Additive manufacturing technologies of porous metal implants

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Abstract: Biomedical metal materials with good corrosion resistance and mechanical properties are widely used in orthopedic surgery and dental implant materials, but they can easily cause stress shielding due to the significant difference in elastic modulus between the implant and human bones. The elastic modulus of porous metals is lower than that of dense metals. Therefore, it is possible to adjust the pore parameters to make the elastic modulus of porous metals match or be comparable with that of the bone tissue. At the same time, the open porous metals with pores connected to each other could provide the structural condition for bone ingrowth, which is helpful in strengthening the biological combination of bone tissue with the implants. Therefore, the preparation technologies of porous metal implants and related research have been drawing more and more attention due to the excellent features of porous metals. Selective laser melting (SLM) and electron beam melting technology (EBM) are important research fields of additive manufacturing. They have the advantages of directly forming arbitrarily complex shaped metal parts which are suitable for the preparation of porous metal implants with complex shape and fine structure. As new manufacturing technologies, the applications of SLM and EBM for porous metal implants have just begun. This paper aims to understand the technology status of SLM and EBM, the research progress of porous metal implants preparation by using SLM and EBM, and the biological compatibility of the materials, individual design and manufacturing requirements. The existing problems and future research directions for porous metal implants prepared by SLM and EBM methods are discussed in the last paragraph.

Key words: additive manufacturing; SLM; EBM; porous metal implant; biocompatibility

Biomedical metal materials with high corrosion resistance, strength and toughness are widely used for bone surgery and dental implant materials. Conventional biomedical metal materials such as Ti6Al4V, 316L, CoCrMo and CoNiCrMo, etc. are mainly used for preparation of bony tissue under high load and complex stress conditions, such as bones, joints, and teeth. Most metal implants used for clinical practice are fully dense, but their elastic moduli are obviously higher than that of bony tissues. This means that loadings can't be transferred efficiently from metal implants to the adjacent tissue, leading to stress shielding and bone resorption.
shielding between implants and bones. As a result, the stresses are always absorbed by the implants, causing atrophy and osteo-necrosis of bones around the implants, distortion of new bone and reduction of bearing capacity of connective parts [3]. Also the smooth surface of dense metal implants makes it difficult to combine well with the host bone. We noticed that surface treatments, such as sand blasting, spraying, or sintering, could increase the roughness and surface area of implants [3] and improve the adhesion strength of the implants with the host bone to a certain degree. However, the state of bone tissues covering around the implants is not changed and the low bonding strength problem has not been essentially solved. The elastic modulus of porous metals is lower than that of dense metals. By adjusting the pore parameters, the elastic modulus of the porous metals can be made to match that of the bone tissues [4]. The open and connected pores of the porous metals can provide space and channels for bone ingrowth, which helps bone tissue growth from the surface into the interior of the implant, strengthening the biological combination between bone tissues and the implants. The preparation technologies of porous metal implants and related research have drawn more and more attention due to the excellent features of porous metals [5].

At present, the preparation methods of porous metal implants mainly include space-holder, polymeric sponge dipping, self-propagation high-temperature synthesis and chemical vapor deposition [6-11]. However, for the bone repair implants, the size varies from one person to another, the shape is complicated and full of tiny details, making it difficult to flexibly control pore structure and liberally manufacture implants by considering different bone defect forms. Such additive manufacturing technologies as selective laser melting (SLM) and electron beam melting technology (EBM) are suitable for fabricating porous metal implants. The two technologies are critical in satisfying the individualized requirements. This is because they can not only produce adjustable elastic modulus and open porous metal implants, with pores connected to each other, but also give the implants complicated interior and exterior shapes simply, rapidly, and accurately. The application of SLM and EBM in the field of additive manufacturing for fabricating porous metal implants is just getting started. Reviewing the research progress, understanding the problems and further research directions for porous implants will provide reference and enlightenment for further study of porous metal implant fabricated by additive manufacturing.

1 The technology status of SLM and EBM

Additive manufacturing technology (also known as rapid prototyping or 3D printing technology) is based on the idea of discrete accumulation forming. This applies the principle of material accumulating fabrication to manufacture complex three-dimensional parts under computer control by virtue of CAD data and model. It is a "bottom to top" material accumulation manufacturing method [12] compared to the traditional material cutting or machining technology. Such a method therefore can substantially shorten the processing because it omits the cutting tool, fixture and multi-step forming processes. In addition, the more complex the product shape, the more obvious the gain in manufacturing speed [13]. Additive manufacturing technology became available in the late 1980s. Since then, about 20 kinds of technology have been developed, such as Stereolithography (SL), Laminated Object Manufacturing (LOM), Laser Cladding Forming (LCF), Selective Laser Sintering (SLS), Selective Laser Melting (SLM), and Electron Beam Melting (EBM). These are found in use in the fields of automation, consumer electronics, medical/dental, industrial machinery, aerospace, etc. The application of additive manufacturing in the medical/dental field shows a rising trend with increases of 13.6%, 15.9% and 16.4% in the years 2011, 2012, and 2013, respectively, as shown in Fig. 1.
SLM and EBM are two kinds of precision forming technologies which are suitable for making metal parts with complex structures. The forming process of SLM is operated in an inert gas chamber, in which a laser scans the metal powder supplied by a powder spreader according to the designed paths. The bulk form is produced by metallurgical bonding of the metal powder layer by layer (Fig. 2) [14]. The EBM forming process is operated in a vacuum environment and the bulk form is produced by an electron beam in a similar way to the SLM method (Fig. 3). The main difference between SLM and EBM is the heat source, i.e., a laser for SLM and an electron beam for EBM. The advantages of the SLM method are the more developed technology, cheaper equipment and higher forming precision, due to a smaller particle size of the powder condensed by the small diameter of the laser spot. However, the forming speed of EBM (80 cm$^3$·h$^{-1}$) is 4 to 5 times faster than that of SLM resulting from its high energy (as high as 3000 W, 10 times larger than that of SLM) supply and larger particle size. Because of the larger particle size, the forming precision of EBM method is relatively low.

SLM was proposed by the Fraunhofer Institute in 1995 and the SLM system was first developed by the MCP Company in Germany in 2003. Since then, commercial equipment has been developed by many German companies, including EOS, Concept Laser, and ILT. A commercial SLM machine named EOSINT M270 equipped with a 200 W solid state Yb fiber laser was recently developed by the EOS Company. The spreading thickness of the powder layer is 20 to 100 μm. The spot diameter is 70 to 200 μm. The typical scanning speed is 750 mm·s$^{-1}$, the dimensional accuracy of parts is 20 to 80 μm, and the surface roughness of parts is 10 to 15 μm. Parts made by SLM equipment have been successfully used in aerospace, automotive, medical, injection molding, and other fields. In China, the Rapid Manufacturing Center of Huazhong University of Science and Technology launched a diode pumped 150 W YAG laser and SLM devices using a 100 W fiber laser in 2003, as a prelude to China’s research on SLM [15]. In 2007, South China University of Technology developed DiMetal-280 type of SLM prototype based on DiMetal-240 equipped with a Yb fiber laser. The spot diameter is 50 to 200 μm, the typical scanning speed is 200 to 600 mm·s$^{-1}$, the dimensional accuracy is 20 to 100 μm, and the surface roughness is 20 to 30 μm. Beijing Institute of Aeronautical Manufacturing Engineering developed China’s largest SLM equipment (type LSF-M360) with forming dimensions of 350 mm × 350 mm × 400 mm. It seems that the research on SLM equipment and technology has made some progress in China. However, the key components of SLM equipment, such as the laser, focusing mirror, and high-speed scanning galvanometer are still dependent on imports.

EBM research began in the early 21st century. In 2003, the Swedish Arcam Company independently developed the earliest EBM for commercial use in the world. The largest forming dimensions are 200 mm × 200 mm × 160 mm. The spreading thickness of the powder layer is 50 to 200 μm. The scanning speed is over 1000 m·s$^{-1}$. The beam spot diameter is 300 to 500 μm. The dimensional accuracy is 200 to 400 μm [16]. Many research institutes and universities in the USA, Japan, Britain, etc. have purchased EBM equipment from the Arcam Company to carry out research in multiple fields. At present, more than 100 EBM units are in use around the world. In 2004, Tsinghua University developed the first domestic EBM prototype (type EBSM150), with a maximum scanning speed of 2 m·s$^{-1}$ [17]. In 2008, Tsinghua University again developed the second generation EBSM250 electron beam forming equipment, and the largest forming dimensions are 230 mm × 230 mm × 250 mm. This is based on the research on several key problems related to EBM process, such as the powder preheating process [18], scanning path planning, and mechanical properties of the part [19]. It was reported that the load-bearing components for Ti6Al4V aircraft manufactured by the EBM process have excellent mechanical properties which equal to that of the parts made by a forging process[20]. Furthermore, the porous titanium alloy made by the EBM method has also stimulated the research interest in Northwest Research Institute of Non-ferrous Metals and much work has been done [21].

The development of additive manufacturing technology of
SLM and EBM is supported by the innovation of equipment, the advancement of the forming process and the development of metal powder. The preparation of metal powder materials is a very important part in the development of additive manufacturing technology. Additive manufacturing has special requirements for the metal powder, such as small particle size, narrow particle size distribution, high sphericity, good liquidity, and low oxygen content. At present, the general preparation methods of metal powder for additive manufacture include Plasma Atomization, Plasma Spheroidization, Vacuum Induction-melting Inert Gas Atomization (VIGA) and Electrode Induction-melting Inert Gas Atomization (EIGA). The varieties of the metal powders are fewer. At present, only Ti6Al4V, CP-Ti, In625, CoCrMo, and 316L are available on the market. AP&C Metal Powder Division of Canada Raymor Industries Company reserves the proprietary right of plasma atomization technology and is one of the suppliers of the main metal powders. Its products are characterized by a high degree of sphericity, high purity, low oxygen content, uniform particle size distribution, and high powder density. The metal powders being used by the Swedish Arcam Company come from the company. The USA’s Ametek employs hydrogenation and dehydrogenation plasma spheroidization and gas atomization to prepare spherically shaped metal powders with diameters of about 100 μm (Fig. 4), while ALD Vacuum Technologies of Germany uses VIGA and EIGA to produce high-quality metal powders marked with fine particle size and narrow particle size distribution (Table 1). However, in China, the high quality metal powders needed for additive manufacturing depend on imports.

![Fig. 4: Additive manufacturing metal powders provided by Ametek: (a) Ti6Al4V powders using Plasma Spheroidization and (b) 316L powders using Gas Atomization](image)

### Table 1: Technical data of additive manufacturing metal powders using VIGA and EIGA

<table>
<thead>
<tr>
<th>Technical data of metal powder</th>
<th>VIGA</th>
<th>EIGA</th>
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<tr>
<td>Typical powder alloy for metal additive manufacturing</td>
<td>CoCrMo, In625, In718</td>
<td>CP-Ti, Ti6Al4V, TiAl</td>
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<tr>
<td>Powder morphology</td>
<td>Spherical</td>
<td>Spherical</td>
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<tr>
<td>$d_{50}$ [PSD mass median]</td>
<td>35–70 μm</td>
<td>60–100 μm</td>
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<td>+10–45 μm</td>
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<td>Typical available size classes for metal additive manufacturing</td>
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<td>+45–63 μm</td>
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2 The progress of porous metal implants made by SLM and EBM

Suitable pore structure, such as the dimension, shape, distribution, spatial direction and mutual connectivity, is conducive to the hard and soft human tissues ingrowth around the implants. The mutual chimeric and the biocompatibility could be realized between implants and human tissues. The porous structure also could reduce the elastic modulus of metal implants. Pore structure, therefore, is the key factor for mechanical compatibility and biocompatibility of bone implants with the host bone. Reports on mechanical compatibility and biocompatibility of pore structure design and related research are seldom found. The recent study on the porous metals made by SLM or EBM is only based on the honeycomb or crystal lattice structures, without a focus on the analysis of...
design principles and pore structure criteria. In addition to the pore structure of implants, connection problems between the implants and host bone with inhomogeneous porous structure (Fig. 5) still need to be considered. Even if the whole structure of the implant were porous and the elastic modulus of the implant matched that of the host bone, it is still possible that requirements for the strength of load-bearing parts and the stability of the connection between the implant and the host bone can not be satisfied. Therefore, an implant with a gradient pore structure was brought into application (Fig. 6).

A porous Ti6Al4V acetabulum cup implant with “a porous surface and a dense inner layer structure” fabricated by EBM has been successfully applied in clinical practice (Fig. 7). The porous surface can not only provide a good environment for bone ingrowth and decrease the elastic modulus difference between the implant and host bone, but also reduce the stress shielding effect; meanwhile, the dense inner layer can satisfy the strength requirement of the load-bearing parts and the connection stability requirement of the implants and host bone. A report on the Ti6Al4V dental implant with a porous surface and a dense inner layer structure by Traini et al. of Chieti-Pescara University, Italy, using the SLM method (Fig. 8) showed that the rough surface of the implant is good for new bone ingrowth, and thus improves the connection strength between the implant and alveolar bone. It also showed that the elastic moduli of the dense layer and the porous surface layer are 104 ± 7.7 GPa and 77 ± 3.5 GPa, respectively, which reduce the elastic modulus difference between the implant and the human alveolar bone (the elastic modulus of a human alveolar bone is 20 ± 7.0 GPa), which is helpful in weakening the stress shielding effect. Lewis Mullen et al. at the UK’s University of Liverpool developed a porous titanium implant by using an SLM facility (made by the MCP Company in Germany) equipped with a 200 W Yb fiber laser. The relationship between laser power and pore size was determined. When the pore dimensions were 600 μm, 800 μm, 1,000 μm and 1,200 μm, the appropriate requirement for laser power was 62 W, 96 W, 132 W and 167 W, respectively. The ideal powder particle size of the feeding material through optimization should be smaller than 45 μm. The fabricated porous titanium implant was characterized by a porosity of 10% to 95%, a pore wall thickness of 150 to 420 μm, a compressive strength of 0.5 to 350 MPa and precise dimensions. In addition, a method to adjust the porosity of porous metal by controlling the beam overlap procedure using SLM was suggested by Stamp et al., and a porous metal implant with a porosity of 71%, average pore size of 440 μm, and compression strength of 70 MPa was obtained. It is worthy of note that Heinl et al. at FAU in Germany, have successfully fabricated a porous Ti6Al4V implant with an elastic modulus close to that of human bone by using EBM S12 made by the Arcam Company in Sweden. Recently an opening cup implant sample with different pore dimensions and porosity made with Ti6Al4V alloy was prepared using the EBM process. The pore dimensions and pore wall thickness are 765 to 1020 μm and 466 to 941 μm, respectively.

In China, there is less research on porous metal implants fabricated by SLM and EBM and there is only a small amount of literature about the investigation of pore structure design.
and forming techniques of porous metal implants. Based on the pore structure modeling method and the SLM manufacturing technology of porous implants, Dr. Xiao [35] of the South China University of Technology, proposed a method for an implicit surface which is represented by the implicit function of porous unit. According to the mechanical performance requirements of different defect locations of implant bone, different units can be constructed by changing the parameters of the implicit function. When the porous structure unit is mapped to the implant design domain, the design rules of porous structure based on SLM technology are then established. The porous structures of a hip joint and skull implants were designed and formed according to their individual requirements, which validate the feasibility of the design rules. Li [23] et al., of Shanghai Jiao Tong University, fabricated honeycomb porous Ti6Al4V implants using EBM equipment under the condition of vacuum < 0.5 Pa. The electron beam power was 4 kW, the processing layer thickness was 0.07 mm and the scan speed was 1 km·s$^{-1}$. Scanning electron microscope observation indicated that porous titanium alloy prepared in this way has a three-dimensional interconnected pore structure consistent with the design shape. It also found that such porous Ti alloy presents about 15% shrinkage.

3 Studies on compatibility of porous metal implants

The compatibilities of the metal implants with the human body mainly include mechanical compatibility and biocompatibility, which relates to many aspects. Two very important aspects are the matching degree of the mechanical properties between the implant and host bone, and the ingrowth condition for new bone provided by the implant. The stability to connect the implant and host bone, and the ability to induce new bone growth, are dependent on these two aspects.

A porous titanium implant [33] with diamond molecular structure was fabricated using the EBM process through the cooperation of Shanghai Jiao Tong University and the Institute of Metal Research of the Chinese Academy of Sciences. The implant was coated with a tantalum film using the Chemical Vapor Ceposition (CVD) method (Fig. 9). As a result, the porous titanium implant simultaneously has the combination of excellent mechanical performance and biological properties. The porosity, pore dimension and elastic modulus of the fabricated Ti6Al4V implant are 70.5±0.6 %, 700±50 μm and 11.3±0.4 GPa, respectively. The elastic modulus particularly satisfies the human cortical bone requirement. The compression strength, however, is lower than that of human cortical bone (Table 2) [36]. The USA Zimmer Company produced a porous tantalum dental implant

Fig. 8: SEM photograph of porous titanium dental implant fabricated by SLM

Fig. 9: A photograph of samples of (a) porous titanium implant; and (b) coated with tantalum
using the CVD method (Fig. 10). The porosity, pore dimensions and elastic modulus are in the range of 15% to 85%, 400 to 600 μm and 1.5 to 3 GPa, respectively. The elastic modulus of the implant is near to that of human cortical bone (Table 2) [37-39].

The biocompatibility of porous metal implants with the host bone depends on the pore structure and surface characteristics of the implant. It has been shown [40] that the ingrowth rate and the quantity of new bone into a porous metal implant are related to the pore structure of the implant. The pore structure characteristics can be described by the geometric parameters of the porosity and the pore diameter; the latter is the major factor influencing the growth of new bone. The results [41] given by Klawitter and Hulbert showed that the pore diameter should be larger than 100 μm in order to ensure blood circulation to the required bone tissue ingrowth. The growth speed and quantity of the new bone increase with the pore diameter. The results were further confirmed by Palmquist et al. [42] who embedded a titanium alloy implant with pore diameters from 500 μm to 700 μm into a sheep’s body for 26 weeks and found a lot of new bone tissues ingrown inside the implant. The porous structure of the implant is helpful to the new ingrown bone and improves the connection strength between the implant and the host bone. However, new ingrown bone cannot be induced if the surface of the porous metal implant is not modified by a coating treatment. Furthermore, the resultant corrosion products and pathological reaction after implantation in the human body will affect the normal biological function. Therefore, the coating modification treatment of the metal implants is a critical way of solving these problems. Nowadays, two kinds of coating materials, i.e. metallic and biological ceramic, are available for the coating of porous metals. One example is Ta coating fabricated by Li [36] who deposited tantalum metal on the surface of porous Ti6Al4V alloy using the CVD method. The tantalum coating combines well with the Ti6Al4V implant and does not possess biological toxicity caused by a metal ion dissolved into the blood. Therefore, it is bio-inert or lacks bioactivity in human body; but cannot induce new bone ingrowth. The biological ceramic coating on the implant surface is not only beneficial to improve the bioactivity, induce the ingrowth of new bone, and promote osseo-integration; but is also helpful to cut off the toxic substances released by the metal substrate. Apatite coatings are representative of biological ceramic coatings. A novel method called bionic solution was developed to coat apatite on a substrate with a complex shape and porous structure. The basic process of this method is that the implant is immersed into the simulated body fluid (SBF), and an apatite coating similar to human bone is induced spontaneously on the implant surface. The biomimetic solution method can provide a reference in exploring surface modification technology of porous implants fabricated by SLM and EBM. But for SBF, technology is still at the stage of laboratory research. Some problems, such as the low bonding strength between the coating and the substrate and the long cycle of growth for the coating, still need to be solved [43].

4 Individual design and manufacture of porous metal implant

The bone shapes and defect morphologies vary between different individuals. Porous metal implants should not only have the mechanical properties and pore structure suitable for the ingrowth of new bone, but also should have the characteristic of matching the individual’s shape.

Additive manufacturing is a simple, rapid method for the
fabrication of a porous metal implant with complex pore structure and matching shape close to the morphology of the bone tissue to be replaced. This is because the design and the manufacture of the individual porous metal implant are based on the data of the patient's own bone. First, the defective bone part of a patient should be scanned layer by layer by using CT imaging. If the bone is totally defective, mirroring images can be generated according to the complete bone data of the symmetrical parts. From this CT image data of bone, a multi-layer section contour can be obtained. Further, the CT data could be translated into 3D CAD data and an individual model can be generated using reverse engineering and computer technology. On this basis, the individual design could be realized and the implant could be fabricated using additive manufacturing. Although the individualized implants can be comparable to the structure and the appearance of bone tissue with the aid of CT scanning and additive manufacturing technology, the implementation process is too complex to complete without the help of professional image processing software. A famous and mature image processing software now being applied in the bone surgery field is named Mimics, which was developed by the Belgium Materialise Company. The Mimics is a set of reverse engineering software which is the bridge between the medical and the mechanical fields. Using this, the CT layer scanning image data can be quickly translated into CAD/CAM data format for the manufacturing of the defective bone parts and can be shown in highly accurate 3D model for design purposes. A porous artificial femoral model with individual features was built by Su et al. through reconstructing CT data in three-dimensions with the aid of the Mimics software. The porous Ti6Al4V artificial femoral implant fabricated using the SLM method is shown in Fig. 11. However, additive manufacturing technology used for individualized bone implant fabrication is still at the exploratory stage, and it still needs further clinical validation and strict examination and approval of the registration process before it becomes reality in practice.

Fig. 11: (a) Porous titanium femoral bone implant digital model and (b) physical photo

5 Existing problems and prospects

SLM and EBM are two kinds of simple, rapid additive manufacturing technology used to fabricate porous implant materials. These two technologies have unique advantages in precisely controlling pore size, spatial distribution, shape, etc. In these new developing additive manufacturing technologies of SLM and EBM, some problems still exist which urgently need to be solved. Therefore, further in-depth and meticulous research in terms of preparation of porous metal implants should be considered.

(1) The accuracy of the equipment, the quality of the metal powder and the level of the forming process are the key factors which influence the development of SLM and EBM technology. In China, research on SLM and EBM is still in its infancy; there is a long way to go considering the great gap between the developments of commercial equipment, metal powder and forming technology, etc. that already exists abroad. It is suggested that China should focus on the research and development of commercial equipment with stable performance, high forming precision, low cost, and independent intellectual property rights; and extend the species of metal powder to the requirement of additive manufacturing. Only if the development of high quality metal powder and the supporting forming processes are simultaneously promoted can the substantive breakthrough for the preparation of porous metal implants using SLM and EBM be realized.

(2) Much work has been done abroad related to the laser power selection, scanning path scheme design, powder particle size optimization and clinical evaluation in preparation of porous metal implant using SLM and EBM. Part of the work has resulted in products on a large scale. China is still at the exploratory research stage in the field of porous metal fabrication using the additive manufacturing method.
Researchers in China should actively carry out the research on pore structure design by considering both the mechanical compatibility and biocompatibility of porous metal implant, and pay particular attention to the development of porous implants with both gradient pore and bionic structures.

(3) The mechanical properties and biocompatibility of the porous metal implants are closely related to the pore diameter and other pore parameters. In order to optimize the pore structure, it is necessary, therefore, to investigate the influence of pore parameters on the mechanical properties and biocompatibility of the porous metal systematically and thoroughly.

(4) A biological ceramic coating on the surface of a porous metal implant can improve the biological activity. As a surface modification technique, the SBF method is suitable for forming a homogeneous ceramic coating on a complex and porous substrate. The problems for the SBF technology, however, are the low bonding strength between ceramics and the substrate and the longer growth circle of the ceramic coating. This should be regarded as one of the important research directions in developing the SBF technology.

(5) Fabrication of porous metal implants using SLM and EBM technologies is an interdisciplinary subject involving research on the thermo-physical properties, the melting and solidifying mechanism of the materials, processing techniques, compatibility with the human body and the clinical application of the implant, etc. This means that exchanges and cooperation between medical workers and materials workers are necessary in order to bridge the gaps and take advantage of the complementary disciplines. We expect that the technical difficulties will be surmounted as soon as possible in order to realize the application and popularization of porous metal implants.

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