Mechanical performance of additively manufactured pure silver antibacterial bone scaffolds

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Abstract

Implant infection is a serious complication resulting in pain, mortality, prolonged recovery, and antimicrobial resistance (AMR). Reducing the risk-of-infection associated with tissue implants require imminent attention, where pure silver (Ag) offers enormous potential. However, the printability, mechanical performance nor microbial resistance of additively manufactured (AM) pure Ag is unavailable in literature. This is critical as Ag is thought to play a vital role in the development of AM patient-specific infection resistant implants in the decade to come. The study therefore additively manufactured 99.9% pure-Ag through selective laser melting (SLM) and systematically investigates its mechanical performance. The validated SLM process parameters were then used to conceive two fully porous bone scaffold each at approximately 68 and 90% (*wt.*) porosity. While the study brings to attention the potential defects in SLM pure-Ag through X-ray nanotomography (X-ray nCT), the mechanical properties of porous Ag scaffolds were found to be similar to cancellous bone. The study achieved the highest SLM pure-Ag density of 97% with Young's modulus (E), elastic limit (σ_e), yield strength (σ_y), ultimate strength (σ_{ult}) and ultimate strain (ε_{ult}) in the range of 15.5-17.8 GPa, 50.7-57.7 MPa, 57.6-67.2 MPa, 82.4-95.9 MPa and 0.07-0.10 respectively. The antimicrobial efficacy of printed silver was tested against the common implant infection-causing *Staphylococcus aureus* and led to 90% and 99.9% kill in 4 and 14 hours respectively. The study, therefore, is a first step towards achieving a new generation Ag-based AM infection resistant porous implants.

Keywords: Additive Manufacturing; Selective Laser Melting; Bone scaffold; Pure Silver; Antibacterial; Mechanical Performance

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1. Introduction

Millions of people across the world suffer from inflammatory and degenerative diseases associated with bone and joints [1,2]. According to the review by Ribeiro *et al.* [3], a significant percentage of all chronic diseases in people over the age of 50 in developed countries fall under this category. Navarro *et al.* [4] predict that the number of people affected by bone diseases and requiring invasive surgeries will double in the next decade. Despite the substantial advances in invasive surgery and aseptic techniques, implant-related infection remains an all too common complication [5]. O'Cearbhaill [6] suggested that even systematically introduced antibiotics were largely ineffective for peri-implant infections leading to life-threatening complications. This is primarily due to antibiotic resistance, insufficient drug penetration, and suboptimal bioavailability at the implant site. Once the infection takes hold, the implant needs removal, followed by the resection of the infected tissue. The surrounding tissue must then be treated with antibacterial agents before a surgical revision can be performed. Overall, implant-related infection accounts for extremely high healthcare costs and long recovery time adding to the economic burden estimated by Kurtz *et al.* [7] to be around \$1.62 billion in the US alone by 2020.

As a response from the research community, multifunctional bone scaffolds that offer protection against infections were being intensively investigated [8–13]. The use of additive manufacturing (AM) for the digital fabrication of permeable implants loaded with antibiotics that can be released into the infected sites was one option being considered [14–17]. Studies conducted by Krakhmalev *et al.* [18] and Macpherson *et al.* [19] identifies the possibility to infuse antibacterial metals to reduce infection. Other notable strategies include the development of bespoke surface coatings [20,21] to achieve infection resistance as reviewed by Chouirfa *et al.* [22] and Orapiriyakul *et al.* [23]. Some of these strategies have been applied by Vaithilingam *et al.* [24] and Zhang *et al.* [25] to porous AM titanium (Ti) based implants for antimicrobial incorporation.

A proportion of literature has also been dedicated to implant surface coatings that featuring metallic ions such as silver (Ag) [26–30], zinc (Zn) [31–34], copper (Cu) [19,35–37] and lithium (Li) [38–40] for the reduction of implant infection. The use of metallic ions as infection resistant agents was being increasingly investigated as a result of wide-spectrum antimicrobial properties [41]. Metallic ions were also favoured over targeted antibiotic delivery due to the growing concerns over antibiotic resistance. Among the different metallic ions experimented, Ag offers the highest potential with well documented antimicrobial properties even for superbugs such as methicillin-resistant *Staphylococcus aureus* (MRSA) [42].

Electrodeposition and anodising have also used to introduce Ag ions in implants, which has also shown infection resistance without altering biocompatibility [41]. In particular studies [43] have shown that anodised surfaces with consistent Ag release can even destroy mature bacterial biofilms. Studies [25,44] also show that the antimicrobial performance of silver can be coupled

with antibiotics to further enhance the membrane permeability of Ag ions; an ability that is particularly advantageous in combating antimicrobial-resistant (AMR) bugs. Direct AM of pure Ag was thus worth investigating as it allows for further incorporation of the best strategy to combine different methods to develop infection resistant implants.

Despite the established use of Ag nanoparticles [45–47], ions [48–50], alloys [51–54], and coatings [55–57], no studies have reported AM of pure silver. This means that little is known regarding the mechanical behaviour of pure silver from an additive manufacturing perspective. While AM pure Ag by itself might not offer the required structural strength for replacing cortical bone, it can lend itself to the development of hybrid implants that offer antibacterial efficacy without secondary post-processing. This not only reduces the cost of developing fully porous antibacterial bone scaffolds but also accelerates the bench to bedside development of patient-specific antibacterial scaffolds. Furthermore, AM pure Ag scaffolds can be used as a cancellous bone replacement as they may offer comparable strength [58]. Furthermore, direct AM of pure Ag will serve as the basis for the development of new functional Ag-based biomaterial and alloys suitable for infection resistant total bone replacement.

When it comes to the manufacturing process, research [28,59] suggest that the infection resistance of Ag-coated implants were influenced by the fabrication process due to its effect on the ion release. While soaking AM Ti implants surface with Ag showed antimicrobial properties, chitosan assisted Ag+ showed no efficacy towards *Staphylococcus aureus* (*S. aureus*) infection. Resistance to *S. aureus* is critical as a significant proportion of all scaffold infections are caused by the *S. aureus* species. Accordingly, this study not only pioneers the development of additively manufactured pure silver but also establishes its mechanical property and antibacterial efficacy towards *S. aureus*; the most common implant infection-causing microbe.

Numerous biomaterials such as CoCr and Ti alloys show excellent mechanical properties such as strength, toughness, corrosion, and wear performance [60,61]. However, these materials are not inherently antibacterial and various surface techniques are often sought for inducing antimicrobial properties. In most cases, the antimicrobial agents used are metal particles such as silver (Ag), copper (Cu), gold (Au), and zinc (Zn). Among these, Ag or Ag ions (Ag+) have long been reported to have excellent antibacterial activity [62]. The most recent study carried out by Liang *et al.* [63] suggests that surface modification of CoCr alloy by silver coatings is generally considered the most efficient way to reduce the bacteria and microbial adhesion. The scope of this study is limited to the performance of additively manufactured pure silver, which is structurally weak in comparison to CoCr or Ti, however, the advantage is its antimicrobial property. The intention of developing AM pure Ag is not to be used as a standalone material to compete with existing biomaterials. Now that the parameters and performance of laser melted pure Ag is understood, effective alloys can be developed that delivers a good balance between mechanical and antimicrobial property.

The investigations on the biocompatibility of metallic Ag, their micro-particulates, and implant devices are still in their infancy. Comprehensive studies are required to fully understand the biocompatibility associated with Ag metal particulate exposure in general. *In vitro* studies shows that a time and dose-dependent impairment of cell viability and cell proliferation of bone-forming cells exists after AgNP incubation. Pauksch *et al.* [69] confirm that the cytotoxicity of the AgNP occurred at higher concentrations, which can be reduced by incorporating AgNP into other biomaterials or using micron particulates. This aspect was demonstrated in studies conducted by Kanwal *et al.* [70] where the *in vitro* biocompatibility of the Ag at varying concentration was verified by the MTT assay. The results showed all concentrations to be non-toxic up to 10 $\mu\text{g ml}^{-1}$ with cell viability values of more than 70%. However, at higher concentrations (up to 20 $\mu\text{g ml}^{-1}$), samples became moderately toxic. A similar trend of increasing cytotoxicity against human cell line U937 by increasing particle concentration was reported by Nyga *et al.* [71]. However, it is important to remember that all these tests were carried out *in vitro* using AgNPs.

One of the most extensive studies on bulk Ag coated medical devices were carried out by Jansen *et al.* [72] where animal studies showed no signs of acute toxicity or adverse tissue reactions during an observation period of 3 days. Long term implantation data from rabbit paravertebral muscle for periods of 3, 10, and 30 days also showed no significant differences between Ag coated and uncoated samples [73]. Overall, Ag coated samples did not reveal differences in biological reactions compared to the control in short term toxicity tests as well as in implantation studies. These biocompatibility data were also consistent with studies reported by Spadaro *et al.* [74] on mouse bone marrow cells treated with pure Ag. Other relevant works include Chu *et al.* [75], who investigated silver-treated nylon sutures that revealed excellent biocompatibility. Overall, *in vivo* analysis using bulk silver coating shows that silver is biocompatible at a low dosage with no adverse local tissue responses.

Even though the data on the influence of Ag on bone regeneration is limited, Zhang *et al.* [76] report that Ag particulates have favourable osteogenic properties. Osteogenesis in general is a complex process. In most cases, it follows the formation of cartilage by chondrocytes, which is then removed and replaced by bone by osteoblasts. Osteoblasts develop from mesenchymal stem cells (MSCs) found in the periosteum and the bone marrow. MSCs differentiate through pre-osteoblasts, transited osteoblasts, and finally into mature osteoblasts. Pre-osteoblasts and transited osteoblasts are mitotically active, expressing alkaline phosphatase, but do not secrete bone matrix. Mature osteoblasts are mitotically inactive but are capable of secretion of the bone matrix that contains the relevant proteins and minerals. The bone matrix protein is mainly collagen, which provides tensile strength. The matrix is then mineralised by the deposition of a calcium-phosphate-hydroxide that provides compressive strength. Ag particulates exhibit new and unusual biological properties on cells in that they increase the rate of wound closure by promoting the proliferation and migration of keratinocyte on one hand, while driving the

differentiation of fibroblasts into myofibroblasts, on the other hand, thereby promoting wound contraction [77–79].

As demonstrated in previous studies [64–68] AM as a process is now a proven and efficient method in fabricating custom geometry with targeted properties including patient-specific bone scaffolds. Nevertheless, further improvements in both materials and processes are still necessary before its clinical adoption to achieve both infection resistance and biomechanical performance. Accordingly, this research investigates the development of additively manufactured 99.9% pure silver bone scaffolds for the first time. The study reports the necessary process parameters to print 97% dense pure Ag using selective laser melting (SLM) technique. The mechanical performance of the printed Ag was then systematically studied and its antibacterial efficacy towards *Staphylococcus aureus* (*S. aureus*) evaluated. The established process parameters were then used to conceive two fully porous bone scaffolds at approximately 68% (*wt.*) and 90% porosity, and their performance validated using both analytical and experimental techniques. Emerging visualisation techniques such as X-ray nanotomography (X-ray nCT) was used to analyse the overall quality of pure-Ag scaffolds and to identify potential internal defects. This is the first study on the SLM of pure-Ag antibacterial bone scaffolds and it is anticipated to accelerate the bench to bedside development of infection resistant Ag-based hybrid scaffolds that do not require significant post-processing treatments or coating.

2. Methodology

2.1. Additive manufacturing and materials

All experimental samples were additively manufactured using gas atomised 99.9% pure-Ag with morphology as shown in Fig. 1a. It can be seen that the powdered particles are spherical and have satellite particles attached. The average size (dia.) of the satellite particles was found to be 2 μm . Particle size analysis was carried and identified that the size distribution was between 18.8–38.6 μm . The particle volume distribution (PVD) was relatively evenly spread with only 10% of particles below 18.8 μm that can be primarily distinguished as smaller satellite particles. This meant that the atomised Ag powder used in this study featured a near-normal size distribution, which is favourable for SLM.

The composition of the powdered post-atomising was evaluated using energy dispersive spectroscopy (EDS), the result of which are presented in Fig. 1b. The EDS analysis was critical to analyse the purity of the feedstock as contaminants can cause significant variation in the thermomechanical behaviour during laser melting. The EDS analysis confirms that the powder was 100% pure with no contaminants and oxidation.

A Powder Bed Fusion (PBF) machine featuring a 400 W fibre laser system was used for SLM. A 370 W overhead laser system was precisely modulated based on the process parameters to generate the pure-Ag melt pool. The scan speeds were varied from 400 to 1000 mm/s at 50 mm/s

increments. This range was selected to maximise the energy input while allowing for a realistic fabrication speed. While the lower scan speed (400 mm/s) allows for a high energy density input to produce single tracks; higher scan speeds are desirable as they reduce build time making the process more cost-effective. Standard materials in comparison such as Aluminium (Al) and Titanium (Ti) have scan speeds of 800 and 1250 mm/s respectively.

Fig. 1. SEM and EDS data showing (a) atomised pure-Ag particles used for SLM and (b) corresponding elemental analysis to assess contamination.

Measurements were taken along each molten track produced and the maximum, minimum, average, and range were recorded for the bead width. All tests were conducted in an argon atmosphere with oxygen content in the process chamber below 0.1%. The laser melting process was carried out on substrates that were heated to 35°C. Finally, the process parameter combination that achieved the highest density (97%) was chosen to fabricate the experimental specimens as listed in Table 1. These parameters were selected as they were found to be optimum for the size of the smallest wall thickness (0.3 mm) being manufactured and resulted in a 97% dense part.

Fig. 2. The scaffold designs and the associated unit cells where SC68 and SC90 feature a porosity of 68.46 and 90.69% (*vol.*) respectively. The scaffolds have a height of 10 mm and a diameter of 14.5 mm.

The CAD data for scaffold fabrication was advised by a previous study [64] and featured a low and high porosity design namely SC68 and SC90 as shown in Fig. 2. The two designs were chosen to conceive a relatively simple and complex bone scaffold while meeting all the porosity requirements associated with bone ingrowth. Due to the porous nature of the scaffolds, a diagonal self-supporting build was carried out without the use of internal support structures. Following the SLM process, submerged wire Electro-Discharge Machining (EDM) was employed to remove the as-built samples for further analysis.

2.2. Morphological analysis

2.2.1. X-ray nanotomography

The overall porosity and defect analysis of the SLM samples were carried out using multiscale X-ray nanotomography (X-ray nCT). X-ray nCT is a rapidly emerging non-destructive analysis tool suitable to study the structural integrity and internal details in a wide range of materials [80]. The non-destructive nature of the technique allows the analysis of internal defects and pores of the same components that can later be subjected to conventional experimental testing. The technique is increasingly being used in AM due to the synergy between the complexity of SLM and porosity defects.

The system used in this study was the Skyscan 2211 multiscale X-ray nCT manufactured by Bruker. The SLM samples were mounted on a rotary platform suspended on air bearings and placed between the X-ray source and the receiver. Each sample was then rotated 360° at 0.1° incremental points. For each sample, approximately 11000 images were collected. A Gaussian smoothening at a kernel size range between 0.5 and 0.7 was used for the reconstruction.

A region of interest (ROI) method using CTVox (Bruker) was applied to the reconstructed data and the volumetric porosity was evaluated using the inverse data method. To do this, a watershed segmentation of the raw data was carried out to obtain a binary histogram where different phases were assigned to voxels corresponding to pores and material. The scan data was visualised using a 3D volume rendering and iso-surface module. Although the technique is emerging, the method results in highly accurate porosity evaluation and widely considered best practice [80]. The individual X-ray scans took ~ 4 hrs and each reconstruction ~ 3 hrs using a 40 core xeon workstation expending 128 GB of RAM supported by CUDA enabled GPU acceleration.

2.2.2. Surface characterisation and elemental analysis

The 2D surface data of the SLM samples were evaluated using a Zeiss scanning electron microscopy (SEM). The surface roughness of the implant resulting from DMLS was evaluated using an Olympus LEXT OLS3100 confocal laser scanning microscope. The laser identified the lowest site and referenced this as zero, from which peaks and valleys were referenced to evaluate the mean surface roughness value. Energy dispersive spectroscopy (EDS) was used to carry out the elemental analysis to assess the purity of both the pure Ag feedstock and printed samples. For the pure Ag powder (Fig. 1), the EDS analysis was used to determine the elemental composition of individual points. A total of three points were analysed. To analyse the scaffold sample (Fig. 13) contamination inclusive of the surrounding base metal (Ag), finite area scans were collected for a spectrum of the lateral distribution of elements from the imaged area.

2.3. Mechanical testing

Both tensile and compressive tests to evaluate the material performance of SLM Ag and the performance of porous scaffolds were carried out using a Zwick Roell Z1474 universal materials test rig (Fig. 3) having a maximum load capacity of 100 kN. The tests were carried out to obtain the stress-strain ($\sigma - \varepsilon$) responses for all the specimens. Before commencing the tests the rig was calibrated and verified following BSENISO 7500-1 [81]. While the tensile coupons were pulled to failure the scaffolds were compressed to a maximum deformation of 60% at 0.01 mm/s.

Fig. 3. Experimental test rig used for the mechanical testing of Ag samples.

In all cases, a crosshead-controlled displacement with an 80 kN threshold was used to keep the compression from stopping at the yield limit. Deformation beyond the elastic limit was essential to study the failure modes and the overall behaviour of both the pure-Ag material and scaffolds. Based on the $\sigma - \varepsilon$ curve, the properties of the test samples were characterised for Young's modulus (E), elastic limit (σ_e), yield strength (σ_y), ultimate strength (σ_{ult}) and ultimate strain (ε_{ult}). To ensure data quality, all the mechanical testing was carried out using non-slip platens whose performance was monitored through a Nikon high definition video capturing device. The device was calibrated, and the captured frame-by-frame data of the sample and strain locations were then analysed to make sure the recorded strain is accurate. The relative motion of both the sample and crosshead was also monitored to identify any sample slippage for both the compressive and tensile test cases. Two samples were tested for each category under identical conditions to ensure the consistency of data observed.

2.4. Antimicrobial evaluation

The antimicrobial properties of the SLM pure-Ag were evaluated *in vitro* against *Staphylococcus*

aureus (ATCC 6538) obtained from the American Type Culture Collection (LGC Standards, Teddington, UK). *S. aureus* was chosen for the initial analysis as they cause two-thirds of all orthopaedic implant infections and they are the principal causative agents of infection affecting bone [3]. The organism was cultured for 24hrs on tryptone soy agar (Oxoid, Basingstoke, UK) at 32°C to allow for colony formation. After incubation, overnight suspensions of bacteria were prepared in Dulbecco's Phosphate Buffered Saline (DPBS) solution at a concentration of 1×10^7 to 1×10^8 CFU/ml. SLM pure-Ag disk was disinfected for 20 min in 70% propan-2-ol followed by air drying. Each disk was placed into 12 well plates containing 1 mL DBPS per well one disk was placed per well and inoculated with 0.01 ml of the overnight suspension. The plates were then incubated at 30°C and 70 rpm. At predetermined time intervals samples were taken, serially diluted, and plated on agar plates using WASP Touch Spiral Plater (Don Whitley, Bingley, UK) and incubated at 32°C for 24 h. The resulting colonies were counted using an aCOLyte 3 HD automated colony counter (Synbiosis, Cambridge, UK), and the reduction in viable bacterial cells was plotted as the log reduction in viability for each time point compared to zero-time viability. DPBS solution was used as a negative control. The methodology is widely used to characterise antibacterial performance when it comes to biomaterials [82].

3. Results and discussion

3.1. Additively manufactured pure silver

3.1.1. Mechanical performance and process parameters

Although materials such as aluminium, gold, and copper have been proven to be suitable for the SLM process, little is known regarding the AM of pure silver, let alone their mechanical performance. In this regard, before demonstrating the AM pure-Ag porous bone scaffolds, it was critical to understand the mechanical performance of laser melted pure Ag material. This was critical as the result of the SLM parametric analysis depends on laser power (P_l) layer thickness (t_l), scan speed (v_s) and hatch spacing (h_s). Overall, the highest part density of ~97% was achieved at three sets of parametric combinations as shown in Table 1.

Table 1. SLM process parameters used to additively manufacture pure silver tensile test coupons.

Part identifier	(W)	t (μm)	v_s (mm/s)	h_s (mm)	ρ_r
4T0.16	370	0.3	400	0.16	97%
4T0.20			400	0.20	
8T0.16			800	0.16	

To evaluate the feasibility of laser melting pure Ag and to identify optimum process parameters the atomised silver powder was laser processed on steel and copper substrates. The common practice for the selection of substrate materials was to use the same substrate as the processing material. However, this was not feasible for this study due to the high cost associated with pure silver. Accordingly, both copper and steel substrate were systematically evaluated and found

steel to be the most suitable at 370 W without any miscibility. Consequently, a steel substrate was used to manufacture all the Ag samples used in this study.

The selection of the most suitable SLM process parameter to print the bone scaffold was to consider not only the part density but also the mechanical performance. Although numerous SLM parameters exist, the laser power, scanning speed, hatch spacing, and layer thickness are widely accepted [83–85] as the most common process parameters that can be adjusted to optimise the process. For pure silver, the tensile test specimens conceived at the three parametric combinations as shown in Table 1 generated a ~97% dense material. The relative density observed for SLM pure silver can be seen to be at par with the highest ρ_r recorded for some of the emerging and reflective materials as listed in Table 2. In particular, the pure-Ag process parameters presented in this study delivers a ρ_r comparable to sterling silver and copper.

Table 2. Highest reported relative density for AM reflective materials compared to pure silver.

SLM material	ρ_r (%)	Ref.
Cu	97.00	Colopi <i>et al.</i> [86]
W	89.92	Deprez <i>et al.</i> [87]
Mg	82.00	Zhang <i>et al.</i> [88]
Au (24 Carat)	89.60	Khan and Dickens [89]
Sterling Ag (purity <92.5%)	96.70	Xiong <i>et al.</i> [51]
Pure Ag	97.00	This work

In addition to ρ_r , the resulting mechanical performance was also used as a benchmark to select the best SLM process parameters to build the scaffold designs. This strategy was chosen as it allows to consider the process parameters that simultaneously deliver both the highest relative density and structural integrity. Having tested numerous specimens, the highest and lowest stress-strain ($\sigma - \varepsilon$) curve observed for all of the process parametric combinations that delivered the highest (97%) relative density are presented in Fig. 4. The respective failed samples are shown in Fig. 6.

Analysing the $\sigma - \varepsilon$ curves (Fig. 4a-c) reveals consistent elastic and plastic regions for all the tensile coupons tested. However, despite featuring the same ρ_r (97%) for all the samples tested, a certain difference in the $\sigma - \varepsilon$ performance can be observed between each of the three process categories. The respective mechanical performance derived from each of the characteristic regions of the $\sigma - \varepsilon$ is summarised in Table 3. Overall, the highest Young’s modulus of 17.8 GPa was exhibited by the Ag specimen manufactured at a scan speed and hatch distance of 800 mm/s and 0.16 mm respectively. The Young’s modulus of 8T0.16 was closely followed by 4T0.16 and 4T0.20 at 15.6 and 15.5 GPa respectively.

From the $\sigma - \varepsilon$ data, it was clear that the hatch distance has a higher influence on the mechanical properties of dense pure Ag when SLM processed at 370 W. Even though, 4T0.16

and 4T0.20 featured the same v_s , the difference in h_s has contributed to the $\sim 0.6\%$ change in Young's modulus between the specimens. Overall, the highest difference of 13.8% in E was observed between the best (8T0.16) and worst (4T0.20) performing case demonstrating the influence of v_s despite the same h_s . Nevertheless, the difference in E was found to be negligible (0.6%) between the two cases sharing the same scan speed (400 mm/s). This shows that the scan speed has the highest influence on the stiffness of the dense pure silver manufactured using SLM.

A similar trend was observed when it comes to elastic limit, yield strength, ultimate strength, and ultimate strain, with 8T0.16 outperforming the other two cases. Out of all the mechanical properties evaluated, ε_{ult} showed the largest difference where 8T0.16 outperformed 4T0.20 by 42.3%. This demonstrates the significance of scan speed and hatch spacing on both the toughness and ductility of the SLM processed pure silver. In all the cases, the failure of the samples was consistent (Fig. 6) confirming the repeatability and validity of the observations.

Fig. 4. Stress-strain curve for 97% dense SLM pure-silver tensile coupons at different process parameters was (a), (b) and (c) shows the highest and lowest performance at the respective scan speed and hatch spacing; (d) compares the average performance between the three SLM process cases considered.

To put these observations into perspective the cumulative parameter, ‘laser energy density (e_{ld})’ has to be introduced [90–93]. It is widely considered as the critical parameter to characterise both the manufacturability [94] and mechanical performance [95] of SLM components and generally represents the energy density in J/mm^3 as shown in Eqn. (1):

$$e_{ld} = \frac{P_l}{v_s \times h_s \times t_l} \quad (1)$$

where P_l is the laser power (Watt), t_l is the layer thickness (mm), v_s (mm/s) is the scan speed and h_s (mm) is the hatch distance. Despite keeping P_l and t_l constant, 8T0.16 results in the lowest energy density of the three SLM process combinations considered. This subsequently affects the thermo-physical properties during laser melting and hence affects the mechanical properties of the manufactured parts.

Table 3. Mechanical performance of SLM manufactured 97% dense pure silver, where E , σ_e , σ_y , σ_{ult} and ϵ_{ult} are the Young’s modulus, elastic limit, yield strength, ultimate strength, and ultimate strain respectively.

Identifier	E (GPa)	σ_e (MPa)	σ_y (MPa)	σ_{ul} (MPa)	ϵ_{ul}
4T0.16	15.6±0.2	52.63±1.48	61.68±1.76	88.42±2.84	0.085±0.0046
4T0.20	15.5±0.4	50.71±2.38	57.56±1.94	82.43±2.74	0.071±0.0035
8T0.16	17.8±0.4	57.74±1.59	67.15±1.61	95.92±2.53	0.101±0.0002

According to Munir and Wen [96], the energy density at the powder bed during SLM primarily governs the microstructure and properties of as-built parts, which subsequently relate to the resulting mechanical performance. This energy density can be measured as the average applied energy from the laser per volume of material during the scanning of an individual layer. h_s is a measure of the separation between two consecutive laser tracks: centre-to-centre distance from one to the next laser scan (Fig. 5a). Scan spacing is directly proportional to the production speed. If it is high, it will take less time for the laser to scan the layer, if low, additional scans were required to execute the whole layer. Even though, an h_s of 0.16 and 0.20 mm in this instance delivered a ~97% part, the density at the higher h_s of 0.20 may have been composed of partly melted or sintered powder, which explains the lower mechanical performance. While at this stage it was merely a hypothesis, X-ray nCT analysis was consulted in later sections of this paper to further explore this aspect.

During the heating and melting of pure silver, consideration should also be given to heat capacity and latent heat [97]. These are heavily dependent on the material and proportional to the mass being melted. Accordingly, insufficient energy as a result of high v_s (>800 mm/s) leads to balling due to lack of wetting of molten pool with the preceding layer; low v_s (<800 mm/s) results in extensive material evaporation and the keyhole effect. In any case, un-optimal h_s leads to regular porosity as a result of insufficient melt overlap preventing the material layers from fusing as

shown in Fig. 5b. This was why despite the three parametric combinations resulting in similar relative density the mechanical performances were different.

Fig. 5. The creation of regular porosity throughout the structure as a result of un-optimal hatch spacing (h_s), where (a) shows two adjacent SLM material beads using an optimal h_s and (b) trapped unmelted material causing porosity as a result of un-optimal h_s .

Vaporisation [98] as a result of high e_{ld} during melting was another aspect which affects the performance. This leads to condensation of volatilised materials resulting in fewer but large pores, which explains the lower performance of 4T0.16. Hence, a suitable combination of P_l , v_s , h_s , and t_l is essential to manufacture repeatable and high-quality SLM parts. Accordingly, for pure silver, the SLM process parameters associated with 8T0.16 summarised in Table 1 delivers the best performing dense parts. As a result, this combination of SLM process parameters was chosen to manufacture the scaffold prototypes considered in this study. Nevertheless, to fully explain the reasons for the difference in mechanical performance despite similar ρ_r , a fracture surface analysis and defect characterisation were carried out.

3.1.2. Fracture and defect characterisation

Analysing the $\sigma - \varepsilon$ curve (Fig. 4) for as-built Ag in conjunction with the failed samples (Fig. 6) further; it was clear that the strain associated with failure was representative of a semi-ductile metallic material. A semi-ductile classification was most appropriate as the performance and associated failure of the SLM pure Ag exceeds a 5% strain at fracture (ε_{ult}) but falls close to or below 10% at failure. The reason for moderate ductile behaviour can be identified from the SEM fractography of the failed samples which reveals a semi-ductile fracture as shown in Fig. 7.

This is different from the pure ductile performance usually exhibited by traditionally manufactured silver. As it can be seen, an appreciable amount of permanent plastic deformation accompanies the fracture that was largely manifested macroscopically across the cross-section in the form of distortion or micro-necking. The normal practice is to test at least five tensile samples of a material to ensure a consistent stress-strain curve. However, for the case under consideration, only two samples of each variant of the tensile specimen were tested. This was primarily because of the high cost associated with pure silver. Therefore, it is crucial to evaluate the failed samples to see whether the failure is consistent. Overall, the failure mode can be seen to be consistent as shown in Fig. 6.

Fig. 6. Failed 97% dense pure-silver tensile-test coupons under different SLM process parameters.

The cross-section of the fracture surface shows a specific pattern where dimples of various size, shape, and distribution can be observed (Fig. 7). Failure was dominated by the growth of coarse voids between dimpled peaks consecutively being strained to failure. This can be seen to create lower size secondary dimples showing an assigned distribution of pores which have consistently grown during plastic deformation. The existence of un-melted or partially sintered Ag particles at the fractured surface explains the lower mechanical performance despite a 97% density. Despite the elongated peaks characteristic of a ductile behaviour, brittle ‘river markings’ [99] can also be observed in Fig. 7 explaining the comparatively low ε_{ult} . For a traditional classification of brittle failure based on $\sigma - \varepsilon$ curve, a plastic strain less than 5% is called for by Callister and Rethwisch [100]. Consequently, a semi-ductile failure is the most appropriate classification for 97% dense laser melted 99.9% Ag.

To further understand the reason for the semi-ductile behaviour a representative section of the tensile specimen was scanned using the X-ray nCT as shown in Fig. 8. The use of X-ray nCT for the non-destructive measurement of density and porosity defects and distribution is an emerging practice that applies particularly well to AM parts. According to Li and Fu [101], semi-ductile fracture of metallic materials is influenced by numerous factors including material state, workpiece geometry, and strain path. Under tension, semi-ductile fracture starts as a result of nucleation growth, and coalescence of micro-voids. Accordingly, looking at the X-ray nCT data of the SLM pure Ag dense samples as shown in Fig. 8, a certain distribution of micro-voids was visible. This not only explains the rapid reduction in mechanical performance but also observed fractography leading to semi-ductile behaviour.

Fig. 7. SEM fractography of the failed tensile sample surface.

Overall, it can be confirmed that the fracture mechanism involves the linkup of voids from multiple planes. Accordingly, the semi-ductile fracture occurs as a mixture of tension- and porosity induced damage. This explains the semi-ductile behaviour where the early nucleation was initiated as a result of crack propagation initiated by the voids representative of the ones shown in Fig. 8. The propagation of large nucleation cracks requires much less energy than pure-ductile crack and can occur at applied stress much lower than that at which failure would normally be expected. In other words, a high localised stress concentration facilitated by the pre-existing micro-voids was the reason for the observed semi-ductile behaviour and the associated mechanical performance.

Fig. 8. Reconstructed X-ray nCT showing selected cross-sections of dense SLM pure silver prints with pores highlighted.

Even though this is the first study on the additive manufacturing of pure Ag, the mechanical performance outperforms the performance of low temperature sintered silver carried out by Bai *et al.* [102]. While Bai reported an E and σ_y of 10 GPa and 43 MPa respectively, the SLM process carried out in this study resulted in an E and σ_y of 17.8 GPa and 67.15 MPa respectively, which accounts for an improvement of 56% and 43% for Young's modulus and yield strength respectively. Overall, it is clear that the mechanical performance of AM pure Ag is largely dependent on the SLM process parameters and the resulting part quality. For 99.9% pure Ag, the parameters associated with 8T0.16, summarised in Table 1 delivers the best performance and hence used to manufacture the scaffold prototypes considered in this study. Having sufficiently characterised the dense material, the manufacturability and performance of what can be considered as the first-ever AM pure silver bone scaffolds are brought forward in the subsequent sections.

3.2. Additively manufactured pure silver porous bone scaffold

3.2.1. Selective laser melting

To assess the suitability of additively manufacturing geometrically porous pure Ag bone scaffolds, two designs were conceived in this study. A low and high porosity cellular structure at

68.5% (*wt.*) and 90.7% respectively; the two versions were conceived to emulate a thick and thin strut thickness representative of porous structures suitable for bone ingrowth informed by previous studies [64]. The additively manufactured scaffolds prototypes using optimal process parameters (8T0.16) in pure Ag are shown in Fig. 9.

Fig. 9. As build AM pure Ag bone scaffold where SC90 and SC68 feature a porosity 90.7% and 68.5% respectively.

A rough surface finish as a result of characteristic stair-step effect during SLM was observed. The resulting morphology of as build Ag scaffolds is as shown in Fig. 10. While the interconnected porosity and the representative geometry closely followed the CAD (ideal geometry), the average strut thickness of SC90 was found to vary by approximately 50 μm . While this is in line with existing literature [103–106] on SLM of thin struts (<300 μm), the comparatively low mechanical performance of silver means that the effects can be significant.

It can be seen from Fig. 10 that despite the existence of semi-molten particles on the surface, SC60 features significantly low thickness variations in comparison to SC90. This clearly shows that even though SLM has clear advantages fabricating complex shapes, there are still challenges when it comes to thickness closer to or less than 300 μm . Consequently the rational design of pure Ag scaffolds requires aspects of the SLM process to be considered. Even though, this was a case explored by some researchers [104,107,108], most porous models do not consider manufacturing artefacts nor implement the manufacturing limitations at the design stage. Studies carried out by Zadpoor [109] also shows this to be a critical aspect if ignored can significantly worsen the mechanical performance of additively manufactured porous structures.

Fig. 10. Morphology of the as printed pure Ag scaffolds using SLM where (a) shows SC68 and (b) SC90.

3.2.2. Mechanical performance of Ag bone scaffold

Before the experimental evaluation of the mechanical properties, Ashby's criterion [65,66,110] for porous structures was used to characterise the relative mechanical performance of the scaffolds. The material properties (Table 3) of SLM Ag at a scan speed and hatch distance of 800 mm/s and 0.16 mm was used as the bulk Ag properties for Ashby's model. The scarcity of literature on the mechanical performance of SLM Ag means that the relative characterisation was crucial in assessing the influence of the AM process on the resulting mechanical performance of the scaffolds. Porous structures can be divided into either stretch or bend dominated depending on their bending/stretching of the cell struts. Similar to a honeycomb structure, both SC90 and SC68 were bend-dominated. The subsequent evaluation under this criterion using Ashby's model results in the theoretical mechanical performance of the scaffold as listed in Table 4.

Table 4. Relative and experimentally measured mechanical performance of SLM pure Ag bone scaffolds.

Scaffold	ρ_{rd}	φ (%)	σ_r (MPa)	$\sigma_{y (exp)}$ (MPa)	E_r (GPa)	E_{exp} (GPa)
SC68	0.32	68.46	37.71	44.58±5.2	1.77	1.86±0.08
SC90	0.09	90.69	20.49	10.16±0.8	0.15	0.13±0.02

According to Ashby's criterion, porosity (φ) has an inverse relationship with both relative density, elastic modulus, and strength. SC68 exhibited a relative modulus and strength of 1.77 GPa and 11.89 MPa respectively at a relative density of 0.32. When the relative density changed to 0.09, a ten-fold reduction in strength was observed resulting in 1.91 MPa in comparison to SC68. When it comes to the relative elastic modulus, a 90% reduction was observed between SC68 and SC90. This means that a high porosity structure using pure Ag may not be suitable for load-bearing applications unless suitable alloying elements can enhance the structural integrity. Nevertheless, the yield strength and elastic modulus were in the range of human cancellous bone suitable for reducing the effects of maladapted stress concentration and stress shielding according to Helsen and Missirlis [111]. Although the elastic modulus of porous pure Ag scaffolds presented here is below the stiffness of most human cortical bone [112].

Now that the theoretical performances are established for the scaffolds, the experimental $\sigma - \varepsilon$ performance for SLM pure Ag scaffolds are shown in Fig. 11 with the corresponding performance parameters summarised in Table 4. Generally, the stress-strain curve for cellular metallic materials represents three characteristic regions representing both the elastic and plastic performance. The first characteristic regions show both the linear and non-linear elastic deformation, where the $\sigma - \varepsilon$ proportionality is followed by a non-linear region ending in a stress peak commonly referred to as the ultimate strength (σ_{ult}). However, due to the considerable ductile behaviour of Ag, neither σ_{ult} or ε_{ult} could be registered. A case similar to what was observed by Wauthle *et al.* while studying AM porous tantalum implants [58]. In both cases,

the reason for this was the ductile behaviour of the material influencing what is otherwise a clear distinction between elastic and plastic performance.

Fig. 11. Quasi-static compressive performance of Ag SLM bone scaffolds showing (a) SC90 and (b) SC68.

It can be seen that as the compression strain increases, the scaffolds were getting plastically deformed (crushed) subsequently undergoing densification. Accordingly, the three distinct regions of the $\sigma - \varepsilon$ curve can be identified as: (i) the linear elastic region followed by (ii) the plateau and finally (iii) densification where a steep rise in stress can be observed. The performance at the plateau region was a result of the interaction between the crushed material. As a result, two distinct slopes can be dictated by the respective porosity of scaffolds being tested. Therefore Fig. 11a is representative of a highly porous structure while Fig. 11b represents a comparatively dense structure characterised by the shorter and steeper plateau region.

Looking at the experimental performance parameters for the pure Ag SLM scaffolds, the highest elastic modulus (E_{exp}) was exhibited by SC68 followed by SC90 at 1.86 and 0.13 GPa respectively. This performance trend was anticipated as SC68 features a lower porosity in comparison to SC90 at 68.46% and 90.69% respectively. The trend was similar for yield strength (σ_y), where SC90 showed the lowest performance at 10.16 MPa and SC68 the highest at 44.58 MPa. Comparing this to the theoretical performance calculated using relative density (E_r and σ_r), the experimental performance was representative of the porosity. Even so, a difference of 5% and 17% was observed between both the methods for the elastic modulus and compressive stress respectively at a relative density of 0.32. In comparison at the lower r_d of 0.09, Ashby's model overestimated the elastic modulus by 14% and compressive strength by 67%. This significant variation between experimental and theoretical performance at the highest porosity of 90.69% may signify structural anomalies introduced during SLM which is further explored through X-ray nCT.

3.2.3. Failure modes

One of the earliest studies on the effects of AM process-induced imperfections on the mechanical performance of SLM biomaterial lattice structures were studied by Campoli *et al.* [84]. The study showed that the weak spots in the geometry as a result of SLM substantially reduces the mechanical properties as compared to what was predicted by analytical models. It is important to realise that quality (continuity of the bulk material) of any SLM component depends not only on the process parameters but also on the part orientation with respect to the powder bed [58,113]. To investigate this aspect selected still frames recorded under quasi-static compression were analysed as shown in Fig. 12. As can be seen, chosen frames correspond to the compressive strain between 10% and 60%. From the deformation pattern exhibited by the structures, two distinct failure modes can be identified. Overall, it is clear that the porosity of the scaffolds has a significant effect on the part quality as SC90 clearly shows localised failure.

Fig. 12. Deformation and failure associated with pure Ag scaffold prototypes where (a) shows SC90 and (b) SC68.

Looking at the deformation of SC90 (Fig. 12a), the failure was dominated by weaker links at the lowest layer. As a result, a buckling induced failure can be observed starting from the first layer where multiple weaker links buckles to initiate plastic failure. To identify the reason for this localised failure, X-ray nCT of the scaffolds was conducted as shown in Fig. 13. This data helps to determine any manufacturing defects that may have been induced during SLM. It is clear from Fig. 13b that an SLM induced geometrical irregularity exists acting as a weak link at the first layer of SC90. The location of the link in Fig. 13b corresponds with the failed band and buckled beams explaining the failure modes observed as shown in Fig. 12a.

In addition to mechanical performance, the failure mechanism of SLM materials is dependant not only on the process parameters but also on the porosity and thickness of the struts. Most importantly thin-walled geometries are more prone to SLM induced defects compared to thick-walled structures. When it comes to the low porosity variant SC68, Fig. 13a shows that the scaffold prototype has good structural integrity and as a result shows failure due to barrel-shaped bulging (Fig. 12b) as opposed to localised failure. The X-ray nCT data reveals a dense part which is performing close to theoretical prediction. This shows that while laser melting pure Ag

using the process parameters reported, the unit cell porosity and wall thickness has a significant impact on the mechanical performance. Consequently, further refinement of the process parameters and powder characteristics are required to conceive geometrically accurate parts $<300\text{ }\mu\text{m}$ at 90% porosity. However, the parameters presented in this paper are capable of manufacturing highly repeatable and geometrically precise parts for beam/wall thicknesses above $300\text{ }\mu\text{m}$ up to porosity of $\sim 70\%$.

Fig. 13. Reconstructed X-ray nCT data of the SLM pure Ag scaffolds showing (a) SC68 and (b) SC90.

3.2.4. Chemical composition and morphology

According to Weber [114] micro/nano characteristics such as surface roughness, topology, and pore interconnectivity are important parameters for bone-interfacing scaffolds and hence influence osseointegration. Therefore, in addition to mechanical properties such as elastic modulus and strength, the surface itself has to provide the vital interconnectedness suitable for the migration and proliferation of osteoprogenitor cells or nucleation sites for *de novo* bone formation within the scaffold. To this end, the Ag scaffolds were SEM analysed to study the surface properties as shown in Fig. 14. While the figure shows the existence of multiple open porosities that allows for osteoblast incubation, the EDS analysis showed the presence of higher oxygen for areas identified by spectrum 2 as listed in Table 5. Looking at the data, these scattered areas of higher oxygen content can be identified as shown in Fig. 14b, which looks like spatter and oxidation reactions occurring at the laser melting course.

Looking closely at the melt pool during melting and solidification of pure Ag is critical to comprehend the reason for the presence of oxygen spatter. The EDS data shows that some Ag spatter was enriched with surface oxides resulting in a higher content of oxygen residue. According to Das [98], the stability of the melt pool and the melting itself depends largely on oxide formed as a result of laser melting. From this perspective, considering the O_2 (diatomic oxygen) partial pressure at the laser melting environment, it was plausible that the $\sim 962^\circ\text{C}$ temperature pure Ag melt pool triggered the oxide film formation on certain spatter. In any

case, the occurrence of oxygen-enriched spatter was found to be sparse and was identified only on the top and bottom surfaces of the scaffold parallel to the powder bed.

Fig. 14. Surface morphology of the scaffold walls showing the respective sections used for EDS analysis where (a) shows SC68 and (b) SC90.

Table 5. Elemental composition derived from EDS spectrum of the respective regions as shown in Fig. 14.

Element	Spectrum 1		Spectrum 2	
	Weight%	Atomic%	Weight%	Atomic%
O	0.73	4.75	5.80	29.35
Ag	99.27	95.25	94.20	70.65
Total	100	100	100	100

To put these observations into perspective, Louvis *et al.* [115] suggest that oxide layers generally found on metals such as aluminium are also stirred at the melt pool, however, the subsequent effects are negligible. The reason for this according to Verhaeghe *et al.* [116] was that the SLM process does not vaporise thicker oxides layers present in these materials due to the limited energy density. Overall, little has been studied on material evaporation during powder bed fusion. Nevertheless, studies conducted on laser welding shows that evaporation influences laser absorption, which in turn gives rise to fumes that were deposited on the component as oxygen-enriched laser spatter [117,118]. This shows that further optimisation of the chamber environment is required to reduce oxygen contamination. However, this change was primarily present on the surface alone and no change in elemental composition or Ag phase change was observed at the melt pool. These spatter spots can also be removed through effective post-processing such as abrasive blasting before implantation.

The mean surface roughness of the as-built pure Ag scaffold measured across the surface boundaries was found to be 51.4 μm . Any unmelted or loose particles on the scaffolds were removed through post-processing by sandblasting followed by cleaning with compressed air. The surface roughness characterisation of the scaffolds was characterised following the post-processing. A rough surface finish was exhibited by SLM scaffolds as a result of partially molten Ag particles that were attached to the surface. In addition, the laser scan marks characteristic of the SLM process representative of the layer-by-layer manufacturing were also identified.

Previous work [119] show that surface roughness of 40 μm is effective in accommodating osteoblasts responsible for bone growth. Even so, it is worth noting that osteoblasts come in numerous sizes and mature osteoblasts can grow over 200 μm . Therefore, the combination of small pores facilitated by the rough surface combined with geometrical porosity satisfies the requirements for osseointegration.

3.2.5. Antimicrobial property

When it comes to strategies for the inclusion of antimicrobial properties into orthopaedic implants, a large proportion of the research experiments with metallic ions such as copper, silver, and zinc. The existing data [52,120,121] on these materials show that they offer broad-spectrum antimicrobial properties suitable for infection resistance. According to Croes *et al.* [121] amongst all of these different materials, Ag is the most promising as it features well-established antimicrobial properties and offers low cytotoxicity to osteoblast responsible for bone regeneration. The mechanism of antimicrobial response in Ag involves two primary phases: In the first phase Ag ions binds with the bacterial membrane causing damage resulting in the condensation of deoxyribonucleic acid (DNA); in the second phase the penetrated membrane generated ‘reactive oxygen species’ (ROS) exerts oxidative stress further extending the antimicrobial capacity.

Fig. 15. Antibacterial time-kill efficacy of additively manufactured pure Ag bone scaffold against *S. aureus*.

Nevertheless, it is worth noting that so far the studies on establishing antimicrobial properties on implants either feature porous implants loaded with Ag nanoparticles [25] or specialist biofunctionalised Ag coatings [28,59,122]. While these routes offer the potential for developing targeted antimicrobial treatments and implants, they do extend the time, cost, and complexity of the manufacturing process itself. In this regard, an alternative route that allows the direct additive manufacturing of Ag influenced antimicrobial scaffolds suitable for on-demand implants is warranted. To this extent, there is no information on the antibacterial performance of laser melted 99.9% pure Ag. Therefore to confirm whether the SLM pure Ag scaffold possesses any antibacterial efficacy; the material was investigated on *Staphylococcus aureus* (ATCC6538).

According to Jaurugei *et al.* [123], *S. aureus* is one of the most common infections causing bacterium in scaffolds. It falls under the Gram-positive classification which is associated with large amounts of soft tissue infections in humans. Infections as a result of Gram-positive bacteria are susceptible to beta-lactam antibiotics like penicillin meaning they have a high efficacy in acquiring antibiotic resistance (an example of this is the superbug MRSA which is a staphylococcus species). Although *S. aureus* infections can often be treated, when associated with scaffold they lead to invasive and life-threatening complications.

Besides, hospital-acquired *S. aureus* infections are further complicated as a result of the multidrug-resistance nature of the strains. They have also been reported to affect medical devices creating biofilms *in vivo* which further contributed to implant-associated infection [124–126]. According to Johnson *et al.* [127], even though there are other types of microbes that are capable of implant infections, *S. aureus* is one of the most common pathogens. Therefore, as an initial test to inform the infection-resistance of the additively manufactured SLM pure-Ag, a 24-hour time-kill experiment was carried out on *S. aureus*. Two SLM pure-Ag samples were investigated and the reduction in viable cells was plotted as the change in log viability for each time point compared to zero-time viability. An identical test was carried out on Dulbecco's Phosphate Buffered Saline (DPBS) as a negative control to fully associate the observance of any antibacterial activities to the scaffold alone.

The antibacterial analysis shown in Fig. 15 confirms that the laser melted pure-Ag scaffolds not only inhibits the growth but kills *S. aureus* bacterium that is commonly associated with biofilm formation and implant-related infection. It can be seen that the antibacterial activity of the scaffold starts around 4 hours and leads to a 90% reduction of *S. aureus* in 7 hours. Overall, a 99.9% reduction in *S. aureus* was achieved by the scaffold in 14 hours showing that the SLM pure Ag scaffolds show antibacterial efficacy validating the direction of the study. However, it is unclear whether the antimicrobial activities of SLM Ag are exclusively mediated by the release of silver ions (Ag⁺) or, instead, are due to a combination of surface and silver ion effects. This aspect needs further investigation.

This is the first study on the SLM of pure silver, consequently, the intention is not for it to be used as a standalone biomaterial. Rather, it is a first step in the development of a new class of antibacterial biomaterials that can be additively manufactured which does not require extensive post-processing such as coating. It is anticipated that the effective understanding of SLM parameters and the performance of pure Ag will accelerate the development of functional Ag alloys that delivers a good balance between mechanical, antimicrobial, and osteogenic properties.

4. Conclusion

Implant related infection is a serious concern and various research efforts are going on around the world to address this issue. To this extent, this study demonstrates the use of selective laser

melting to fabricate porous (68-90%) Ag bone scaffolds with antibacterial properties and reports the respective processing parameters for the first time. The morphological evaluation of laser melted pure Ag shows that the process parameters reported can deliver a relative density of 97%. The material exhibits Young's modulus and yield strength of 17.8 GPa and 67.15 MPa respectively when printed at a scan speed, hatch distance, and layer thickness of 800 mm/s, 0.16 mm and 0.3 μm respectively using a 370 W laser system. Nevertheless, the X-ray nCT scan analysis revealed the presence of porosity defect, which contributed to the fracture mechanics of the tensile coupon. A high localised stress concentration facilitated by the pre-existing microvoids was the reason for the observed semi-ductile behaviour and the associated mechanical performance. The subsequently developed low (68%) and high (90%) porosity pure Ag bone scaffolds exhibited the yield strength and elastic modulus in the range of 44.58-10.16 MPa and 1.86-0.13 GPa. For the 68% porous scaffold, the theoretical prediction was conservative and underestimated Young's modulus and compressive strength by 5% and 17% respectively. However, at lower r_d of 0.09, the theoretical calculation overestimated the elastic modulus by 14% and compressive strength by 67%. This significant variation between experimental and theoretical performance at the highest porosity of 90.69% was identified using X-ray nCT analysis as the result of porosity defects introduced during SLM process. Nevertheless, with a yield strength of 44.58 MPa, an elastic modulus of 1.86 GPa, and a ductile deformation mechanism at 68% porosity, SLM pure Ag scaffold exhibits mechanical properties that are in the range of cancellous bone. While EDS analysis showed a 5.8% oxygen contamination at the surfaces of the pure Ag scaffolds, the antibacterial efficacy was unaffected. The SLM pure-Ag scaffold not only inhibited the bacterial growth but showed a 99.9% kill of the most common implant infection-causing *Staphylococcus aureus* in 14 hours. Overall, the study shows that SLM powder bed fusion is suitable to fabricate porous Ag bone scaffolds with interesting mechanical behaviour and promising antibacterial efficacy.

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