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Corrosion, fatigue and wear of additively manufactured Ti alloys for orthopaedic implants

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ABSTRACT
Additive manufacturing (AM) allows for the fabrication of custom orthopaedic implant devices which have complex geometries and similar mechanical properties to bone. This paper reviews the corrosion, fatigue and wear properties of AM Ti alloys to confirm their safety for use in orthopaedic implants. Specifically, AM Ti lattice geometries are highlighted due to their improved osseointegration and better modulus matching with that of bone, making them an attractive option for more durable implant devices. Finally, the properties of current implants made via AM are compared with that made via conventional manufacturing methods to confirm their overall safety.

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Introduction

Titanium alloys are commonly used in the biomedical industry due to their compatibility with body tissue [1]. Specifically, orthopaedic implant devices require good biocompatibility with the surrounding tissue, good corrosion resistance and matching mechanical properties with that of bone, all of which are fulfilled with the use of Ti alloys [2]. Since the first joint arthroplasty involving Ti alloys in the 1940s [3], Ti orthopaedic implants have become widespread, mostly as standardised, commercially available devices [4] that are mass-produced to address a specific clinical need. In the United States alone, it is projected that 572,000 hip and 3.48 million knee arthroplasties will be performed in 2030 [5], demonstrating the large market for orthopaedic implants. Furthermore, as life expectancy and obesity rates increase worldwide [6], so does the incidence of arthritis requiring joint replacements [7].

In recent years, additive manufacturing (AM) of orthopaedic implants is becoming increasingly popular due to its ability to produce complex geometries, therefore overcoming the limitations of conventional manufacturing techniques [8], such as shown in Figure 1. It is predicted that the market for additively manufactured devices will reach USD 26 billion by 2022 [9]. Furthermore, AM allows for the production of custom-shaped implants, as well as the production of hollow structures, or lattices, to better mimic the structure of porous bone. In cage-like lattice implants, osseointegration and bone ingrowth determine their long-term success and durability, ensuring bioadhesion with surrounding bone which reduces the probability of implant loosening and the need for revision surgeries [10–12].

Porous orthopaedic Ti-based implants have been extensively studied in the literature for a wide variety of applications. Bittredge et al. [11] have fabricated and optimised Ti–6Al–4V lattices for total shoulder implants, where 21% of orthopaedic revision surgeries is caused by the loosening of shoulder implants from the humerus bone [14]. Ramhamadany et al. [15] developed a lattice cage structure for talus replacements, which allows 75–90% of its entire volume to be filled with bone graft, enhancing osteogenic potential [16]. This is especially important to treat avascular necrosis of the talus which has very little blood supply [17]. Kuslich et al. [18] have successfully trialled an open-cell lumbar cage in 196 patients presenting with degenerative invertebral disc disease, with a 95.1% bone fusion rate in four years. Popov et al. [19] have successfully implanted a lattice structure to replace bone affected by osteosarcoma resection in canines, thereby providing a viable alternative to amputation. These examples show the breadth of applications of AM Ti implants, which allow for a patient-specific design in a cost-time competitive manner [20–22].

Moreover, current orthopaedic implants are required to be approved by relevant regulatory bodies, such as the Food and Drug Administration in the United States [23] and the CE (Conformite Europeenne) mark in the European Union [24]. Devices such as newly developed implants need to pass pre-clinical testing through clinical studies to assess the efficacy, safety and durability of the devices by means of

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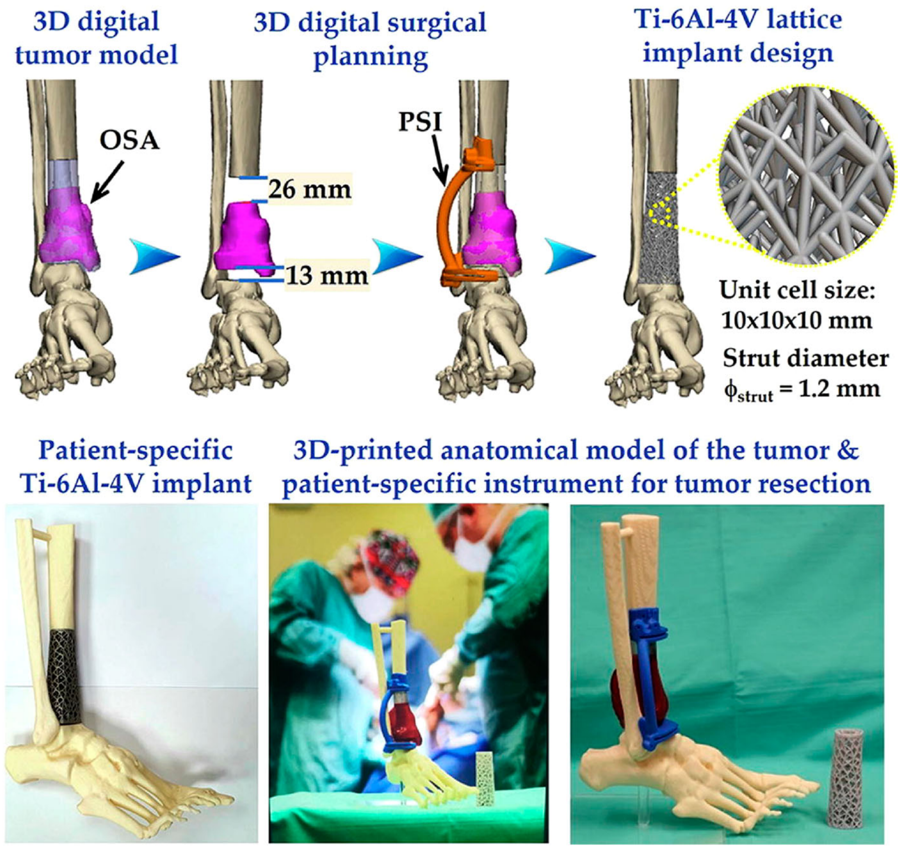

Figure 1. Patient-specific AM Ti–6Al–4V implant. (Reproduced from [13].)

mechanical and biocompatibility testing. Furthermore, post-market follow-up is also required, where patient monitoring post-operation aims to minimise the need for revision surgeries [25]. In contrast, there is a lack of regulatory requirements for custom-made orthopaedic implants using AM as they are not mass produced [26], lowering the barriers to entry for AM companies in producing orthopaedic implants. It remains the responsibility of surgeons who implant these devices to ensure that the quality and safety of the devices are adequate [27].

Despite being corrosion-resistant by means of a protective surface oxide layer, Ti alloys have been found to corrode *in vivo* especially in the presence of fatigue loading [28], shown in Figure 2. This is a non-trivial issue as it often causes implant failure [29–31] as well as the release of metallic ions and debris, which can lead to tissue inflammation and infection [32]. Corrosion in head-neck taper connections in femoral hip prostheses made from Ti alloys occurs in 16–35% of cases according to Gilbert et al. [33]. Furthermore, there are significant challenges in AM implants, such as the critical angle of overhanging structures [34] in addition to inherent defects such as porosity and surface roughness from the AM process [35].

This paper aims to review the corrosion, fatigue and wear behaviour of AM Ti alloys for orthopaedic implants by systematically looking at:

- Manufacturing methods of orthopaedic implants using conventional methods as well as additive manufacturing
- Corrosion behaviour of conventionally manufactured and AM Ti alloys for orthopaedic implant applications
- Fatigue and corrosion fatigue behaviour of conventionally manufactured and AM Ti alloys for orthopaedic implant applications
- Wear and corrosion wear behaviour of conventionally manufactured and AM Ti alloys for orthopaedic implant applications
- Comparing AM and conventionally manufactured orthopaedic implants
- Future prospects of AM orthopaedic implants.

Manufacturing methods

Conventionally manufactured orthopaedic implants are typically made via forging, where metal is shaped using compressive forces through dies, presses and/or hammers [37]. Most orthopaedic implants are manufactured using closed-die forging to its final shape [38]. Implant fixation is often necessary to provide stability to the surrounding bone, such as using acrylic cement [39]. Depending on the location and complexity of the implant, it is also possible to have cementless fixation which involves a press-fit between the bone and implant

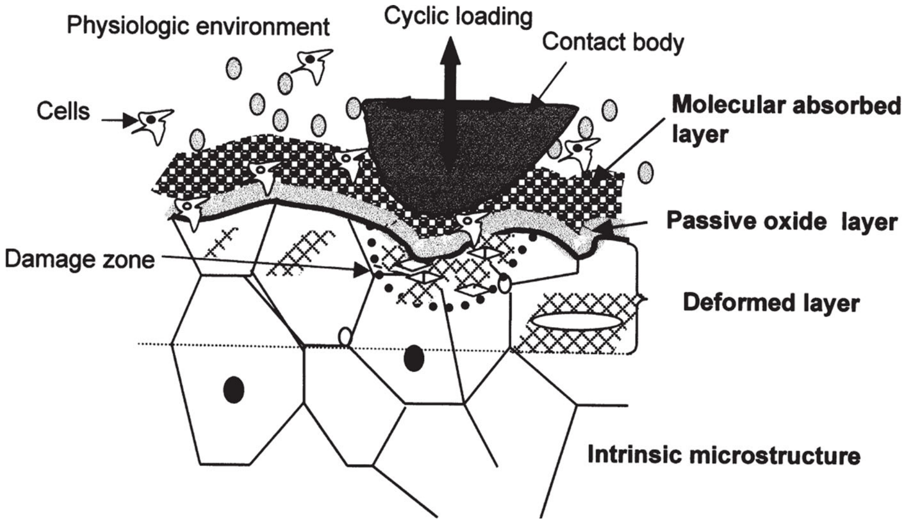

Figure 2. Schematic illustration showing the complex interactions between the material's surface and the physiological environment. (Reproduced from [36].)

[40]. Fixation methods are highly dependent on the implant design, patient's age and surrounding bone quality. However, in the past the use of cement fixation has only proved successful in older patients and less so in younger patients who tend to be more active [41]. Some implants, such as hip tapers, are manufactured from wrought alloys [42, 43].

AM of implants

AM of orthopaedic implants is extremely lucrative as it has the ability to produce components with mechanical properties as similar to native bones [44, 45]. The complex structure of bone can be replicated, such as its anisotropy [46] and specific internal architecture containing macropores [47]. Moreover, the use of computer-aided design (CAD) to manufacture the implants allows for patient-specific components which significantly reduces operation time while also catering for patients with unique implant requirements [27]. For example, AM of implants can often salvage the joint, allowing joint preservation post-operation such as shown in Figure 3.

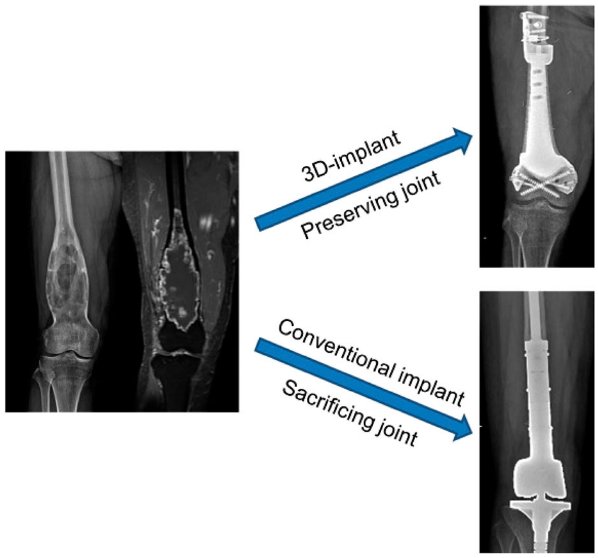

Figure 3. Using additive manufacturing to produce complex implant geometries allowing for joint preservation. (Reproduced from [48].)

Various types of experiments involving additively manufactured Ti–6Al–4V implants have been performed in the literature. These involve different AM techniques, which include selective laser melting (SLM), electron beam melting (EBM), laser powder bed fusion (L-PBF) and wire arc additive manufacturing (WAAM), among others. A few examples of previous work involving corrosion of AM Ti–6Al–4V for biomedical applications are summarised in Table 1.

Different AM techniques produce different microstructures which therefore influence corrosion resistance. As such, studies involving different AM techniques may not be directly comparable with each other. For example, Bai et al. [49] showed that the corrosion resistance of electron beam melted Ti–6Al–4V is better than that of wrought Ti–6Al–4V due to a more refined microstructure and a more compact protective oxide layer. Meanwhile, it has been found that Ti–6Al–4V manufactured by SLM typically has a higher corrosion rate in 3.5% NaCl compared to one conventionally manufactured due to a highly strained martensitic structure and therefore a more porous oxide layer [50–52]. In contrast, the oxide film of SLM Ti–6Al–4V has been found to be more protective in PBS and SBF compared to EBM Ti–6Al–4V [53–55]. Metalnikov et al. [56] also corroborated this finding by comparing EBM and SLM Ti–6Al–4V, with the latter having a slightly better corrosion resistance than the former.

Table 1. Previous studies involving corrosion of additively manufactured Ti–6Al–4V in the literature.

Source	AM method	Loading conditions	Corrosion medium
Leon et al. [51]	SLM	N/A	3.5% NaCl
Ettefagh et al. [57]	L-PBF	N/A	3.5% NaCl
Yang et al. [58]	SLM and WAAM	N/A	3.5% NaCl
Jesus et al. [59]	SLM	Compression fatigue $R = 0.05$, $f = 1$ and 10 Hz with varying stresses up to 10 kN	3.5% NaCl and Ringer's
Chen et al. [60]	L-PBF	N/A	Hank's
Sharma et al. [52]	SLM	N/A	NaOH, NaCl, H_2SO_4, SBF
Gayathri et al. [61]	SLM	N/A	PBS
Bai et al. [62]	EBM	N/A	PBS
Lu et al. [63]	SLM	Wear test with ball-on-disc configuration	SBF
Wu et al. [64]	EBM	N/A	SBF
Wegner et al. [65]	L-PBF	Compression fatigue $R = 10$, $f = 5$ Hz with varying stresses	SBF

AM lattice geometries

One of the main advantages of AM is the fabrication of lattice structures through CAD, allowing the production of complex shapes requiring minimal post-processing [66, 67]. A significant proportion of studies [68–71] involving lattices utilised repeated unit cells throughout the geometry. These geometries comprise beams with a certain thickness, called struts, joined at nodes throughout the lattice. An advantage of strut-based lattices is the ease of design and manufacture, as well as the simplicity of design which allows for reliable computational modelling [70]. However, strut-based structures have sharp curvatures [8], with nodes acting stress concentration sites susceptible to fatigue failure [71]. Additionally, thin struts (1 mm or thinner) are more prone to edge effects [69] further reducing fatigue life.

A better option than struts are surface-based lattices, which have smooth curvatures [72, 73], therefore promotes improved cell attachment and better osseointegration for orthopaedic implants. Among surface-based lattices, triple-periodical minimum surfaces (TPMS), shown in Figure 4, have been found to have zero mean curvature with a single interconnected domain [8]. The lack of stress-concentration sites also reduce fatigue initiation sites, prolonging implant life. However, surface-based lattices are difficult to manufacture. Several studies have investigated different TPMS geometries [74–76]. Kapfer et al. [77] used finite element analysis to determine the mechanical properties of different TPMS geometries, however, it did not take into account surface roughness from the partial sintering of unmelted powders typical of AM.

Lattice geometry is vital for bone regrowth. Zadpoor [73] has studied this extensively, noting that the pores within the lattice should allow for not only mechanical support but also tissue invasion, cell nutrition and oxygenation, and that a minimum pore size of around 300 μm is essential for bone ingrowth and formation of capillaries [79]. Similarly, Gotz et al. [80] noted a minimum of 200 μm pore size is required for osseointegration. Moreover, the curvature of pores also strongly influences bone growth rates [81, 82]. In particular, the rate of bone regeneration process has been found to increase with curvature [83, 84]. Most importantly, a lattice geometry reduces the overall stiffness of the implant, therefore reducing the extent of stress-shielding [85].

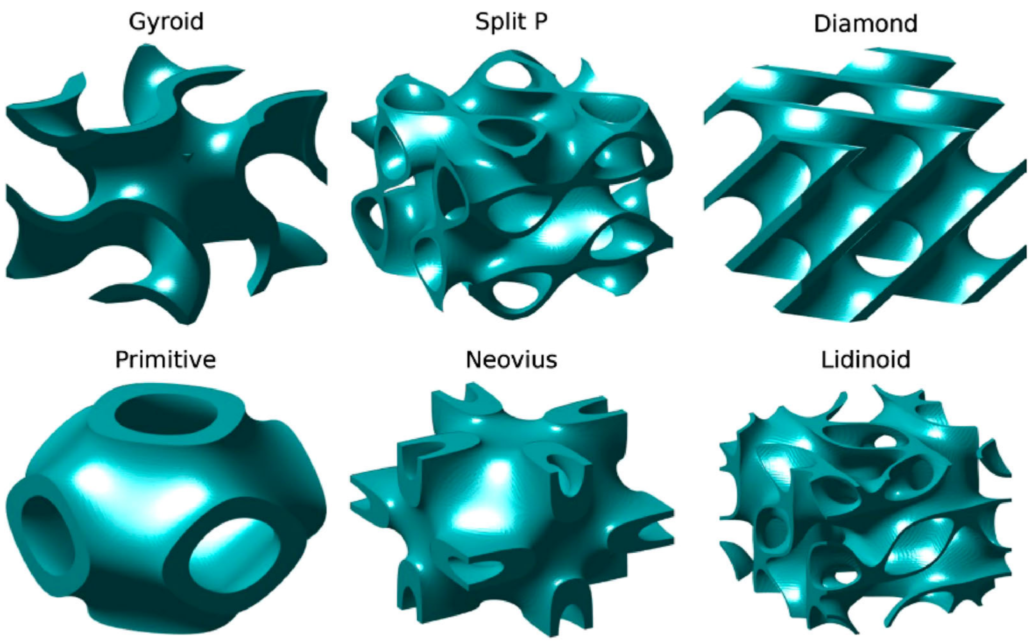

Figure 4. Various TMPS geometries. (Reproduced from [78].)

Besides lattices with repeating unit cells, some studies have looked at stochastic lattices, which are anisotropic and therefore more similar to bone. These types of lattices are much less studied than periodic ones. One type of stochastic lattices is Voronoi, based on a mathematical model initially described by Dirichlet and Voronoi in the early twentieth century. This model started as 'seeds' in a three-dimensional shape and a circle is drawn around each seed, expanding in radius at the same rate until two circles touch. This results in a three-dimensional diagram of circle boundaries which correspond to lattice scaffolds. Voronoi structures therefore are random lattices, dependent on the number of seeds initially used, which in turn is related to a target pore size [86]. Another type of stochastic lattices can be generated using a Poisson disk algorithm which populates a three-dimensional volume with random points, studied by Ghouse et al. [87]. These points are later joined with lines to form struts. This structure is therefore governed by the minimum proximity between points which determines the strut lengths [88]. For stochastic lattices, it is even more difficult to predict the mechanical properties of these structures, however some models have described a relationship between density and mechanical properties of porous structures. One such model was developed by Gibson and Ashby [89], which show that the strength and modulus of a porous material increase with density by a power law. This has been found to apply to additively manufactured lattices by Yan et al. [90].

Subsequently, orthopaedic implant applications comprising different AM lattice geometries have varying corrosion properties. The following section aims to investigate corrosion mechanisms and properties of AM Ti alloys for orthopaedic implants.

Corrosion

Ti alloys are generally considered to be corrosion resistant owing to the spontaneous formation of a protective oxide layer on the metal surface, acting as a physical barrier between corrosive species and titanium metal [91]. However, Ti alloys are not immune to corrosion once the passive layer is compromised [92], and many studies have investigated the mechanism and extent of corrosion of Ti alloys under different environmental conditions, such as in Ringer's solution [93], in NaCl solution [94] and in HCl solution [95].

The most direct way to measure the extent of corrosion is via electrochemical experiments such as potentiodynamic polarisation curves. Cyclic potentiodynamic polarisation (CPP) curves directly measure the corrosion potential, E_{corr}, the corrosion current density, i_{corr} and the corrosion rate. In Ti–6Al–4V, this measurement involves a three-electrode polarisation cell with Ti–6Al–4V as the working electrode, Ag/AgCl as the reference electrode, a Pt mesh counter electrode and the corrosion medium as the electrolyte. A more negative E_{corr} value, a larger i_{corr} value and a higher corrosion rate suggest a more corrosive environment. ASTM F2129-19a states that breakdown potentials of implant devices should be at least 800 mV within a physiological environment [96]. Cyclic potentiodynamic polarisation measurements can accurately assess the extent of corrosion, however, this is limited to a stationary sample and not for corrosion wear or fretting corrosion [97–99].

The protective TiO_2 layer on the surface of Ti alloys can undergo dissolution, therefore resulting in localised corrosion (i.e. pitting corrosion) [58]. This dissolution of TiO_2 layer occurs at high enough potential, termed critical pitting potential, which have been well documented to have a linear function of the logarithm of chloride ion concentration in metals [100]. The mechanism of corrosion therefore occurs by adsorption of chloride ions into the metal substrate and subsequent dissolution reactions occurring at the metal/oxide interface [101, 102]. Soltis [103] has reviewed this mechanism and showed agreement between computational models and experiments in Ti alloys.

Sivakumar et al. [104] found that the corrosion potential is also further influenced by kinetic parameters, and once fretting removes the passive oxide film on the implant surface, there is a potential difference which further increases corrosion rate. This is owing to the fretting contact being the anode and the large area outside the fretting zone being the cathode, the former increasing in area as fretting worsens.

Corrosion medium

There have been many attempts to determine the corrosion mechanism of Ti alloys in the presence of stress and exposure to the body fluid environment, such as by Dimah et al. [105] using phosphate buffered solution (PBS), Yaya et al. [106] using 0.9% NaCl solution, Cvijovic-Alagic et al. [93] with Ringer's solution and Bidhendi and Pouranvari [107] with Hank's solution. Since these solutions contain different concentrations of various chemical species, it is difficult to elucidate the exact mechanism of corrosion by comparing these studies. Table 2 highlights a few studies involving corrosion of AM Ti alloys using a variety of corrosion media. Furthermore, these solutions are not similar to body fluid in composition. For a more systematic study of body fluid corrosion, a solution which has similar ionic concentrations to that of body fluid is crucial. Such solution was developed by Kokubo and Takadama [108] in 2006, which has very similar ionic concentrations to blood plasma. The chemical composition of these solutions as well as human blood plasma is shown in Table 2.

It is important to note that body fluid compositions do not remain constant throughout and also vary from

Table 2. Typical chemical composition of human blood plasma [112] and different physiological solutions [108, 113].

	Ionic concentrations (mM)			
Ion	Blood plasma	Hank's	Ringer's	SBF
Na^+	142.00	142.00	113.60	142.00
K^+	5.00	5.00	1.88	5.00
Mg^{2+}	1.50	1.50	0.00	1.50
Ca^{2+}	2.50	2.50	1.08	2.50
Cl^-	103.00	103.00	115.30	147.80
HCO_3^-	27.00	27.00	2.38	4.20
HPO_4^{2-}	1.00	1.00	0.00	1.00
SO_4^{2-}	0.50	0.50	0.00	0.50
pH	7.20-7.40	6.82	5.92	7.40

person to person [97]. For example, studies involving pH of body fluid before and after a metal implant is placed *in vivo* have found that it can drop from 7.4 to 5.5 due to the disruption of blood supply to the bone [109] while a bacterial infection post-surgery can result in a pH of between 4.0 and 9.0 in the vicinity of implant surface. A localised decrease in pH can result in severe pitting corrosion of the implant. This has also been verified in *in vitro* experiments [110]. Moreover, oxygen content in blood is lower than that of artificial physiological solutions in air due to the presence of haemoglobin in blood, thus repassivation of implant surface, once corroded, is much more difficult in the body [28, 111]. Meanwhile, carbon dioxide in blood will reduce its pH which accelerates corrosion.

Corrosion in AM lattices

There has been very few studies involving corrosion of additively manufactured Ti–6Al–4V lattices in the literature. Sharp et al. [114] studied the effect of porosity on LPBF Ti–6Al–4V Gyroid structures and found out that a lower porosity lattice is more susceptible to pitting corrosion. This suggests a trade-off between lattice optimisation and corrosion behaviour and thus a potential link between void volume, surface area and corrosion. As expected, corrosion appeared to initiate at corners and raised edges, and breakdown potentials for all samples tested were above that found in a physiological environment showing that the implants should not corrode in the body. Losiewicz et al. [115] investigated the severity of corrosion among different TPMS lattices and found that Gyroid lattices are most corrosion resistant. Gabay et al. [116] studied AM Ti–6Al–4V lattice that is infiltrated with a biodegradable Zn–2%Fe alloy and determined adequate corrosion resistance, improved osseointegration and satisfactory bonding between the two materials.

The lifetime of orthopaedic implants containing AM lattices is therefore difficult to predict accurately. In addition to corrosion due to exposure to body fluid environment, the application of mechanical loading will likely reduce their overall lifetime. The next section aims to investigate the effect of fatigue in AM Ti alloys for orthopaedic implants.

Fatigue

Fatigue is well known to be the most common cause of premature orthopaedic implant failure [117], where the application of cyclic stress leads to the initiation and growth of cracks. Upon reaching a critical crack length, catastrophic failure occurs [118]. Fatigue properties of Ti alloys are well studied and are greatly influenced by microstructure types and grain sizes, with bimodal microstructures possessing the highest fatigue strength, followed by lamellar and equiaxed [119]. In many materials including titanium alloys, fatigue cracks often initiate at stress concentration sites such as inclusions, pores, residual surface stresses and grain boundaries. In dual-phase alloys such as Ti–6Al–4V, the softer phase is more prone to fatigue crack nucleation compared to the harder phase [120].

Corrosion fatigue

Corrosion fatigue is a well-known behaviour in alloys where failure occurs under cyclic loading when exposed to corrosive species [121]. Typical fatigue crack initiation in Ti alloys occurs as a result of plastic deformation during cyclic loading, resulting in regions termed persistent slip bands [122].

Ti alloys are generally considered to be corrosion resistant owing to its protective TiO_2 film that spontaneously forms in air on its surface [123]. However, while this oxide film is resistant to chemical attack and corrosive environments under static conditions, it is not sufficiently stable under loading conditions [124] when exposed to body fluids [125]. Corrosion fatigue behaviour of conventionally manufactured Ti alloys is relatively well studied, especially owing to its widespread use in aero-engine and other industrial applications [126]. Dawson and Pelloux [127] studied the different crack propagation behaviours in Ti–6Al–4V when immersed to different environments. Similarly, Baragetti and Arcieri [128] found that Ti–6Al–4V exposed to 3.5% NaCl solution has a 20% reduction in stress concentration factors in fatigue tests compared to that exposed to laboratory air.

Vallittu and Kononen [129] also found that crack propagation in Ti alloys under cyclic loading is relatively rapid compared to cobalt-chromium and gold alloys. In the field of metallic implants, pitting corrosion is common owing to the dissolution of protective passive film due to contact with aggressive species such as chloride ions [117]. Fatigue cracks have been found to nucleate near these pits, which continue to grow until fracture occurs. Studies by Azevedo [130] and Magnissalis et al. [131] have shown that metallic implants have reduced fatigue lives and faster fatigue crack propagation due to corrosion.

Corrosion fatigue behaviour in Ti alloys is strongly influenced by microstructure. This has been studied by Bache and Evans [132], who found that the lamellar microstructure is the most sensitive to environmental conditions, presenting with a significant increase in crack growth rates, as compared to bimodal and mill annealed microstructures. Similarly, Gregory [133] found that corrosion fatigue behaviour is influenced by crystal orientation (texture), showing only significant reduction in fatigue life when the basal planes of the hexagonal α phase are perpendicular to tensile stress. In contrast, Roach et al. [134] found that the presence of a notch, rather than environmental conditions, contributes more to a reduced fatigue lifetime of Ti alloys, due to higher tri-axial stress state at the notch.

It is important to note that to date, there have been no studies to determine the relationship between microstructure, surface finish and fatigue crack growth in Ti alloy implants under body fluid environments [117], owing to the difficulty in replicating these conditions in laboratory conditions, i.e. physiological medium containing proteins, enzyme and ions at 37° C with varying wear and cyclic loading conditions.

Fatigue in AM lattices

Fatigue behaviour in AM components is notoriously difficult to predict and is highly dependent on the sample geometry, surface finish and defect concentration. Typical AM components contain porosity which generally reduce their overall fatigue strength compared to conventionally manufactured components of the same geometry [135]. The fatigue resistance of Gyroid lattices has been extensively studied by Yang et al. [136], who found that the internal strut topology of lattices play a crucial role in its fatigue resistance. Specifically, lattices such as Gyroid with helical surfaces and round pores are effective in reducing tensile stress around nodes driving crack initiation and propagation. Other studies have presented that fatigue resistance improves with lower internal defect percentage. Specifically, Kelly et al. [137] found that void defects within struts are typical sites for crack initiation, whereas Mahmoud et al. [138] found that thicker struts require more laser exposure, leading to worse heat accumulation and unstable melt pools. These yield internal defects thereby decreasing fatigue resistance. Similarly, Li et al. [139] investigated the compression fatigue behaviour of lattices with high porosities (60–85%), showing that fatigue failure in highly porous lattices is mainly dominated by cyclic ratcheting leading to fatigue crack initiation and propagation. An improved fatigue strength is observed in lattices with higher relative density.

Burton et al. [140] performed finite element analysis on different lattice geometries and found that the Schwartz primitive (pinched) lattice has the strongest unit cell, with a 10% volume fraction sample displaying good fatigue properties. Moreover, controlling the volume fraction the overall stiffness of the implant which can be tuned to match that of the surrounding bone. In fact, it has been widely hypothesised that having a slightly lower Young's modulus than bone is beneficial for bone ingrowth and having some implant deformation promotes better bone formation [141, 142]. In the study of stochastic lattices, Mhurchadha et al. [143] have found that strut thickness in Voronoi structures has a significant influence on fatigue lifetime.

Few studies have studied the combined effect of corrosion and fatigue in AM Ti–6Al–4V. Jesus et al. [59] found that fatigue crack propagation rate increased by 290% in Ringer's solution and more than 330% in 3.5% NaCl solution in comparison to an air environment.

Premature failure from corrosion and fatigue is common in orthopaedic implants. Another common cause for early failure of implants is due to fretting/wear leading to the release of metal debris harmful to tissue health. The following section aims to investigate the effect of wear on AM Ti alloys for orthopaedic implants.

Wear

Ti alloys are known to have relatively poor wear resistance [144]. This is largely due to the low protection of the surface oxide layers against mechanical wear; oxides of Ti easily spall and fragment [145]. This is especially the case for Ti on Ti contact.

Orthopaedic implants, especially joint replacements, may contain bearing surfaces which may wear during its lifetime. Metal on metal bearing surfaces typically cause less wear over time compared to metal on polymer and ceramic on polymer surfaces [146], hence are preferable for younger patients [147]. However, metal wear debris and metal ion release are detrimental to human health [148]. Furthermore, the release of metal ions due to tribocorrosion can also negatively affect the mechanical stability of the implants, causing implant loosening [105]. An example of bone loss due to wear debris causing implant loosening is shown in Figure 5.

Corrosion wear

Tribocorrosion is the combined effect of interactions between corrosion and mechanical wear [150]. Tribocorrosion studies usually perform wear tests with and without body fluid solutions, such as that following ASTM G77-83 using block-on-disc configuration [93], using pin-on-plate tribometer [151] or using ball-on-disc tribometer [105]. These methods produce different wear tracks and are not directly comparable with each other.

In the case of Ti–6Al–4V, Dimah et al. [105] have found that the protective oxide layer formed when

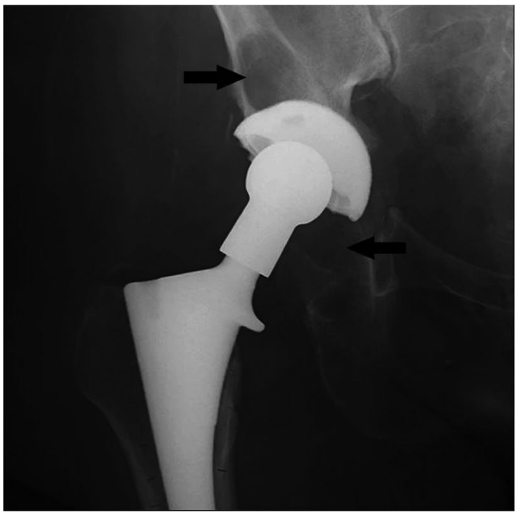

Figure 5. Bone loss surrounding a hip implant. (Reproduced from [149].)

exposed to body fluid solution consists of mainly TiO_2, with TiO and Ti_2O_3 in the metal/TiO_2 interface and Al_2O_3 in the TiO_2/solution interface. Milosev et al. [152] found that the wear resistance of the alloy therefore depends on the mechanical properties of the oxides on the surface layer. This is due to the slow regeneration of the passive film once damaged by wear [93]. Wear of implant devices can lead to implant loosening as well as surrounding tissue inflammation due to wear debris, which is highly undesirable [40].

Wear in AM lattices

Tribocorrosion studies in AM Ti–6Al–4V lattices are limited and have generally shown worse wear properties compared to conventionally manufactured Ti–6Al–4V. This is mainly owing to the increased surface roughness in AM Ti–6Al–4V components. The inherent irregularities on the surface of AM components are due to the solidification of the melt pool, leading to partially melted powder particles [153] on the component surface, which are in the range of 10–60 μm in diameter for SLM powders and 50–150 μm for EBM powders [154]. Furthermore, the layer-by-layer deposition in AM causes surface defects such as the stair-step effect [155]. Both these factors contribute to poorer tribocorrosion properties as surface irregularities can act as initiation sites for pitting corrosion [156].

Interestingly, Shahsavari et al. [154] found that corrosion resistance in AM Ti–6Al–4V is improved with decreasing surface roughness, which can be achieved using techniques such as electron beam surface remelting. However, this may be undesirable as a decreased surface roughness will be detrimental in aiding bone ingrowth and decrease bone adhesion.

Furthermore, tribocorrosion studies are only applicable in cases where the Ti implant has an articulating surface such as to replace ball-and-socket joints. However, where articulating surfaces are required, orthopaedic implants are often highly polished to remove surface roughness and coated with a ceramic layer such as titanium nitride (TiN) which has very high wear resistance, hardness and smoothness [32]. Where articulating surfaces are not required, corrosion wear is unlikely to occur as there is no friction experienced by the implant. In these cases, bone fusion is a more significant issue to prevent implant loosening [157], hence a porous implant such as open-cell lattices is beneficial. The pore size and shape can be optimised by varying the geometry of the lattice to ensure maximised bone ingrowth and adhesion to the implant [72].

To improve tribocorrosion properties in AM Ti alloys, many studies have investigated various surface finishing methods, such as sandblasting [158], chemical etching [159], passivation and electropolishing [64]. For example, adding diamond like coating (DLC) on the surface of implants has been found to improve the overall wear resistance, as well as encouraging cell growth [146]. However, it is important to note that for orthopaedic implant applications, increased surface roughness is favourable for bone ingrowth to prevent loosening.

Discussion

Additive manufacturing inherently produces components with more defects than conventional manufacturing owing to its high cooling rate [160], high residual stresses [161], porosity [162] and surface roughness from partially melted metal powders [155]. As such, AM implants are more susceptible to localised corrosion as well as tribocorrosion [53]. Subsequently, metal ion release to body fluid is a safety concern as these can cause toxic and allergic symptoms [163]. In Ti–6Al–4V implants, there is a preferential release of Ti ions but also significant amounts of Al and a small amount of V ions, the latter being a concern as it is toxic to humans [164].

The fatigue properties of AM lattices are challenging to predict due to its complicated geometries. This is in comparison with conventionally manufactured Ti implants for which corrosion fatigue behaviours have been extensively studied. In AM lattices, modelling fatigue crack growth is tricky as there are significant variations in strut surfaces between manufactured components and CAD data [87]. Consequently, predicting the fatigue behaviour of stochastic AM lattices is even more difficult, however, some studies have attempted this and compared their results with experimental data [88] with good consistency. Even so, AM artefacts such as residual stress and surface roughness are known to reduce the fatigue life of AM components. Under corrosive environments such as body fluid, fatigue life of these lattices is expected to be further reduced [133], although the extent of this remains a gap in literature that needs to further studied.

In terms of wear, Chiu et al. [165] have investigated and compared the performance of both conventionally manufactured and AM Ti–6Al–4V and showed worse corrosion wear behaviour in the AM specimen. This has been attributed to microstructure defects and porosity on the AM specimen causing an enrichment in oxygen vacancies on the surface layer when exposed to body fluid solution. Interestingly, the formation of passivating TiO_2 layer on the surface of AM Ti–6Al–4V has been found to be faster in specimens with a finer grain structure. This suggests that the corrosion wear behaviour in AM Ti alloys can be improved through specific heat treatment cycles aimed at grain refinement.

Future prospects

The demand for orthopaedic implants is ever-increasing for a variety of medical needs. While the corrosion, fatigue and wear behaviour of AM implants are challenging to predict, this paper has shown the breadth of experiments conducted in the literature to determine these properties in AM implants, ensuring they are safe for clinical use. With its inherent defects and complicated geometries, the lifetime of AM implant devices might be reduced. However, with its lattice geometries, the devices allow for faster bone ingrowth, ensuring that the surrounding bone can quickly bear load thus reducing the reliance on implant devices themselves.

AM of orthopaedic implants will also continue to be the lucrative option especially in specialised cases such as oncology and paediatric cases, where commercially available modular implants are not appropriate and limb salvage is a priority [48]. Furthermore, revision surgeries for patients with bone loss surrounding existing implant devices require custom-shaped implants which is only possible with AM [166].

Conclusion

The development of AM technology allows for the fabrication of complex implant geometries, opening new opportunities in the field of custom-made implants. This paper reviews the corrosion, fatigue and wear behaviour of Ti alloys for orthopaedic implants fabricated by additive manufacturing.

- AM lattice geometries enable implants to have mechanical properties that are similar to surrounding bone while allowing bone ingrowth, both of which are desirable for implant durability.
- There are limited studies on the behaviour of these lattices under both mechanical loading and body fluid environment. However, due to fast cooling rates and poor surface finish in AM components, it is expected that both corrosion fatigue and wear properties of AM Ti–6Al–4V orthopaedic implants to be worse.
- There is therefore a trade-off between osseointegration of implant and surrounding bone and their long-term mechanical and corrosion properties.

Overall, this opens up possibilities for research in the promising area of AM lattice geometry optimisation for optimum mechanical properties, enhanced osseointegration and corrosion resistance.

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