of Some Medical grade Titanium biomaterials [13,24, 25,55,58–62].

S/ N0.	Grade	Name	Type of Crystal Structure	Tensile Strength (MPa)	Fatigue Strength (MPa)	Hardness	Young's Modulus (GPa)	Elongation (%)
Commercially Pure Titanium (CP Titanium)								
1	Grade 1	Titanium CP4	α-type	241		70 HRB	102.7	24
2	Grade 2	Titanium CP3	α-type	345		80 HRB	102.7	20
3	Grade 3	Titanium CP2	α-type	448		90 HRB	103.4	18
4	Grade 4	Titanium CP1	α-type	552		100 HRB	104.1	15
Titanium Alloys								
5	Grade 5	Ti-6Al-4V	α+β-type	931	620–725	36 HRC	110–114	6–10
6	Grade 6	Ti-5Al-2.5Sn	α-type	828–972	290	36 HRC	110	10–16
7	Grade 7	Ti-0.15Pd	α-type	485		150 HV	103	28
8	Grade 8	Ti-7.35Al-1Mo-1V	Near α-type	938		37 HRC	121	17
9	Grade 9	Ti-3Al-2.5V	α+β-type	690		24 HRC	102	21
10	Grade 10	Ti-11.5Mo-6Zr-4.5Sn	β-type	730	380	36 HRC	80	13
11	Grade 11	Ti-0.15Pd	α-type	345		120 BHN	103	37
12	Grade 12	Ti-0.3Mo-0.8Ni	Near α-type	483–607		180 HBN	103	18–22
13	Grade 13	Ti-0.5Ni-0.05Ru	α-type	275				24
14	Grade 14	Ti-0.5Ni-0.05Ru	α-type	410				20
15	Grade 15	Ti-0.5Ni-0.05Ru	α-type	484			106	19
16	Grade 16	Ti-0.06Pd	α-type	345–483		185 HRB	≥ 103	20–28
17	Grade 17	Ti-0.06Pd	α-type	241–345		120 HRB	≥ 103	24–37
18	Grade 18	Ti-3Al-2.5V-0.05Pd	Near α-type	621–740		15 HRC	91–107	15–17
19	Grade 19	Titanium Beta-C (Ti-8V-6Cr-4Mo-4Zr-3Al)	β-type	793	275	39 HRC	103	12
20	Grade 20	Ti-8V-6Cr-4Zr-4Mo-3Al-0.06Pd		795			105	8
21	Grade 21	Ti-15Mo-3Nb-3Al-0.2Si	β-type	793			100	15
22	Grade 23	Ti-6Al-4 V ELI	α+β-type	862	598–816	32 HRC	101–110	10–15
23	Grade 24	Ti-6Al-4V-0.06Pd	α+β-type	895				10
24	Grade 25	Ti-6Al-4V-0.5Ni-0.06Pd	α+β-type	895				10
25	Grade 26	Ti-0.1Ru	α-type	345				20
26	Grade 27	Ti-0.1Ru	α-type	300				24
27	Grade 28	Ti-3Al-2.5V-0.5Ru	Near α-type	625		25 HRC	107	15
28	Grade 29	Ti-6AL-4V-0.1Ru ELI	α+β-type	898			110	16
29	Grade 36	55Ti-45Nb	β-type	546			63	22
30		Ti-6Al-7Nb	α+β-type	900–1050	580–710		114	8.1–15
31		Ti-5Al-2.5Fe	α+β-type	1020	740–780		112	15
32		Ti–5Al–1.5B	α+β-type	925–1080	720–750		110	15–17
33		Ti–15Zr–4Nb–4Ta –0.2Pd	α+β-type	881			100	27
34		Ti-15Mo-5Zr-3Al	β-type	920	580		80	18–25
35		Ti-12Mo-6Zr-2Fe	β-type	1085	580–620		74–85	20
36		Ti-15Mo	β-type	874			78	21
37		Ti-15Mo-2.8Nb-0.2Si	β-type	990			83	16–18
38		Ti-13Nb-13Zr	β-type	1005	490–550		81	13
39		Ti-35Nb-7Zr-5Ta	β-type	597	260–300		55	19
40		Ti-35Nb-7Zr-5Ta-0.4O	β-type	1012			66	18
41		Ti–35.3Nb-5.1Ta–7.1Zr	β-type	596.7			55	19
42		Ti-16Nb-10Hf	β-type	852			81	11
43		Ti-29Nb-13Ta-4.6Zr	β-type	912			80	13
44		Ti-24Nb-4Zr-7.9Sn	β-type	830			46	15
45		Ti-35Nb-7Zr-5Ta-0.4O	β-type	1012			66	20
46		Ti–15Sn–4Nb–2Ta–0.2Pd	α+β-type	860			89	21
47		Ti–25Pd–5Cr	β-type	880		261 HV		5
48		Ti-55Ni	Intermetallic	875		190 HV		8
49		Ti-6Al-2Nb-1Ta-0.8Mo	α-type	860	490		110	12

Notes:.
The values stated may vary depending on the heat treatment subjected to the alloy.
For fatigue strength, the values stated depend on the number of cycles used.

structure, Ni-Ti can exist in two phases: austenite and martensite, depending on its temperature [78]. When the atomic lattice undergoes martensitic transformation between austenite and martensite phases, the space within it changes, allowing it to alter its shape. This behaviour is known as "super-elasticity" or "pseudo-elasticity" [77]. Super-elastic materials can regain their initial shape after the removal of the deformational stress, and they will subsequently recover their deformation strain [3]. This unique behaviour can be utilized for biomedical applications at varying temperatures [78,79].

Ni-Ti alloys have been identified as appropriate materials for medical implants due to their excellent biocompatibility, mechanical properties, and corrosion and wear resistance [80]. A protective Ti-based oxide surface layer on Ni-Ti products improves the corrosion resistance of the alloy. They feature a high strength-to-weight ratio with an ultimate tensile strength of 1240 MPa [24]. They also have an elastic modulus of roughly 48 GPa, which is virtually comparable to bone's elastic modulus [25,76,81], making them ideal for long-term orthopaedic implants [82]. The recoverable strain in Ni-Ti is roughly 8%, whereas it is more than 1% in bone [76,79]. The same behaviour of Ni-Ti and bone deformation in the body demonstrates the biomimetic behaviour of Ni-Ti implants under loading and unloading conditions [76]. The stress-strain curves in Fig. 3.1 depict the comparison amongst Ni-Ti, stainless steel, and human bone.

Because of the good qualities of the Ni-Ti alloys, they have been used for orthopaedics, cardiovascular, and orthodontic arch wires, guide wires, minimally invasive surgical instruments, and stents [83]. It has a shape-memory effect [34], which causes compressive stresses that help damaged bones knit together [79]. It has also been found to dilate blood vessels, hence its use as a self-expanding cardiovascular stent, thus increasing the flow of blood to vital organs [28,79]. Flexible self-expanding NiTi stents are currently used as a replacement for urethrotomies in current urethral strictures, keeping the biliary duct lumen open, minimizing dysphagia and helping in the restoration of swelling and nutritional intake for patients with malignant strictures [84,85]. The drawback of Ni-Ti is weak interfacial bonds for using these metallic implants in orthopaedic surgery [76]. In some studies [25], Ni concentrations above a certain level have also been observed to produce severe local tissue irritation, necrosis, and toxic reactions. The amount of Ni released by implants, on the other hand, is insufficient to elicit these effects [25].

Titanium-zirconium (TiZr) alloy

Because of the issues with the first generation of titanium alloys, new second-generation titanium alloys have been researched, manufactured, and brought onto the market. Beta stabilisers like zirconium (Zr), niobium (Nb), tin (Sn), palladium (Pd), tantalum (Ta), and indium (In) are used as alloying elements in these innovative Ti alloys and are

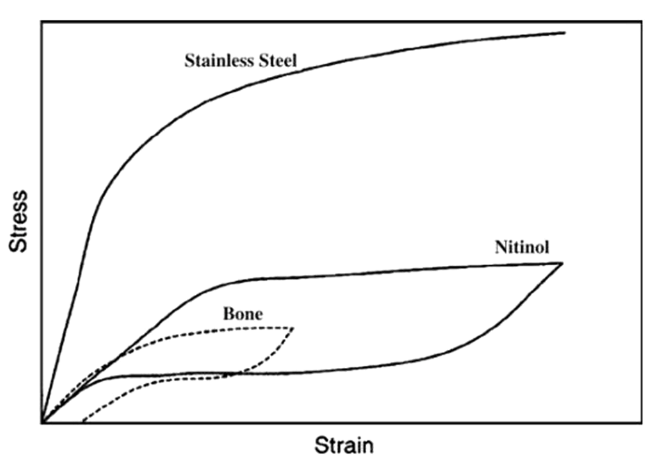

Fig. 3.1. Stress-strain curves for stainless steel, Ni-Ti and the human bone [76].

deemed extremely biocompatible and relatively harmless (i.e. non-toxic) when compared to aluminium and vanadium [6,56,72,86]. Titanium alloys in this category include binary titanium alloys, near β-type and β-type titanium alloys [57] and were developed to avoid the "stress shielding" phenomenon caused by the modulus difference between human bone and implant [56, 87]. These new varieties of titanium alloys include TiTa, TiHf, TiZr, TiNb [36], Ti-Nb-Zr, Ti-Mo-Zr-Fe, Ti-6Al-7Nb, Ti-Nb-Zr-Ta alloys etc. have been developed [71]. They offer excellent workability and mechanical properties and are alloyed with inert elements [71]. In terms of biocompatibility, corrosion resistance, modulus of elasticity, and wear resistance, innovative beta-type titanium alloys are regarded to be more favourable for biomaterial applications [25].

The TiZr alloy was created by Roxolid, Institut Straumann AG in Basel, Switzerland. TiZr is formed when zirconium is alloyed with titanium. From a metallurgical perspective, Zr is easier to alloy with Ti [36]. Originally, it was made by mixing Ti with 13–15% Zr. Ti and Zr are both transition metals that belong to the same periodic table group and have similar chemical characteristics. Zirconium is a biocompatible, bio-inert material and has high resistance to corrosion with a good biological response [88]. In TiZr, there is no limit to how much Zr can dissolve in the titanium matrix since the Zr element forms a complete solid solution with both α and β Ti phases [89]. Zirconium is added to Ti to enhance its mechanical strength, plasticity, and hardness and decrease its elongation, hence ensuring fine processing properties [72]. It lowers the fusion temperature of the titanium (1670 °C) as its amount increases, thus enhancing the castability process of titanium alloys [90]. When compared to other titanium alloys, most investigations have demonstrated that the Ti-Zr alloy has greater corrosion resistance and biocompatibility [91]. A binary TiZr alloy does not suffer from corrosion problems after long exposure to body fluids [88]. It has a tensile strength of more than twice that of the Ti alloy or even of Zr [89].

TiZr alloy is used for medical implants, especially as dental prostheses, although its use to its full potential is yet to be realised [72,92]. It was developed specifically for smaller diameter dental implants (≤3.5 mm) [72] because TiZr had increased fatigue strength and maintains the biocompatibility and osseointegration properties the same as for cpTi, the gold standard [72]. Saulacic et al. [93] explained that cpTi and TiZr implants have faster osseointegration than Ti6Al4V implants. Lee et al. [92] discovered that the Ti-Zr alloy composition influenced the *in vitro* biological response. They discovered that Ti-Zr with 30 mol% had the best physiologic response and the strongest strength.

Ti-13Nb-13Zr (TNZ)

Ti-13Nb-13Zr (TNZ) alloy is amongst various near β-type titanium alloys [38,94] and was developed by Davidson and Kovacs in the early 1990s [32]. It possesses a beta-phase stabiliser in the form of niobium, and zirconium is isomorphous with titanium's alpha and beta phases [95]. It consists of hcp martensite in water-quenched conditions and submicroscopic bcc beta precipitates as it ages [32]. It has a lower elastic modulus (65–79 GPa) [94], great biocompatibility, and strong corrosion resistance than standard grade 5 titanium alloy [96] due to its alloying components. It's a superior Ti alloy because both niobium, Nb and zirconium, Zr are recognized as non-toxic [94,97], biocompatible and non-allergenic elements with no adverse effects on the human body [54, 67]. Its tensile strength is estimated to be over 1300 MPa [95] and it has been proven that when Nb is alloyed with titanium at a specific amount, it lowers Young's modulus [37].

The American Society for Testing and Materials (ASTM) ASTM F1713–96 standard specifies the TNZ and it has already been certified for use as an orthopaedic implant material by ASTM and the US Food and Drug Administration (FDA) [94]. The TNZ alloy is an attractive alternative alloy for hard tissue implants due to the strengthening of the TiO_2 passive coating by Nb_2O_5 and ZrO_2 oxides [67]. Hard tissue implants made of Ti–13Nb–13Zr alloy are also intriguing because they have been established to have some antibacterial effect against

Gram-negative bacteria [67]. ZrO_2 is not cytotoxic, carcinogenic, teratogenic, or genotoxic oxide and it is harmful to different kinds of bacteria [98]. Because of Nb_2O_5's good lubricating qualities, titanium alloys with a high Nb concentration have better wear resistance [5]. Ti alloys with a high Nb content have a thin passive layer on the surface that lasts longer than Ti alloys with a low Nb content [25]. When compared to Ti-6Al-4 V alloy, Ti-13Nb-13Zr alloy has been established to have higher corrosion resistance [50].

According to Lee et al. [38], warm-rolled TZN has higher mechanical qualities in comparison to other types of TZN. It has a tensile stress of 1050 MPa, compared to 850 MPa for both hot rolled alloy and ASTM TZN [99]. It also has a higher hardness of HV 335±5 than the hot-rolled alloy and the solution-treated and cold-rolled alloy with hardness of HV 215±4 and HV 285±7, respectively [54]. It also has enhanced corrosion resistance, and it is attributed to its sub-microcrystalline structure [35].

Ti–15Zr– 4Nb–4Ta (Ti-15–4–4)

It is a titanium alloy created in Japan for long-term biomedical applications [100] and has been standardized in Japan with the Japanese Industrial Standard JIST 7401–4. Because the amount of released metallic ions into the medium for Ti, Zr, Ta, and Nb particle extractions is modest (0.3 mg/L), the alloy's constituents are biocompatible [51,59]. When compared to the Ti–6Al–4 V alloy, this alloy has a higher mechanical strength [101], a lower elastic modulus, and greater corrosion resistance due to the addition of Zr [64] and at 1×10^8 cycles, it has a fatigue strength of around 730 MPa [102]. This alloy is free of cytotoxic components and has a high ability to create apatite [103]. In comparison to Ti-6Al-4 V and Ti-6Al-7Nb, Ti-15–4–4 releases the fewest metal ions in various physiological solutions [104]. In their investigation, Okazaki [59] found that cells in Ti-15–4–4 grow faster than in Ti-6Al-4 V. Ti-15–4–4 has been successfully used to fabricate bone plates, artificial hip joints, and tooth implants [65]. Ti-15–4–4 is predicted to be the favoured titanium alloy for future biomedical applications because of its exceptional corrosion resistance [65].

Other medical-grade titanium alloys

Another titanium alloy that has also found application in biomedical industries is Ti-15Mo-5Zr-3Al [14]. It is used to make wires for sutures and implant fixation. This prevents galvanic corrosion when it comes into contact with other implant materials such as cobalt-base alloys and stainless steel [39]. Ti–6Al–2Nb–1Ta is an α+β type titanium alloy that has been created for advanced prosthetic hip joints for cementless types. In a neutral environment, it meets mechanical and anticorrosive requirements [105]. The Ti-15Mo-5Zr-3Al alloy is β type alloy intended for advanced prosthetic hip joints made of cement [58]. The α+β type Ti–6Al–2Nb–1Ta alloy has been designed for advanced artificial hip joints for cementless types [59]. In spinal fixation systems, Ti–17Mo has been examined and shown to be a feasible option for spinal rods [52]. The alloy demonstrated a changeable modulus of elasticity, prevented stress shielding effects, small spring-back, ductile and high tensile [52].

Grinding of titanium alloys

Machining is a technique for removing material that often involves cutting metals with a range of cutting tools. It is a vital process carried out to manufacture components to the desired dimensional accuracy to perform their functions satisfactorily [71]. It's a one-of-a-kind manufacturing technique in that it may be utilized to fabricate as well as used for finishing processes [106]. Turning, milling, shaping, grinding, drilling, and other machining techniques are only a few examples of the machining processes [107].

Grinding is an abrasive machining technique that involves removing material with abrasive grains [108,109]. Grinding abrasives are classified as conventional abrasives such as alumina or super-abrasives such as cubic boron nitride, and their usage in the grinding process is dictated by economic considerations. Grinding is used to manufacture precise and high-quality medical products with close tolerances [110,111]. It is the process used to machine implants from difficult-to-machine biomaterials such as titanium alloys to achieve a particular surface roughness that will promote osseointegration. Smooth surfaces are considered hygienic since they reduce the accumulation of plaque and bacteria on the implants [112]. The precision obtained through surface grinding is much better than with any other conventional machining [109]. In the grinding process, the removal of the material takes place through both shearing and ploughing modes [108,109,113].

Machining titanium alloys is the most difficult process amongst biomedical materials and usually results in metallurgical changes at the machined surface, rapid wear of the tools, and degradation of the surface integrity of machined parts [12]. These difficulties arise from titanium alloy's inherent properties, such as limited thermal conductivity, chemical reactivity at high temperatures, and so on [114]. They have a low thermal conductivity of approximately $6.7W/m.K$ which makes it difficult to conduct heat away from work-piece material [11]. At high temperatures (> 550 °C), they chemically react with most machining tools [11]. Their low elastic Young's modulus decreases rapidly during machining even at mild temperatures, causing deflection, chattering, and undesirable tool rubbing on the freshly generated surfaces. Other challenges are the maintenance of high hardness at elevated temperatures and rapid strain hardening during machining. The machining cost of Ti alloys exponentially rises when using high-speed machining, necessitated by the poor machinability of Ti alloys [115].

The grinding of titanium alloys generates a lot of heat in the grinding zone as a result of the poor heat conductivity of titanium alloys [113]. The heat produced when grinding degrades the surface integrity of the work and leads to the chemical reactivity of the titanium, hence causing rapid wear of the cutting tool [116,117]. Nandy et al. [118] explain that approximately 80 per cent of the heat produced during production is retained on the cutting tool, while the chips remove the remaining heat.

To mitigate against the unfavourable effects of high temperature and friction during the grinding of titanium alloys, suitable cutting fluids can be used in the grinding process as one of the remedies. Cutting fluids perform cooling, lubrication, and cleaning purposes during the machining processes. Cutting fluid should be used in an appropriate amount and the delivery method used during grinding determines its functionality and is a factor that influences the grindability of titanium alloys [119]. Depending on the cutting fluid delivery method, there are conventional and sustainable machining methods, which are discussed in detail in subsequent sections.

Conventional grinding

In conventional grinding, conventional grinding machines and conventional cutting fluids are mainly used. Cutting fluid is critical in the grinding process since it is responsible for maintaining the workpiece's surface integrity [110]. Flood cooling has a better surface finish than dry machining, although it falls short of MQL [63]. Many varieties of cutting fluids are available and vary in their application process, methods of treatment, and thermo-physical properties. Flood cooling is the oldest, most basic, and most widely used cooling and lubricating technology [63]. In the flood cooling lubrication method, a huge amount of cutting fluid is pumped into the grinding zone as a continuous feed flow [120]. Cutting fluids lubricate, cool, and remove chips at the tool-workpiece interface [121,122]. The flow rate of conventional cooling is in the range of 50–120 L/h [120].

In machining operations, cutting fluids are the most typical cause of environmental hazards [123,124] and human diseases [113]. Cutting fluid consumption is estimated to account for roughly 15% of all machining costs, with prices increasing by as much as 20%−30% when machining titanium alloys [113]. The costs include high purchase costs, which are factored into the cutting fluid disposal costs, cleaning costs, maintenance costs, depreciation costs, and personnel and health-associated costs [121]. As a consequence, cleaner cutting liquids such as vegetable oil-based cutting fluids including castor oil, sunflower oil, coconut oil, etc., have been developed [124]. Vegetable oils are derived from plants and are a viable source of renewable and environmentally friendly oils. Vegetable oils usually have excellent lubricating characteristics, such as excellent inherent lubricity, high viscosity index, low volatility, great lubricant additive solvency, high flash points, and easy mixing with other liquids [124,125]. They are biodegradable because the glycerine ester group can be hydrolysed easily and microorganisms can easily attack the double bond that is unsaturated in the chain [126]. The natural fatty acids, which make up vegetable oil, can enhance the biodegradation process.

Sustainable grinding

There are a variety of solutions for improving machinability and thereby lowering manufacturing costs, such as improving machining effectiveness, but amongst these, employing cooling and lubrication strategies receives the most attention. To ease the impacts on the economy and environment [127], some cooling techniques that are sustainable and environmentally friendly are currently being explored. In the manufacturing process of titanium-based implants, various stakeholders are advocating for a sustainable process. Sustainable manufacturing is a method of manufacturing products using techniques with minimal environmental side effects, preserving natural resources and energy, and being safe for people and economically feasible [128]. Sustainable manufacturing is one of three functional elements that compose sustainable development: sustainable goods, processes, and systems. It requires simultaneous consideration of social, economic, environmental, and safety aspects [129] during the production of medical implants. Sustainable products are environmentally friendly throughout their whole life cycle [121].

Process parameters considered in the fabrication of biomaterials, lubrication and cooling techniques, application of cutting fluids, machining tool material, work-piece material, optimum machining conditions, and tool geometry influenced the sustainability of the machining processes [123, 128, 130]. Due to the challenges associated with the cutting fluids, alternative methods of dissipating the heat generated at the machining interface are currently being explored.

Several methods can be applied to avoid or reduce the usage or control the amount of cutting fluids during the grinding of titanium alloys as a measure of sustainability. Dry machining, MQL, high-pressure cooling, cryogenic cooling, and cooling using nano-fluids and solid lubricants are examples of some green, sustainable, and environmentally friendly cooling and lubrication processes [131–133]. They've been proven to be better alternatives to traditional cooling approaches and have been established to improve the machining performance of titanium alloys [134]. These methods will be reviewed in detail as far as grinding titanium alloys is concerned. A few reviews on the grinding of titanium using sustainable cooling techniques necessitated this review.

Dry grinding (DG)

This is machining without the use of cutting fluids [135] and is the earliest grinding processing technique that is environmentally friendly [120]. Simply eliminating the use of cutting fluids in favour of DG and high-performance cutting tools would result in significant gains for sustainable technology [121]. All problems associated with the use of cutting fluids, such as contamination, disposal, and hazardous components, are eliminated with this machining procedure [135]. Hence, there is no air and water pollution and hence the cost of disposing of the cutting fluids is reduced. To extend the life of the cutting tool, DG is performed at lower cutting speeds and at a reduced production rate [135]. Previous studies on the dry grinding of titanium alloys are discussed in the following paragraphs.

Rai et al. [130] used an alumina wheel to grind grade 5 titanium alloy under various grinding pass counts and depths of cut (10 μm and 20 μm) in a dry grinding environment. Surface roughness, grinding ratios, hardness and visible surface burns were examined. The experimental results show that the surface finish is better at 10 μm than at 20 μm under the prevailing grinding settings. The hardness of the grade 5 titanium alloy decreased after grinding due to work softening caused by air cooling (dry environment). The grinding ratios increase when infeed values and the number of passes increase. Nosenko et al. [136] researched the grinding of VT3–1 titanium alloy in a dry grinding environment using a SiC grinding wheel. They established that grinding of titanium alloy without cutting fluids results in the production of a multi-layer structure with a thickness of 2 μm or more on the machined surface as a result of adhesive and cohesive processes on the machined surface. The experimental result of X-ray spectral microanalysis indicated that the mass concentration of atmospheric gases increases by more than 20 times. Banerjee et al. [137] performed grinding of grade 1 titanium under a dry grinding environment using an alumina wheel. The depth of grinding of 10 μm and 20 μm was utilized for the experiments, and grinding force, grinding ratios, surface roughness, ground chip and workpiece surface morphology were examined. The experimental findings show that the surface finish is better at 10 μm in comparison to at 20 μm. The morphology study of the chips and ground surface indicated the generation of high temperatures.

Despite its limitations, dry machining is a process that is more environmentally friendly [120] and reduces manufacturing costs as compared to machining involving coolants [138]. Researchers have made some attempts to make the application of dry machining viable and one of the attempts is the use of cutting tool materials with a coating [139]. Dry machining demands the use of modern cutting tool materials as well as the proper selection of cutting parameters. These cutting tools have the disadvantage of being extremely expensive, which raises machining costs [139].

Minimal quantity lubrication (MQL)

MQL is a machining process that combines the benefits of both wet cooling and dry machining [140]. MQL machining is a process that improves the machinability of materials, eliminates environmental damage, and reduces health problems faced by operators [11]. The MQL practice's environmental friendliness is aided by the use of vegetable oil-based cutting fluids [63]. The MQL system is made up of several components and includes an air compressor, fluid reservoir, flow regulator, air atomizing nozzle, and air pressure regulator [108]. Fig. 4.1 shows the schematic diagram of the MQL setup.

In the MQL process, compressed air atomizes a small amount of exceptionally efficient cutting fluid and propels it into an aerosol form to the cutting zone [141]. External and internal supply systems are the two main types of aerosol delivery systems used in MQL procedures [142]. The flow rate in MQL is often below 0.2 L/h as compared with 120–720 L/h of conventional coolant flow [11]. As a result, the MQL process is regarded as an effective, clean, and environmentally friendly way of improving machining results [63]. It also helps to reduce machining costs because just a little amount of cutting fluid is used, resulting in possible cost savings [63].

There is an improvement in machining performance in MQL machining as enough cutting fluid under pressure can penetrate the chip–tool interface. Because of the convection and evaporation modes of heat transfer in MQL, it has been demonstrated that MQL outperforms the traditional flood cooling technique [66]. Surface quality, cutting forces, machining temperature, and tool wear have all shown noticeable improvements in MQL machining. [143]. The limitation of MQL machining is that it produces abundant oil vapours in the air, which can threaten human life if the oil used is hazardous.

Research on the use of MQL in the grinding of titanium alloys has been carried out by various researchers. De Mello et al. [110] examined the surface grinding of Ti-6A-l4V alloy under three grinding environments (i.e. wet, MQL, and dry conditions) using a SiC grinding wheel. In comparison to dry and fluid grinding, they discovered that MQL grinding produced a better surface finish and caused less surface damage

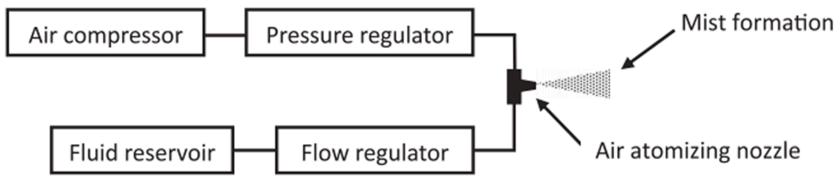

Fig. 4.1. Schematic diagram of the MQL setup [108].

due to effective cooling caused by good fluid penetration into the grinding action zone. According to Wojcik and Wejman [144], MQL grinding Ti-6Al-4 V and TIGER 5 titanium alloys with polypropylene glycol reduces cutting forces and limits thermal deformations on the top layer of thin objects. Mahata et al. [109] used soap water and a liquid CO_2 jet cooling technology to grind titanium alloys (Grade 1 and 5) using small quantity lubrication (SQL). They reported a considerable reduction in grinding forces and an improvement in surface finish under MQL conditions and liquid CO_2 compared to MQL conditions and an alkaline-based coolant.

Mukhopadhyay and Kundu [114] compared the results of dry environment, flood cooling and SQL method when grinding grade 5 titanium alloy using alumina wheel under identical process parameters. The results show that the SQL outperformed flood cooling and dry environment in terms of lowered grinding forces, reduced average surface roughness by roughly 58 per cent and 23 per cent, respectively, and enhanced ground surface morphology, according to their research. They also investigated the use of restricted quantity lubrication (RQL), flood cooling conditions, and a dry atmosphere in grinding grade 5 titanium alloy with an alumina wheel in another investigation [119]. They discovered that RQL was more successful at improving Ti-6Al-4 V grindability than flood cooling and a dry atmosphere.

Guo et al. [145] investigated surface grinding of titanium grade 5 (Ti6Al4V) alloy under three grinding environments (i.e. wet, MQL and dry condition) using a SiC grinding wheel. Due to excellent cooling generated by good fluid penetration into the grinding action zone, they discovered that MQL grinding gave a better surface finish with less surface damage than dry and fluid grinding. Sadeghi et al. [146] investigated the grinding of Ti-6Al-4 V alloy using synthetic and vegetable oil under MQL conditions. Grinding with synthetic oil yielded superior surface quality than with vegetable oils under MQL conditions, according to the findings. Better surface quality was achieved because the synthetic oils reduced friction force more effectively, had good lubricity, and could maintain higher integrity at higher temperatures than vegetable oil. Ronoh et al. [147] investigated the wet and MQL grinding of Ti-6Al-4 V alloys using sunflower oil, sunflower oil-based cutting fluids, and conventional coolants. MQL grinding with sunflower oil-based cutting fluids provided superior surface quality with less micro-hardness disturbance layer than other cooling methods, according to their findings.

Biswojyothi et al. [141] studied the grinding of Ti6Al4V under MQL parameters (coolant flow rate, coolant concentration, and air pressure) and found that the cutting forces and surface roughness showed a decreasing trend when air pressure, coolant concentration, and coolant flow rate were increased. When coolant concentration increased, the tendency of the coolant to form film increased due to higher viscosity and the energy formation decreased, resulting in a decrease in cutting forces. When air pressure increases, the chips from grinding are flushed away, hence maintaining a clean surface. This performance is credited to the ability of the coolant to reach the grinding zone in MQL grinding. According to the literature review in this section, the MQL method has shown considerable advances in the sustainable grinding of titanium and related alloys.

Nanoparticle jet MQL (NPMQL) grinding

When grinding difficult-to-cut materials, the MQL technique [148] may not be an appropriate alternative to flood cooling [149] due to the shortcomings of the lubricating medium. Researchers are looking into methods to increase the MQL system's capabilities, and the enhancements they're looking into are still within the realm of sustainable manufacturing. Fig. 4.2 shows a redesigned structure diagram with new classifications and prospective study areas [150].

Using certain cutting fluids, such as nano-fluids or nano-lubricants, to develop the wettability, convection, and conduction elements of the MQL technique is an emerging factor to be employed to enhance the cooling and lubrication functions in the MQL approach [151]. Solid nanoparticles are introduced to the cutting fluid to establish suspensions in the base fluid, such as water, lubricating oil, or glycol, in NPMQL grinding [108,148]. The pulverized nanoparticle lubricant with compressed air (0.60 MPa) is sprayed into the grinding area [120]. The most common nano-particles include the oxides (Aluminium oxide (Al_2O_3), Zirconium dioxide (ZrO_2), Copper (II) oxide (CuO)), molybdenum disulphide (MoS_2), graphene, silicon dioxide (SiO_2), metal nitrides, non-metal nitrides, carbide, and carbon nanotubes (CNT) and they have hardness and unique lubricating properties [108,142,151]. CNT has the lowest grinding temperature of all the nanofluids and consequently has the best heat transfer performance [148].

The nano-additives have different diameters that vary from 1 to 100 nm and their lengths range from 10 to 10×10^6 nm [108,151]. Due to the rolling movements of billions of nanofluids [148], the nano-particles utilized in the machining process act like ball bearings, lowering the cutting tool's rolling friction with the work material [142]. Fig. 4.3 shows the lubrication and cooling concepts of MQL nano-lubrication.

Cutting fluid consumption is approximately 70 mL/h per unit width of the grinding wheel in this technique [120]. Nano-cutting fluids have been used in a variety of cutting processes, including turning, milling, and grinding, and have shown promising cutting performance [140, 152]. MQL nano-cutting fluid has improved MQL performance in terms of good surface integrity, cutting force reduction, and tool wear, particularly when machining titanium alloys [151,153]. Because of their high thermal conductivity, they have shown a significant improvement in heat transfer during machining operations [63,153].

Despite their advantages in machining, nanoparticle applications have several limits in terms of production [142]. First, the nanoparticle concentration must be greater than 1% to have a substantial impact. Second, it contaminates and clogs the delivery system's pipe network, causing particle contamination and clogging. This is due to the increase in the fluid viscosity and subsequent changes in the flow dynamics in the internal system caused by the addition of nano-particles. The third is the issue of preparation of the fluid and maintenance, which require special procedures. Last is the emission of the aerosols with nanosized particles which cannot be filtered with conventional filtering methods.

Considerable research work has been devoted to the NPMQL grinding of titanium alloys. Setti et al. [108] investigated the MQL grinding of grade 5 titanium alloy with nano-cutting fluid (Al_2O_3 and CuO nanoparticles) and without nano-cutting fluid. They discovered that the MQL nano-fluid technology outperformed the traditional MQL technique in terms of tangential force reduction and coefficient of friction. Furthermore, the chip morphology analysis revealed excellent findings due to good lubrication and improved heat transfer of the nano-particles. Ibrahim et al. [154] investigated the role of graphene nanoplatelets (GNPs) lubricant film on the workpiece following MQL

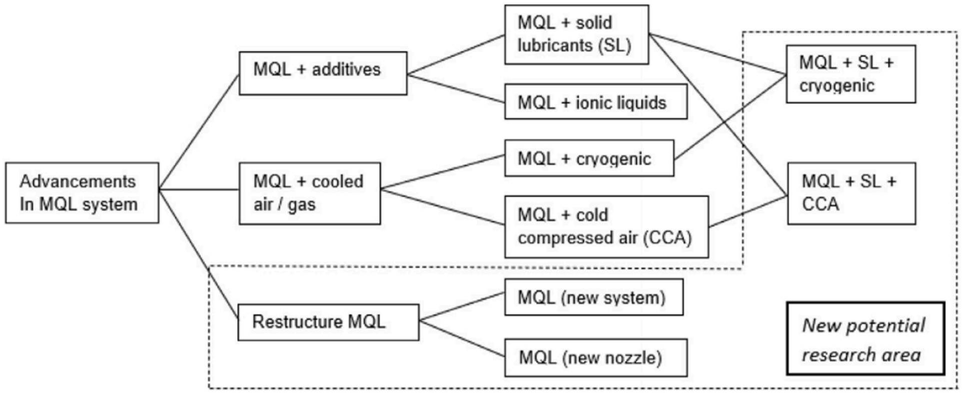

Fig. 4.2. Classifications of MQL advancements [150].

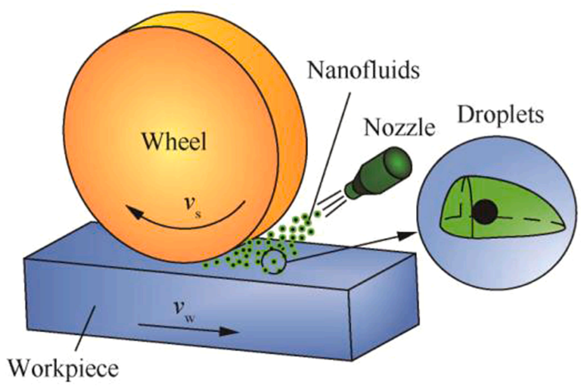

Fig. 4.3. Lubrication and cooling concepts of MQL nano-lubrication [120].

grinding and revealed that the graphene lubrication film functions as an antifriction and antiwear agent. Singh et al. [113] investigated the performance of graphene-assisted vegetable oil-based MQL while grinding TI-6AL-4V-ELI. The results revealed that the surface quality obtained while using graphene nanoparticles is very good with minimal surface roughness. Sarhan et al. [155] investigated the performance of MQL nano-lubrication with various levels of SiO_2 as an alternative to dry environment and flood grinding while grinding titanium alloy, and the results showed that MQL nano-lubrication worked well. Another study by Setti et al. [156] looked into the usage of a nanofluid in MQL mode to improve the grinding properties of Ti-6Al-4 V alloy. The MQL grinding process uses water-based Castrol cutting oil, water-based Al_2O_3 nanofluids, and pure water. With a larger concentration of Al_2O_3 nanoparticles, their results indicated that grinding with MQL nanoparticles reduces grinding forces and enhances surface smoothness.

Paul et al. [140] used MQL multi-walled carbon nanotube (MWCNT) and alumina nanofluid to grind Ti-6Al-4 V alloy. They discovered that the MQL MWCNT technique increased Ti-6Al-4 V grindability by reducing specific grinding forces and specific energy when compared to the traditional flood cooling technique. Because of its excellent thermal conductivity, MWCNT performed better as a coolant than Al_2O_3 nanofluids. However, as compared to the flooding process employing soluble oil, MQL Al_2O_3 nanofluid performed badly. Singh et al. [66] used the MQL technique to analyse the performance of different molybdenum disulfide (MoS2) nanoparticles, graphene, and graphite as additives in base canola soybean and olive oil for the surface grinding test of grade 23 titanium alloy. The effects of five different weight concentrations of MoS_2, graphene, and graphite nanoparticles on grinding performance were investigated. They discovered that MQL canola nano-fluids with 1.5 per cent graphene weight concentration produced the best outcomes. They also discovered that nanofluids in canola oil outperform MoS_2, graphite, and graphene in terms of performance.

Liao et al. [157] discovered that using the MQL of nano-modifier at a rate of 50 ml/h when grinding grade 5 titanium alloy with a vitrified bonded diamond grinding wheel resulted in lower grinding forces, lower grinding forces, and higher surface quality than using water-based cutting fluid. Bhargavi and Kumar [158] investigated the use of water-based Castrol cutting oil, water-based Al_2O_3 nanofluids, and pure water in MQL grinding of Ti-6A1–4 V. They found that with a larger concentration of Al_2O_3 nanoparticles, their results indicated that grinding with MQL nanoparticles reduces grinding forces and enhances surface smoothness. Liu et al. [152] studied the applications of water-based grinding fluid, pure synthetic lipids, and 2% Al_2O_3 nanofluid dispersed in pure synthetic lipids, through MQL in grinding Ti-6Al-4 V based on grey relational analysis. They noticed that using NPMQL gave them the best surface quality and rate of material removal. The literature evaluations in this part have revealed that by employing the MQL nano-lubrication technology, there are considerable improvements in the long-term grinding of titanium and its alloys.

Solid lubrication machining

In this process, solid lubricants are employed in the machining of titanium alloys in this process. Solid lubricants can significantly reduce heat generation in the grinding zone. This will aid in lowering the frictional components of the grinding force when grinding titanium alloys [159], resulting in enhanced titanium alloy grindability [160]. Because most cutting fluids are hazardous to one's health when used, solid lubricants can be used instead. The usage of solid lubricants has no negative effects on the environment [117].

There are different types of solid lubricants available on the market, for example, graphite, molybdenum disulfide, tungsten disulfide, hexagonal boron nitride (hBN), carbon nanotubes (CNT), SiO_2 and polytetrafluoroethylene, etc. [117,160,161]. The majority of solid lubricants have a hexagonal planar structure with strong covalent or ionic interactions between molecules and weak van der Waals forces between adjacent layers [117]. Due to its low shear strength, the planar structure slips with each other when a shear force is applied on any plane, resulting in the lubricity attribute [160].

Some studies have reported on the solid lubrication of titanium alloys during grinding and are presented in the following paragraphs. Malik et al. [161] investigated the use of solid powder lubricants in grinding grade 5 titanium alloy material using a SiC wheel. The usage of solid lubricants was found to be beneficial in boosting grinding performance in the study. In the research work done by Sahoo et al. [117], they investigated the performance of solid lubricants in grinding Ti-6Al-4 V using a SiC wheel. Solid lubricants have been demonstrated to be superior to dry grinding when it comes to reducing grinding forces by up to 20%.

In their work on the grindability of Ti-6Al-4 V alloy employing solid lubricants (graphite and MoS_2), Vemula and Khan [159] found that grinding forces and specific energy consumption were lowered by roughly 5% to 20%. They also stated that when grinding using MoS_2 rather than graphite, the decrease is greater. The studies reported in this section clearly show that there is a need for more research on the solid lubrication of titanium alloys during grinding.

Cryogenic machining (CM)

CM is a type of machining that uses cryogenic coolants rather than regular cutting fluids [162]. It entails a cutting procedure at a very low temperature [142]. According to Aggarwal et al. [163], cryogenic machining has several advantages when compared to machining using conventional cutting fluids. Being clean, safe and environmentally friendly, the CM process is considered a sustainable machining process [21,128]. The technique provides for better productivity by promoting a higher material removal rate while limiting the wear rate of tools and reducing the frequency of cutting tool changeovers during machining.

The cryogenic gases that are used as main coolants are nitrogen, carbon dioxide, and helium [11]. They are usually used in a liquid state and are harmless when they evaporate into the atmosphere [128]. This setup, depicted in Fig. 4.4, is designed to use liquid food-grade CO_2 as a coolant. Liquid nitrogen (LN_2) and liquefied carbon dioxide (CO_2) are usually used as the primary refrigerants to mainly dissipate the generated machining heat and enhance the machinability of the titanium alloys [115]. These two types of coolants are preferred for usage due to their extremely low temperature (approximately −183 °C), are environmentally friendly, and can prolong the life of the tool [115,123]. They have an effective heat absorption capacity, which can transfer away heat generated during machining. The LN_2 has a higher heat absorption capacity than liquefied CO_2, is less expensive, and is environmentally friendly, so it is frequently used as a coolant [11]. When machining hard metals, Pereira et al. [164] explain that CO_2 cooling is more suited and less expensive than other cryogenic gases.

LN_2 evaporates quickly into the environment in the cryogenic machining system. Hence, no residue is left on the machined surface, chips, or machine tools; hence, no contamination can occur and the cost of disposing of the coolants is eliminated [121]. There are different techniques for performing cryogenic machining [63]. The first method is a cryogenic bath, where the workpiece is fully submerged into cryogenic fluid before or during the machining process. The second method, which is the most commonly used method, is applying emulsified liquid coolants to the cutting zone. The third method is by injecting pressurised cryogenic fluid into the application nozzle, which abruptly expands at the exit of the nozzle and results in a combination of gas and solid particles.

Currently, CM is considered the best solution for the heat-induced problems associated with high temperatures generated when machining titanium alloys [115]. It is an alternative technique of cooling to eliminate the negative effects of petroleum-based cutting fluids and reduce tooling and energy costs [123]. Applications of the refrigerants hardened the tools hence reducing the tool wear, extending the tool life and improving the accuracy of the dimensions and geometry desired. Surface integrity is also improved by CM, which results in a decrease in surface roughness, an increase in surface hardness, and the formation of refined layers with ultra-fine or nanograins. Refined grains improve wear and corrosion resistance while also creating compressive residual stresses on the machined surface, improving product fatigue life [71].

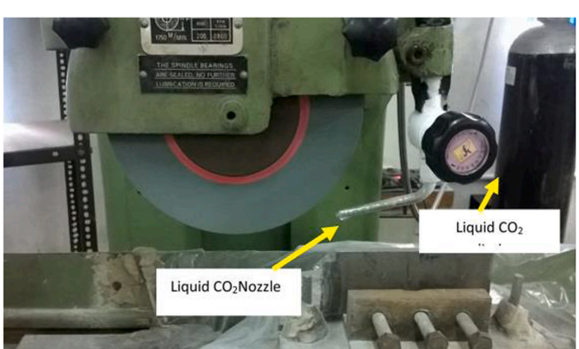

Fig. 4.4. A liquefied carbon dioxide (CO_2) cooling setup [115].

However, the delivery of LN_2 induces the production of ice on the jet flow channel, which is an unwanted side effect of the CM process [11]. The coolant circulatory system becomes clogged because of this event, causing the cryogenic machining process to fail. Furthermore, the low temperature of the coolant enhances the hardness of the workpiece, resulting in a stronger cutting force [115]. This low-temperature coolant also promotes brittleness in the work material, which reduces the work material's machinability [142]. Park et al. [165] found that while machining titanium alloy with only cryogenic cooling, the cutting tool suffered from severe adhesion due to insufficient lubrication. The other disadvantages of CM are the additional costs of equipment and the costly LN_2, which is not reusable.

Difficult-to-machine materials, hard materials, extremely abrasive materials, and superalloys can all benefit from CM [166,167]. It is now used in the aerospace and automobile industries for high-speed machining of hard and superalloys [121]. Studies that have been done on cryogenic machining are reported. Cui et al. [111] observed that cryogenic nano-lubricants exhibit outstanding grinding performance due to their greater viscosity and they significantly increase convective heat transfer capacity. Jawahir et al. [21] explained that cryogenic machining of titanium alloys enhanced the hardness and microstructural characteristics of biomedical implant materials. Setti et al. [116] examined the grinding of Ti–6Al–4 V alloy to evaluate if cryogenic cooling outperforms dry and wet cooling methods in machining. In comparison to alternative cooling methods, they found that CM with LN_2 was successful in minimizing grinding force and surface damage, as well as obtaining a better surface finish. According to An et al. [168], grinding titanium alloys with high-pressure cryogenic air (−20 °C) produces lower grinding forces and surface roughness than grinding with conventional cooling conditions.

Elanchezhian and Kumar [169] investigated the effects of cryogenic carbon dioxide (CO_2) and traditional coolant conditions on Ti–6Al–4 V alloy grinding with a CBN wheel. They found that using CM with CO_2 reduced grinding temperature and surface roughness by 48 and 333 per cent, respectively, when compared to conventional grinding. The experimental results also indicated that CM with cryogenic CO_2 reduces the tangential force and normal force by 3 to 21% and 2 to 99%, respectively. Further, Elanchezhian et al. [170] reported that applying LN_2 during the grinding of Ti-6Al-4 V reduces tangential forces by up to 27% and surface roughness by up to 38%.

Mahata et al. [109] used MQL-aided liquid CO_2 and liquid CO_2 techniques to investigate the grindability of grade 1 and grade 5 titanium alloys. They found significant grinding improvements, including reduced grinding force and surface roughness, as well as improved surface shape and grinding ratio. The medical-grade Ti–6Al–7Nb alloy was investigated by Sun et al. [71] and they found that in comparison to dry and flood cooling, CM with LN_2 greatly improved the surface integrity characteristics. In a study on the machining of Ti-6Al-4 V alloy by Rottela et al. [171], they discovered that cryogenically machining enhanced the hardness of the surface layer compared to dry and MQL machining. They also discovered that CM generates the lowest surface roughness value. When Kaynak et al. [172] compared cryogenic machining of austenitic NiTi to dry and MQL machining, they discovered that CM enhanced productivity and lowered production costs. In addition, the roughness of the surface has been substantially improved. Suhaimi et al. [115] examined the use of LN_2 coolant in the CM of difficult-to-cut materials. They discovered that combining indirect cryogenic cooling with a nano-MQL lubricant approach can increase

Ti-6Al-4 V machinability. The cutting force was lowered by 54% while tool wear was improved by 90%. The studies reported above show that the cryogenic grinding of titanium alloys improves machining performance. It is important to note that most of the studies that have been done on the application of cryogenic coolants are about turning and milling operations. There are a few studies that have been done on the cryogenic grinding of titanium alloys.

High-pressure jet-assisted machining (HPJAM)

HPJAM is one of the methods for maintaining or even improving machining performance by replacing traditional machining processes [121]. HPJAM lubricates and cools both the tool-chip interface and the cutting zone with a high-pressure jet of water or emulsion. Cutting fluid is fed to the cutting tool tip through small diameter nozzles (0.20 mm) at a speed of 97–140 m/s [135] under extremely high pressures in the range of 80–360 MPa. Because of the high pressure, the cutting fluid jet may perfectly enter and cool the cutting zone [135].

This machining improves machining performance by providing cooling effects on the machined surface and allowing management of the friction condition in the cutting zone [173]. The primary goal of HPJAM was to increase the rate of material removal and the efficiency of the manufacturing process of sophisticated materials, including Ti-based alloys and CrCo alloys, which are used in prostheses [174]. Thanks to HPJAM [173], surface roughness is reduced significantly, and tool life is extended while cutting difficult-to-cut materials. In comparison to conventional machining, the HPJAM is more sustainable because of lower cutting fluid flow rates, lower cutting forces, and longer tool life due to less flank wear [63]. It also promotes process productivity by extending the operational ranges of the machining parameters, which improves chip breakability. The disadvantages of the HPJAM include higher initial capital investment for equipment [121]. Mineral oil-based cutting fluids used in the HPJAM are associated with health and environmental effects. The basic principle of the HPJAM set-up is schematically shown in Fig. 4.5.

Studies that have been done on high-pressure cooling in grinding titanium alloys are presented in this section. Shi et al. [175] investigated the feasibility of grinding titanium alloys with electroplated CBN wheels while using a high-pressure coolant. The output responses measured were surface roughness, grinding power, and grinding forces. The study revealed that it was feasible to achieve high material removal rates by grinding the titanium alloys under high-pressure coolants. In another similar study by Shi et al. [176], they found that wheel wear rates were reduced. Nadolny et al. [177] investigated whether a high-pressure hydro-jet might be used to regenerate the active and clean surface of a grinding wheel during the grinding of grade 2 titanium alloy. The findings revealed that using a high-pressure hydro-jet with a pressure of 25 MPa allows for successful grinding wheel surface cleaning. The studies above show that there are improvements in the grinding performance of titanium alloys when using HPJAM. It can also be noted that several studies concerning HPJAM are in the turning of titanium alloys; studies on grinding operations are scarce.

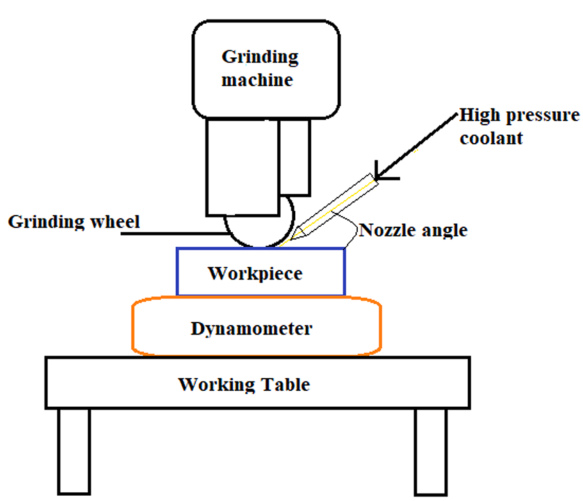

Fig. 4.5. HPJAM schematic diagram.

Conclusion

Several attempts were made in this review to provide a complete overview of the types, grades, and primary requirements of titanium alloys for medical applications. It was established that the biomaterials for manufacturing medical implants ought to be biocompatible so as not to elicit toxic reactions in the human body; corrosion and wear resistance to avoid releases of metallic ions into the human body; good mechanical properties to ensure proper functionality once implanted; and have good osseointegration behaviour to ensures proper interlinking and attachment with natural human parts. Processing methods and thermo-mechanical processes should be tailored to ensure the biomaterial being manufactured has the right chemical composition and microstructures as both have effects on the mechanical and chemical properties of the titanium implant. All biomaterials are chosen and developed with the primary goal of extending the implant's life in the living organism and improving the patient's quality of life.

Titanium biomaterials are known for their great biocompatibility, strong corrosion resistance in body fluids, low Young's modulus, high fatigue strength, and high tensile strength, amongst other qualities. Examples of titanium used in medical applications can be classified as the first and second generations of titanium alloys. The first generations of titanium and its alloys are CP titanium, Ti-6Al-4 V, and Ti-6Al-7Nb, amongst others, and had some problems such as long-term health problems. Because of the issues with the first generation of titanium alloys, new second-generation titanium alloys have been released to the market. This type of titanium alloy has good workability, and good mechanical qualities, and is alloyed with non-toxic and non-allergic components.

Machining titanium alloys is a difficult process because of their low heat conductivity and their chemical reactivity at high temperatures. Cutting fluids have been utilized in the machining of titanium alloys for cooling, lubricating, and cleaning. The advantages of using a traditional flood cooling system are effective heat removal from the workpiece, low coefficient of friction, chip removal, and tool wear rate. However, they have environmental, health, and disposal difficulties, so their use is discouraged. Hence, the concept of sustainable manufacturing through environmentally-friendly machining is becoming relevant in the 21st century.

Sustainable manufacturing entails manufacturing products using techniques with minimal environmental side effects, that are economically feasible, safe for people, and that preserve energy and natural resources. In the manufacturing of titanium alloys, sustainable manufacturing processes have been used with great success. The use of vegetable oil-based cutting fluids has further enhanced the sustainability of the machining as they have neither environmental effects nor health effects and have superior grinding performance compared to mineral oils. Even though dry machining has some limitations, it is a process that is fully environment-friendly and reduces manufacturing costs as compared to machining involving coolants. Some techniques, like MQL, have been researched greatly and have outstanding benefits. Cutting fluid volumes used in machining processes are reduced in MQL, and this makes MQL machining amongst the sustainable methods used in machining titanium alloys. Tool wear in general for MQL machining produced favourable results, extending tool life.

As progress is made in improving the machinability of titanium alloys, nano-particles have been incorporated into the MQL techniques. Applications of different types of nanoparticles that are beneficial to MQL grinding were reviewed. Because of the unique lubricating properties and enhanced heat transfer capability of the nano-fluids, MQL machining has been greatly improved. It was discovered that they improved the cutting tool's efficacy and reliability by lowering the coefficient of friction and wear. In terms of grinding forces, surface roughness, wear rate, and surface finish, NPMQL displayed the best grinding performance. When examining the data published in the literature, it is clear that cryogenic machining is a very promising technique that can increase surface integrity when used to grind titanium alloys in various machining processes. It is a sustainable and clean machining process that has economic, social, and environmental benefits. It is useful in the machining of titanium alloys, where it reduces premature tool wear and gives a good surface finish. Other sustainable techniques like solid lubrication machining and high-pressure machining have been researched and have proved to contribute to the sustainability process in machining titanium alloys. Grinding forces, grinding temperature, and surface roughness have all been reported to be reduced in HPJAM. The technology also enables an effective chip fragmentation process. In summary, the sustainable cooling and/or lubrication strategies employed for grinding titanium biomaterials outlined in the preceding sections perform significantly better than conventional flood cooling.

Future research and future trends

Many research articles on the investigation of titanium alloy machining processes were evaluated in this review. From the review, it was noted that they are research gaps that need further investigation. The usage of cutting tools made of the latest materials in the dry machining of titanium alloys to improve grindability performance could be a future trend. Future research in the grinding of titanium alloys is to incorporate fluid mechanics and machining mechanics in MQL systems. This will aid in the full realization of the advantages of MQL titanium alloy machining, which will have a substantial impact on the worldwide titanium processing industry. Future research can be undertaken to investigate the best sustainable machining parameters involved in the MQL machining of titanium alloys for the best performance. Sustainable air-cooling machining should be examined for its practicality because it provides the required cold with minimal axillary equipment and no unnecessary safety requirements.

According to the literature research, there is a research gap in the study of nanofluids technology effects on titanium alloy grinding. The lubricating aspect of nanofluids as well as the chip generation mechanism in a nanofluid MQL grinding environment for titanium alloys must be investigated. The nanofluids' homogeneity and distribution, as well as their penetration into the machining zone, must be examined further. More research could be done in the future to determine the appropriate nanoparticle concentration for the greatest frictional reduction.

More research into the use of more cost-effective, biodegradable, health-friendly, and ecologically friendly cutting fluids as a future trend in titanium alloy grinding is needed. There is a need also to develop robust commercial cryogenic machining technology to gain from its benefits as far as grinding of titanium alloys is concerned. According to a review of the literature on high-pressure coolants used in titanium alloy grinding, there are few publications. As a result, further research is needed in that field, particularly on elements such as power consumption, so that grinding of titanium alloys might benefit from the technology.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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