Research on Producing Complex Metal Parts with Lattice Structure, by Selective Laser Melting

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Abstract. This study evaluates the manufacturability and performances of periodic cellular lattice structures designed by repeating a cubic unit cell and produced by SLM using titanium powder. The effects of unit cell size on the manufacturability, density, compression and bending properties of the manufactured cellular lattice structures were investigated. Lattice structures manufactured with various unit cell sizes ranging from 0.5 to 1.2 mm could be produced free of defects by the SLM process, with a novel type of supports. By the increasing of the cell size, a decrease of the applied load together with an enhancement of the flexure extension were observed. Specimens with a cell size higher than 1 mm manifested an excellent flexibility during flexure tests.

Introduction

Selective Laser Melting (SLM) is one of the few rapid prototyping (RP) techniques which enables the quick production of nearly unlimited complex geometries directly from metal powder with no or minimal pre-processing and/or post-processing requirements. SLM allows the manufacture of complex lattice structures which otherwise cannot be built by conventional technologies. A growing interest in the SLM-fabrication is the production of porous objects, because they manifest similar or superior properties to solid counterparts. For instance, porous objects usually have larger internal surface areas and higher strength-to-weight ratios. For this reason, interconnected pores can facilitate the passage of fluids and biological cells through which biological tissues can be gradually grown and docked [1].

Micro-lattice block structures offer a significant potential for the usage in the design of lightweight cellular structures. Experiments have shown that mechanical response of micro lattice structures is governed by the micro-architecture of the structure [2].

The research presents the production and testing of such lightweight and at the same time sustainable SLM-parts, with their possible applications. The focus is on the microstructure - mechanical properties–local deformations relationship, which allows the optimization of load adapted lattice structures. The final aim is to obtain personalized implants with mechanical properties close to those of bone and to ensure their osseointegration. The SLM is considered a viable alternative for achieving a precise control over the scaffolds architecture, the pore shape and especially interconnectivity, which is crucial for the bone ingrowth. The manufactured and tested samples offer important information about the degree of dependence of the compression and bending properties of the lattice structures on the various unit cell sizes.

Porous Structures Made by SLM

Yan et al. [3] investigated the design and manufacturing of periodic cellular lattice structures using a novel unit cell type called “Schoen Gyroid”, which possesses circular and smooth struts and a spherical core. The inclination angle of the circular and smooth struts of the Gyroid unit cell continuously varies along the spherical core, what leads to the fact that during the SLM process, the
previously manufactured layer can support next layer, and that the Gyroid lattice structures are self-supporting. Fig. 1 shows this special core-struts architecture, with the same volume fraction of 15% and unit cell sizes of 2, 3.5, 4.5, 5.5, 6.5 and 8 mm, which were manufactured from stainless steel powder (316L). The effect of unit cell size on the manufacturability, strut density and compression properties of the lattice structures made by SLM was investigated.

The researchers didn’t manufacture titanium based - lattices and there is no experience for cell sizes under 2 mm.

Fig. 1. Gyroid lattice structures
Fig. 2. BCC (a) and BCCZ (b) lattice structures

Smith et al. [4] manufactured from stainless steel powder (316 L), on basis of SLM, lattice structures based on two typical structures, with body-centered cubic structure (BCC) and a similar structure with vertical pillars (BCCZ) (Fig. 2). These blocks consist of periodic cells in a scaffold-like structure, in which each cell is based on a number of truss elements or beams connected to each other. The manufacturing technique and resulting structures are given by Mines [5]. The porosities were in the range 86.1 – 96 %. The predictions about using both analytical and beam element FE models demonstrated that the stiffness and yield strength could be improved by varying the unit cell geometry. The authors limited the research to the cell size-range 1.25 – 2.5 mm.

Customized Implants Requirements

Load-bearing bones are sometimes damaged due to diseases or injuries so severely that they have to be replaced by prosthetics, situation often found in surgeries. Conventional designs may not be able to meet the multiple requirements which are required for the load-bearing bone implants. A number of researchers developed some techniques to fabricate scaffolds, because the porous materials have an additional desired effect on promoting bone ingrowth [6]. Anatomically, the external geometry and size of the scaffolds should match to the bone defect of each patient so that the scaffolds can fit and anchor into the defect. The implant materials should be designed to meet simultaneous multiple requirements, such as sufficient stiffness and porosity. Usually, clinical failure of prosthetic solutions is due to stress-shielding generated by the mechanical properties discrepancy between bone and implant.

In order to achieve rapid and high volumes of bone ingrowth, the structures must be manufactured from a biocompatible material with many interconnected porosities, with a pore size in the range 100 - 700µm, porosity between 50% and 70%, compression strength >50Mpa [7].

Titanium is the most biocompatible and corrosion-resistant metal, its elasticity modulus corresponds to the elasticity modulus of the bone more than any other metal does and for this reason it can be used successfully for surgical implants.

Additive Manufacturing of Titanium Samples with Lattice Structure

For a better understanding of the way how the micro-lattices are built, Fig.3 shows the pattern of the most usual unit cell topologies.

The aim of the present work was to develop and characterize a specific porous structure, on basis of a BCC unit cell. For this purpose, the mechanical properties of titanium lattice structures manufactured by SLM were established through experimental testing. This paper is investigating the flexure and compression characteristics under quasi-static rates of loading.
Material and Methods. The most fundamental requirement of any lattice structure design is that it must be self-supporting. Supports are strictly required for the SLM process and must fulfill the following vital functions:

- releasing through deflection of the residual stresses which remained in the working part;
- the taking over and transmission of the released heat towards the building platform, during the metal powder melting process.

In this paper a novel lattice support structures for metal additive manufacturing is proposed. The particularity of these supports is that they are built with an uniform distribution on the whole surface of the samples (Fig.4). A significant advantage is that they can be easily removed using an usual cutter in the first stage from the construction platform together with the sample and then in the second stage, using a simple nipper, from the sample itself.

16 beams 7x7x40 mm and 16 cubes 5x5x5 mm, each type with cell sizes in the range 0.5 – 1.2 mm were manufactured using an MCP Realizer II SLM 250 - at the Technical University of Cluj-Napoca/ROMANIA. Commercially Pure Porous Titanium (CPPTi) powder was used, with a grain size distribution in the range 20 – 63 µm (TLS Technik GmbH-Germany). 75 W Laser power, 30 µm point distance (PD) and 60 µs exposure time (ET) as parameters for the machine and the new command “Point structure“ of the ReaLizer Software were used.

The porosity is calculated as follows:

\[ P = 100\% \cdot \left(1 - \frac{\rho_s}{\rho_0}\right) \]  

(1)

where \( \rho_s \) is the density of the sample, obtained by the ratio weight to volume calculated by the geometrical data of the specimen and \( \rho_0 \) the material standard bulk density.

The calculated porosity was in the range 50.8 % for cell sizes of 0.5 mm and 88.9 % for the cell sizes of 1.2 mm, as shown in Table 1.

<table>
<thead>
<tr>
<th>Cell size [mm]</th>
<th>0.5</th>
<th>0.6</th>
<th>0.7</th>
<th>0.8</th>
<th>0.9</th>
<th>1</th>
<th>1.1</th>
<th>1.2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Porosity [%]</td>
<td>50.8</td>
<td>63.7</td>
<td>71.4</td>
<td>76.4</td>
<td>81.2</td>
<td>84.5</td>
<td>97.8</td>
<td>88.9</td>
</tr>
</tbody>
</table>

Measurements, Results and Discussions. The specimens were tested as following: beams at bending and cubes at compression. For both type of samples, for all cell sizes, the load was applied on two different directions: 0°, respectively 90°, measured against the supports direction. In Fig.5, the 8 different lattices with their respective cell size can be easily observed. Only 3 representative cell sizes (0.5, 0.8, 1.2 mm) of the bended beams are represented in the picture. It could be noticed the macroscopic crack onto the sample with the 0.5 mm cell size, while the 1.2 mm cell size sample undergone rigid body movement during the bending tests.
The tests were carried out using an Instron 3366 testing machine (within the Technical University Cluj Napoca), with a 1000 kN load-capacity, under a constant loading rate of 0.5 mm/min.

**Flexure Tests.** For the flexure measurements the 3 points bending test was used. In the Table 2 flexure load at TS (Tensile Strength) and extension at TS values are presented. Some variations between the two different orientations (0° respectively 90°) could be observed. Some of the samples are more rigid for the 0°-position, other for 90°-position. It can be noticed that the samples with the cell size in the range of 0.5 mm to 1 mm/0°-position (with a maximum flexure load ranging from 1010 N to 87 N) cracked during bending tests and those from 1mm/90° to 1.2mm/90° (with a maximum flexure load ranging from 198 N to 139 N), could not be broken, the TS value being not reached. As a consequence, the values marked in italic style do not represent the values for flexure load at TS, but only the values where the machine has been stopped.

During the tests of those samples, it has been noticed that the specimens undergone rigid body movement, so that the load values presented in Table 2 correspond to an extension of 10 mm. The value of 198N for the sample 1mm/90° is related to the moment when the testing machine was for the first time manually stopped. For the samples with a cell size ranging 0.5 – 1mm/90°, the machine stopped by itself, at the TS values. From the figures presented in Table 2, it can be observed an increase of the flexure extension together with the increase of the cell size, as the applied load decreases.

**Compression Tests.** For the compression tests, because the cubes dimensions were small (5x5x5mm), a constant value of 2 mm for the compression extension - which represents 40 % from the cube edge - was maintained for all cell sizes. In those conditions, the values for compressive load were obtained, as presented in Table 3. For each cell size sample, within the compression tests, a significant difference was noticed between the obtained values, according to the way the sample was oriented, 90° versus 0°, with respect for the supports surface. The values obtained for compression load are smaller if the part is subject to compression in the perpendicular direction to the surface where the supports are, without exception. It can be observed that the compression load decreases by the increasing of the cell size.

### Table 2. Flexure measurements

<table>
<thead>
<tr>
<th>Cell size [mm]</th>
<th>0.5</th>
<th>0.6</th>
<th>0.7</th>
<th>0.8</th>
<th>0.9</th>
<th>1</th>
<th>1.1</th>
<th>1.2</th>
</tr>
</thead>
<tbody>
<tr>
<td>0°</td>
<td>1010</td>
<td>1063</td>
<td>761</td>
<td>664</td>
<td>441</td>
<td>484</td>
<td>222</td>
<td>234</td>
</tr>
<tr>
<td>90°</td>
<td>1190</td>
<td>128</td>
<td>133</td>
<td>139</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexure load [N]</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Extension [mm]</td>
<td>1.48</td>
<td>1.51</td>
<td>3.1</td>
<td>1.9</td>
<td>3.6</td>
<td>3.8</td>
<td>4.7</td>
<td>5.8</td>
</tr>
<tr>
<td></td>
<td>6.4</td>
<td>6.2</td>
<td>8.7</td>
<td>10</td>
<td>10</td>
<td>10</td>
<td>10</td>
<td>10</td>
</tr>
</tbody>
</table>

### Table 3 – Compression load at 40% compressive strain

<table>
<thead>
<tr>
<th>Cell size [mm]</th>
<th>0.5</th>
<th>0.6</th>
<th>0.7</th>
<th>0.8</th>
<th>0.9</th>
<th>1</th>
<th>1.1</th>
<th>1.2</th>
</tr>
</thead>
<tbody>
<tr>
<td>0°</td>
<td>8048</td>
<td>5707</td>
<td>4943</td>
<td>2358</td>
<td>2346</td>
<td>1027</td>
<td>859</td>
<td>560</td>
</tr>
<tr>
<td>90°</td>
<td>415</td>
<td>318</td>
<td>236</td>
<td>127</td>
<td>163</td>
<td>97</td>
<td>97</td>
<td>49</td>
</tr>
<tr>
<td>Compression load [N]</td>
<td></td>
<td></td>
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</table>
Conclusions

The mechanical behavior variation depends on the cell size of the lattice structures and also on the orientation of the samples. For a higher cell size, which means a higher porosity, the bending extension increased during the tests and the loading force decreased respectively. A higher porosity leads to the increase of flexibility. Specimens with a cell size higher than 1 mm manifested an excellent flexibility by flexure tests.

The critical cross-sectional area is depending on the sample’s orientation. The moments of inertia are also different. Future studies will focus onto the calculation and variation of the Young modulus of elasticity, depending on the cell size of lattice samples. The corresponding ultimate tensile strength needs also to be calculated. The final goal of this research is to be able to develop customized medical implants, with a lattice structure, with a similar mechanical behavior with the natural bone of the patient, with respects to his/her weight, age, shape of the implant, functional purpose of reconstruction.

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