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Published in: Materials and Design

Published: 05/07/2019

Document Version: Final Published version, also known as Publisher's PDF, Publisher's Final version or Version of Record

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Publication record in CityU Scholars: Go to record

Published version (DOI): 10.1016/j.matdes.2019.107773


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Octet-truss cellular materials for improved mechanical properties and specific energy absorption

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HIGHLIGHTS
- Optimization design with an objective of maximum specific modulus for octet-truss cellular material was performed.
- In situ static and loading-unloading compressive behaviors of 3D printed cellular materials were studied.
- Young’s moduli obtained from simulations for 3D printed cellular materials are consistent with those obtained from tests.
- 3D printed cellular materials are able to recover their original shapes after a compression up to 30% strain.

GRAPHICAL ABSTRACT

ABSTRACT

Optimization method has been widely acknowledged as an effective approach to design engineering structures, and yet few studies adopt this method to design cellular materials. Here, we firstly adopted a Kriging assisted Multi-objective Genetic Algorithm to guideline the design of octet-truss (OCT) cellular materials with the maximum specific modulus. Subsequently, additional struts were artificially introduced into the optimized OCT to further mechanically reinforce performances. All the cellular materials were precisely fabricated using a Stereolithography 3D printing technique. By reasonably optimizing the sizes of OCT, the optimized OCT with a 1.112 mm diameter and 8.282 mm cell length was achieved, which displays a superior modulus-to-mass ratio. The highest modulus and strength of as-designed cellular materials achieved 34.12 MPa and 2.64 MPa, reinforced by ~3.11 and 4.81 times, respectively. Additionally, the absorbed energy efficiencies of them improved from 74.75% to 90.80%, which are significantly higher than other cellular materials. By in situ tests and fracture analyses, the high recoverability is attributed to the comprehensive effect of net-shaped architecture and elastic-plastic deformation.

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ARTICLE INFO

Article history:
Received 13 January 2019
Received in revised form 29 March 2019
Accepted 30 March 2019
Available online 2 April 2019

Keywords:
Cellular materials
Optimization design
3D printing technique
In situ tests
Deformation mechanism
Recoverability

1. Introduction

The octet-truss (OCT) cellular material has been widely studied due to the advantages of high specific properties and superior energy absorption capacity, which has a potential application to alternate foams or honeycomb cellular materials [1–6]. The unit cell of OCT cellular
material contains 12 nodal or vertex connections and comes into being a face-centered cubic architecture. Its mechanical behaviors are determined by stretching, which are stronger than bending-dominated cellular materials with the similar density, such as foam [7–9]. Along with the development in computer-aided design (CAD) [10–12] and advanced 3D printing techniques [13–18], producing weight efficient, high modulus-to-mass ratio and superior energy absorption engineered cellular materials have been availability achieved.

Gümruk et al. [19] studied the static compressive behaviors of OCT cellular materials fabricated using selective laser melting (SLM), and found that the failure at connection points as well as material overlap should be taken into consideration to elaborate the mechanical responses. To further improve the mechanical properties, they experimentally examined the responses of OCT cellular materials with different relative densities and the relative density was revealed to be a primary parameter to characterize the mechanical properties of these materials [20]. Mohsenizadeh et al. [21] investigated the mechanical properties and energy-absorbing feature of OCT cellular materials manufactured using Stereolithography (SLA). Experiments found that the energy absorption efficiency of the materials was 11% higher than that of the aluminum foam and honeycomb. Dejean et al. [22] examined the mechanical performances of OCT with five specified relative densities and the scaling of the modulus and strength with the corresponding relative density was established. Ahmadi et al. [23] also found that the compressive properties of cellular materials increased with the rise of strut size, illustrating a power law relationship, which was also obtained by Wang et al. [24]. Apart from the influence of relative density, cell topology of unit cell plays an important role in the load-bearing capacity of cellular materials. Kaur et al. [25] investigated the mechanical properties of SLA-printed octet-truss and octahedral cellular materials, and experiments show that the stress-strain curves, deformation and failure mechanisms are remarkably different, even though the cell topologies are similar. Jin et al. [26] designed and fabricated four cellular materials to explicit the relationship between

![Fig. 1. Illustration and practical images of geometric architecture of four as-designed cellular materials. a–c, OCT. d–f, HR_OCT. g–i, VHR_OCT. j–l, BVHR_OCT.](image-url)
failure as well as energy absorption characteristics and cell topology. Results confirmed that the two aforementioned parameters have a significant impact on the mechanical properties and energy absorption. Mieszala et al. [27] fabricated microscale metal-coated polymer cellular composites with various cell topologies, and by in situ experiment, results show that the deformation mechanisms and buckling behaviors are entirely dependent with the corresponding cell topology. Aforementioned discussions have shown that the cell size, cell topology and material are significantly crucial to the mechanical responses of cellular materials. Apart from the selected materials commonly determined by 3D printers, it is therefore of significant concern that the challenges that how to find out a set of appropriate size parameters and design the cell topology in advance to further reinforce the mechanical properties. CAD assisted by optimization design could be an effective route to achieve the two objectives above [28,29]. The optimization methods mainly contain that size optimization, shape optimization, topology optimization (i.e. density-based method, level-set method and evolutionary method) [30] and intelligent optimization algorithms (i.e. Multi-objective Genetic Algorithm, Ant Colony Optimization and Particle Swarm Optimization) [31]. Recently, Meza et al. [32] fabricated a series of nanoscale OCT cellular materials with different geometric sizes, and reported that optimizing the wall thickness-to-radius of the tubes can lead to ductile-like deformation and recoverability. Tommasi et al. [33] performed a morphological optimization design to achieve a minimal weight design of tensegrity cellular materials. Apart from the selected materials commonly determined by 3D printers, it is therefore of significant concern that the challenges that how to find out a set of appropriate size parameters and design the cell topology in advance to further reinforce the mechanical properties. CAD assisted by optimization design could be an effective route to achieve the two objectives above [28,29]. The optimization methods mainly contain that size optimization, shape optimization, topology optimization (i.e. density-based method, level-set method and evolutionary method) [30] and intelligent optimization algorithms (i.e. Multi-objective Genetic Algorithm, Ant Colony Optimization and Particle Swarm Optimization) [31]. Recently, Meza et al. [32] fabricated a series of nanoscale OCT cellular materials with different geometric sizes, and reported that optimizing the wall thickness-to-radius of the tubes can lead to ductile-like deformation and recoverability. Tommasi et al. [33] performed a morphological optimization design to achieve a minimal weight design of tensegrity cellular materials. Aage et al. [34] carried out the topology optimization to design the cell topology of aircraft wing with the optimization objective of maximum modulus-to-weight ratio. Song et al. [35] also adopted topology optimization to design the cell topology of compression-resistance cellular materials, and the resultant topological layout resembles to that of cuttlebone that has a good anti-compression capacity.

Among different cellular materials, the octet-truss cellular architecture was chosen in this work as it not only possesses high specific mechanical properties and superior energy-absorbing characteristic, but also provides a nearly isotropic elastic response. Firstly, through taking the cell length and diameter of struts, viz. relative density, as optimization variables, the OCT cellular materials with the objective of maximum modulus-to-mass ratio was achieved based on the Multi-objective Genetic Algorithm assisted by the Kriging surrogate model. Based on the optimal topological architecture of OCT, three mechanically enhanced cellular materials were man-made designed to further reinforce the load-bearing and energy-absorbing capacities, all of which necessarily remain an isotropic elastic characteristic. Secondly, using SLA 3D printing technique, four categories of designed OCT cellular materials were fabricated. Thirdly, the mechanical responses of them under static and loading-unloading processes were in situ investigated. By in situ tests, finite element simulation and fracture morphology analysis, deformation behaviors, enhancement mechanism, mechanical properties and energy absorption characteristic were systematically elaborated. Finally, the modulus, strength and energy absorption efficiency of four as-designed cellular materials were compared with those of other cellular materials. These studies exhibit that it remains an opportunity to explore how 3D printing techniques in combination with carefully designed and tightly controlled geometries, can be adopted to create cellular materials with exceptional mechanical properties.

2. Analytical preliminaries

2.1. Geometric architecture

The OCT cellular materials with 3 × 3 × 3 unit cells were selected as the research objective. After optimizing the diameter of struts and cell length, the optimized OCT with the maximum of modulus-to-mass ratio was illustrated in Fig. 1a–c. Additionally, to further reinforce the mechanical properties making use of the three-dimensional (3D) spiral of cellular materials, three improved OCT cellular materials with isotropic and multi-level characteristics were artificially designed, see Fig. 1, respectively. Specifically, by insetting two horizontal struts (pink color) in the cell topology of optimized OCT, a horizontally reinforced OCT (HR_OCT) cellular material was obtained (Fig. 1d–f). Based on the cell topology of HR_OCT, additional one vertical strut was
incorporated to expectably improve the load-bearing ability, abbreviated by VHR_OCT (Fig. 1g–i). In the unit cell of VHR_OCT, additional twelve struts were added in the boundary to further enhance the mechanical properties, called by BVHR_OCT (Fig. 1j–l). In accordance with the cell topologies illustrated in Fig. 1b, e, h and k, four cellular materials have a nearly isotropic characteristic. Furthermore, four SLA-printed cellular materials possess relatively smooth and uniform architectures, see Fig. 1c, f, i and l.

2.2. Basic material properties of photopolymer

The basic material was a category of synthesized photosensitive resin (L101, Nova3D™), mainly composed of non-toxic acrylic polyester and curable using a 405 nm light [18]. Five dumbbell-shaped samples were fabricated by SLA method (see Section 4.1-Fabrication materials) and compression tests were performed in accordance with ASTM D638 standard. The related mechanical properties of photopolymer were given in Table S1, Supplementary information (SI).

2.3. Finite element analysis (FEA)

Based on ANSYS 17.0 software, FEA was established to evaluate the elastic properties and elastic behaviors of each as-designed cellular material. In each cellular material, taking into consideration computational efficiency and geometric symmetry at X = 0 and Y = 0 planes (Fig. 2a), a quarter of geometric model was obtained by cutting the full model using the two planes (Fig. 2b and c). The 1/4 geometric model was constructed in NX 10.0 software (All of resultant models were exhibited in Fig. S2, SI). The elastic components shown in Table S1 (SI) were adopted as the material properties for simulation. Ahead of meshing, mesh sensitivity was elaborated and a mesh size of 0.2 mm was thus used to mesh the geometric model using the path conforming method, see Fig. 2d (the mesh sensitivity analysis can be found in Fig. S4, SI). The finite element model was imposed four boundary conditions, in which the normal displacement of bottom face (Fig. 2e, “A”) was constrained, two symmetric constraints were imposed at X = 0 and Y = 0 planes (Fig. 2e, “C, D”) and the top face was subjected to normal displacement increasing linearly over 10 sub steps up to 1 mm, viz. 4% strain (Fig. 2e, “B”). Note: The rationality of boundary conditions was also discussed in Table S2, SI.

3. Optimization design method

Multi-objective Genetic Algorithm assisted by Kriging surrogate model was used to find out the optimized OCT with the maximum modulus-to-weight capacity, in which Central Composite Design was
adopted to create a series of sampling points with minimum possible trials to construct the Kriging surrogate model.

3.1. Design of experiments

In this work, a Central Composite Design [36] was adopted to create sampling points with two parameters of diameter of struts and cell length. The ranges of them were specified as 8.250–11.250 mm and 0.525–1.125 mm, respectively, depending on the 3D printer. The resultant experimental factors and design matrix were listed in Table 1.

3.2. Optimization algorithm

3.2.1. Multi-objective genetic algorithm (MGA)

The optimization problem can be described as follows:

\[
\begin{align*}
\text{max } & f_n(x) = E/m \\
\text{s.t. } & F = Ku \\
& x_1 \in [0.525, 1.125], x_2 \in [8.250, 11.250]
\end{align*}
\]

where \( f_n(x) \) is the objective functions corresponding to \( n \) individual. \( E \) and \( m \) are the modulus and mass. \( x \) is the design variable vector. \( F \) and \( u \) are the load and displacement vectors. \( K \) is the stiffness matrix. \( x_1 \) and \( x_2 \) are the design variables, viz. diameter of struts and cell length.

The algorithm starts with a current population of individuals (population number: 50), or called design points, in which the design points can be divided into two parts, viz. elite and offspring points. The elite ones (less than 70% of the population) are nondominated points that can be directly inherited from the previous generation. The offspring design points should experience crossover and mutation operations so that they can get access to next generation, where 0.95 and 0.05 were used as the probability values for crossover and mutation, respectively. Once the population is produced, the corresponding response values (objective functions) are obtained by simulation and the design points are also generated. Subsequently, a fitness evaluation is operated to find out the nondominated and dominated point designs. Next population is emerged by operating crossover and mutation operations, until to the maximum iteration number (1300). The flowchart can be seen in Fig. 3 (yellow dashed block).

3.2.2. Kriging assisted multi-objective genetic algorithm

The idea is to obtain the nondominated design points from Kriging surrogate model (KSM), rather than calculating responses by FEA [37].

### Table 2
Initial sets of design points.

<table>
<thead>
<tr>
<th>Runs</th>
<th>( x_1 )</th>
<th>( x_2 )</th>
<th>Mass (g)</th>
<th>Force reaction (N)</th>
<th>Modulus (MPa)</th>
<th>Specific modulus (MPa kg(^{-1}))</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.825</td>
<td>9.750</td>
<td>2.458</td>
<td>10.86</td>
<td>5.09</td>
<td>2070</td>
</tr>
<tr>
<td>2</td>
<td>0.525</td>
<td>9.750</td>
<td>1.046</td>
<td>4.11</td>
<td>1.92</td>
<td>1830</td>
</tr>
<tr>
<td>3</td>
<td>1.125</td>
<td>9.750</td>
<td>4.339</td>
<td>22.03</td>
<td>10.32</td>
<td>2380</td>
</tr>
<tr>
<td>4</td>
<td>0.825</td>
<td>8.250</td>
<td>2.027</td>
<td>13.39</td>
<td>6.27</td>
<td>3090</td>
</tr>
<tr>
<td>5</td>
<td>0.825</td>
<td>11.250</td>
<td>2.890</td>
<td>9.15</td>
<td>4.29</td>
<td>1480</td>
</tr>
<tr>
<td>6</td>
<td>0.525</td>
<td>8.250</td>
<td>0.872</td>
<td>4.98</td>
<td>2.32</td>
<td>2670</td>
</tr>
<tr>
<td>7</td>
<td>1.125</td>
<td>8.250</td>
<td>3.537</td>
<td>27.95</td>
<td>13.09</td>
<td>3700</td>
</tr>
<tr>
<td>8</td>
<td>0.525</td>
<td>11.250</td>
<td>1.221</td>
<td>3.51</td>
<td>1.64</td>
<td>1340</td>
</tr>
<tr>
<td>9</td>
<td>1.125</td>
<td>11.250</td>
<td>5.140</td>
<td>18.22</td>
<td>8.54</td>
<td>1660</td>
</tr>
</tbody>
</table>

Note: \( x_1 \)- diameter of struts; \( x_2 \)- cell length. Mass of samples were obtained from NX 10.0 software.

### Table 3
Comparison of the moduli of four as-design cellular materials based on the optimized geometric sizes obtained by the test and simulation.

<table>
<thead>
<tr>
<th>Cell topology</th>
<th>Sample-1</th>
<th>Sample-2</th>
<th>Sample-3</th>
<th>Average</th>
<th>Simulated modulus (MPa)</th>
<th>Error (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>OCT</td>
<td>9.83</td>
<td>13.23</td>
<td>9.83</td>
<td>10.96</td>
<td>12.64</td>
<td>1.68</td>
</tr>
<tr>
<td>HR_OCT</td>
<td>12.49</td>
<td>14.10</td>
<td>16.30</td>
<td>14.30</td>
<td>14.63</td>
<td>0.33</td>
</tr>
<tr>
<td>VHR_OCT</td>
<td>16.75</td>
<td>25.10</td>
<td>33.45</td>
<td>25.10</td>
<td>20.15</td>
<td>4.95</td>
</tr>
<tr>
<td>BVHR_OCT</td>
<td>37.32</td>
<td>38.11</td>
<td>26.94</td>
<td>34.12</td>
<td>26.92</td>
<td>7.20</td>
</tr>
</tbody>
</table>

where \( f_n(x) \) is the objective functions corresponding to \( n \) individual. \( E \) and \( m \) are the modulus and mass. \( x \) is the design variable vector. \( F \) and \( u \) are the load and displacement vectors. \( K \) is the stiffness matrix. \( x_1 \) and \( x_2 \) are the design variables, viz. diameter of struts and cell length.
Nevertheless, the response values obtained from KSM inevitably have predicted errors. If the domination design is not accepted, the points in dominated room are calculated by simulation; otherwise, it is acceptable to use KSM instead of simulation (green dashed block, Fig. 3). In this regard, a quantitative criterion [38], called minimum of minimum distance, was used to evaluate whether the predicted value from the KSM should be accepted. In the objective space, the MMD corresponding to the minimum distance between nondominated and dominated design points can be computed, viz.

$$\text{MMD} = \min \{ \| f(x_{\text{nd}}) - f(x_d) \|^2 \}$$

(2)

where \(x_{\text{nd}} \in \{ \text{nondominated set} \} \), \(x_d \in \{ \text{dominated set} \} \). The norm is defined in the F space, viz. \(\{ f_1, f_2, \ldots, f_m \} \). Then, the MMD, \(m = 1, \ldots, M \) can be obtained.

Moreover, in accordance with the theory in [38], the root mean squared error, RMSE, is used to criticize the nondominated points, viz.

$$s(x^*) = \sqrt{\frac{1}{2} \left( 1 - r^2 R^{-1} r + \frac{(1 - R^{-1}r)^2}{p^2 R^{-1}p} \right)}$$

(3)

where \(s(x^*) \) is the RMSE that means the predicted deviation from the KSM from the actual response. If any design point \(x\) satisfies the following holds:

$$s_m(x) \leq \text{MMD}_{f_m}, \quad m = 1, \ldots, M$$

(4)

then the predicted response values obtained from KSM is regarded as “good” values. Otherwise, the simulation is performed to obtain the actual responses.

After conducting the KSM or simulation, all the design points in the current generation can be obtained, in which the optimized \(f_m(x)\) can be simultaneously found. Operating the crossover and mutation operations, next population of designs can be generated unit to the maximum iteration number. Fig. 3 shows the flowchart of optimization design.

### 4. Experimental procedures

#### 4.1. Fabrication of as-designed cellular materials

A computer-aided design software tool, NX 10.0, was adopted to create the geometric models of OCT, HR_OCT, VHR_OCT and BVHR_OCT cellular materials (Section 2.1-Geometric architecture). The sizes of OCT were optimized using Kriging assisted Multi-objective Genetic Algorithm based on Ansys Workbench 17.0 software (Section 3-Optimization design method) and others have the same sizes with OCT. Each model contains a unit cell with dimensions of 8.282 mm × 8.282 mm × 8.282 mm, made of circular cross-sectional struts with a 1.112 mm diameter. 3D spatial repetition of these unit cells produced the bulk 3D cellular materials with dimensions of 24.846 mm × 24.846 mm × 24.846 mm for follow-up printed geometric model. The 3D geometric models were printed using a projection stereolithography (SLA, NOVA3D™ Printer).

Due to the exceptional geometry adaptability offered by 3D printing, complicated architectures can be easily fabricated by varying the geometric file. Referring to our previous work [35], the Standard Tesselation Language file of geometric model outputted from NX 10.0 software was processed by a slicer layer-by-layer, in which the slicing distance was set at 25 µm, with a curing time of 8000 ms for each layer. Subsequently, 3D printing was conducted with a layer thickness of 25 µm resolution in the vertical direction (z). The printed layers were directly bonded to each other and one-body cellular materials were created. The resultant cellular materials possess a relatively high-precision surface and porous characteristic, in which the horizontal, vertical and inclined struts are well-bonded at nodes, see Fig. 1. The geometric characteristics of all the printed cellular materials were given in Table S3, SI.

#### 4.2. In situ mechanical experiments

To elaborate the static and loading-unloading responses of four as-designed cellular materials, in situ characterization techniques assisted with a high-resolution telephoto macro lens were developed, see Fig. S3, SI. The experimental platform consists of an MTS RT/30 compression tester with a 30 kN load cell and a high-speed video camera (Canon™ EOS-1D X Mark II) equipped with a telephoto macro lens (Canon™ EF 100-400 mm f/4.5-5.6 L IS II USM Lens with Canon™ 77 mm 500D close-up lens attachment) linked to a remote controlling computer which also controls the MTS RT/30 tester. Here,
the compressive load was steeply imposed on the top surface of isotropic cellular materials, while the bottom surface was freely placed on the lower grip. The static compression displacement rate was 2 mm/min (related to a strain rate of $3 \times 10^{-3}$ s), which follows the test conditions stipulated in the ASTM polymer compression test standard. Five loading-unloading processes were conducted in a displacement control mode, viz. $0 \leftrightarrow 1.5$ mm, $0 \leftrightarrow 3.5$ mm, $0 \leftrightarrow 5.5$ mm, $0 \leftrightarrow 7.5$ mm, and $0 \leftrightarrow 9.5$ mm, respectively. In each process, the loading-unloading compression test was also controlled at a prescribed displacement rate of 2 mm/min. The shooting speed of camera was 12 frames to gather the deformed images of cellular materials. According to the record time, the load data and the corresponding image can be matched.

4.3. Scanning electron microscopy

The fracture surface morphologies of cellular materials were explored by scanning electron microscope (SEM, Quanta™ 450 FEG). Each sample was coated with a conductive layer of gold using a vacuum sputter coater (Hummer 6.2, Anatech USA) prior to imaging.

5. Results

5.1. Optimization study

5.1.1. Creation of initial design points

Probability matrix for samples with two parameters were created based on Central Composite Design, and the responses of modulus and modulus-to-mass ratio were then simulated based on ANSYS Workbench software. The initial sets of design points were generated and listed in Table 2.

5.1.2. Construction of Kriging surrogate model

The response surfaces were constructed using KSM, which represent relationships between design points and objective variables, viz. mass, modulus and specific modulus (modulus-to-mass ratio). Based on the initial sets of design points created by Central Composite Design in Table 2, the aforementioned response surfaces were shown in Fig. 4a–c. The response surface of modulus monotonically increases with the rise of diameter and cell length (Fig. 4a). Meanwhile, the corresponding mass synchronously increases (Fig. 4b), causing that it is difficult to assess the variation trend of modulus-to-mass ratio with the increase of geometric sizes (Fig. 4c). Therefore, it is necessary to perform the optimization analysis to theoretically find out the optimized design point with the maximum specific modulus.

5.1.3. Optimization analysis

The optimization analysis was performed based on the Kriging assisted MGA and Fig. 4d–f displays the changes of design variables and optimization objective with the increase of iteration. The optimization objective in term of specific modulus undergoes a significant zigzag variation process and gradually achieves a steady optimized value (Fig. 4f). It converged after 819 iterations and the optimized design corresponded to the one with a diameter of 1.112 mm and cell length of 8.282 mm. To verify the accuracy of response surface, the specific modulus of the optimization design was also simulated by FEA. The calculated value based on the response surface in the optimization point is 2986 MPa kg$^{-1}$, which is well accorded with the simulated value of 2970 MPa kg$^{-1}$. It indicates that the Kriging response surface can be used to evaluate the mechanical responses instead of FEA in this work.

Fig. 6. Stress-strain curves obtained from compression tests of a. OCT, b. HR_OCT, c. VHR_OCT and d. BVHR_OCT. Here, the slopes of dashed lines represent the predicted moduli.
5.2. Simulation study

Based on the optimized geometric sizes, the moduli of four as-designed cellular materials were also predicted by FEA. The predicted results were shown in Table 3 and also plotted in Fig. 6. The modulus errors between the measured and simulated values were 1.68, 0.33, 4.95 and 7.20 MPa, respectively, indicating that the established model and aforementioned boundary conditions basically predict the modulus of four cellular materials. Furthermore, the maximum error was 7.20 MPa for the BVHR_OCT, which could be attributed to the premature buckling behavior of boundary short struts. Moreover, the predicted moduli were in good agreement with the slopes in stress-strain curves of four cellular materials, demonstrated in Fig. 6. Therefore, it is reasonable for the finite element models to simulate the as-designed cellular materials and be selected as the optimization model to perform optimization design.

Apart from predicting the modulus, FEA was also used to simulate the stress distribution cases of cellular materials. It can be seen from Fig. 5a, b that the stress is mainly concentrated at the nodes of OCT and HR_OCT, making them the weak points at which fracture is initiated. However, the high-level stress domains vary from nodes to vertical struts in VHR_OCT and BVHR_OCT (Fig. 5c, d). It indicates that the vertical struts have a great contribution to bear the compressive force, which are more likely to experience an elastic-plastic or buckling deformation that have been observed by in situ tests (Sections 5.3.2 and 5.3.3-In situ tests). In addition, the maximum stress level steeply increases from 8.45 MPa to 14.33 MPa, which could be attributed to the vertical reinforcement effect.

Fig. 7. In situ deformation processes of four as-designed cellular materials.
Fig. 8. In situ loading-unloading processes of four as-designed cellular materials.
5.3. Experimental study

5.3.1. Stress vs. strain curves of four as-designed cellular materials

Fig. 6 illustrates the experimental stress-strain curves for four as-designed cellular materials, in which each category of cellular materials was repeatedly tested three times. At first, each curve exhibits a line relationship where the elastic modulus was calculated and summarized in Table 3. From Table 3, the modulus progressively increases from 10.96 MPa to 34.12 MPa, giving an indication of reinforcement effect of insetting struts in unit cell of OCT cellular material. Afterwards, there are noticeable fluctuations in the stress-strain curves of OCT and HR_OCT cellular materials, but more gradual increases in the curves of VHR_OCT and BVHR_OCT cellular materials. The reason could be ascribed to the effect of vertical reinforced struts in the VHR_OCT and BVHR_OCT, which would be further discussed in Sections 5.3.2 and 5.3.3 by combing with in situ images. Moreover, the predicted moduli are also plotted in the stress-strain curves for comparison, where the moduli are totally similar to the slopes of stress-strain curves.

5.3.2. In situ static behaviors of four as-designed cellular materials

To understand the static behaviors and reinforcement effects, the static compressive behaviors of four as-designed cellular materials were in situ tested (Fig. 7), and the corresponding deformation processes can be found in Movies S1–S4, SI. In the OCT, the stress-strain curve initially exhibits an elastic deformation up to the yield point, followed by a strengthening process. During the process, elastic-plastic deformation and local buckling of struts are in situ observed (Fig. 7a, i–iii). Subsequently, failure phenomena at the nodes result in a remarkable decrease in the stress-strain curve (Fig. 7a, iv). In the following loading process, due to the net-shaped architecture of OCT, elastic-plastic deformation, buckling of the struts and local failure at the nodes increasingly occur, causing that the struts overlap with each other. The following stress-strain behaviors also exhibit a fluctuant feature as the synergistic effect of damage and densification. Similar deformation behaviors were observed in the loading case of HR_OCT, but it demonstrates a remarkably higher yield and peak strength. The enhancement effect could be ascribed that the additional horizontal struts to some degree limit the transversal deformation capacity, leading to the enhancement of load-bearing capacity due to the Poisson’s effect (Fig. 7b, i–iv). Although the second peak is higher than the first one due to the densification effect, the modulus, yield and peak strength were calculated based on the data ahead of densification, viz. the i–iv process in Fig. 7b. In the VHR_OCT, since the additional vertical struts are parallel to the loading direction, the load-bearing capacity is significantly strengthened in comparison to that of OCT and HR_OCT (Fig. 7c). The additional vertical struts display a significant buckling deformation (Fig. 7c, iii), and the inclined struts also emerge a local buckling deformation as well as fractured at the conjunction nodes (Fig. 7c, iv). Besides, because the selected photopolymer possesses a comparatively good ductility (Fig. S1, SI), the vertical long struts and inclined struts have a relatively synchronous contribution to bear the compressive load (Fig. 7c, ii–iv). In the BVHR_OCT, the additional vertical short struts in the boundary not only further enhance the load-bearing capacity, but also exhibit a larger extent of deformability (Fig. 7d, i–iv).

In brief, the reinforcement effects of HR_OCT, VHR_OCT and BVHR_OCT attribute to the comprehensive effect of horizontal limitation (Poisson’s effect) and vertical enhancements (elastic or plastic deformation local buckling). Although the static behaviors confirmed the mechanical properties can be effectively improved, the recoverability and energy absorption capacity are another two significantly important performances for cellular materials, which would be elaborated by the following in situ loading-unloading tests discussed in Section 5.3.3.

5.3.3. In situ loading-unloading behaviors of four as-designed cellular materials

Fig. 8 illustrates the in situ loading-unloading behaviors of four as-designed cellular materials, respectively, and the corresponding deformation processes can be found in Movies S5–S8, SI. In Fig. 8a, b, because no additional vertical struts exist in OCT and HR_OCT, the inclined and horizontal struts completely bear the compressive force. The elastic-plastic capacity of photopolymer has a great contribution to recoverability (Fig. 8a, b, ii–iii), with samples recovering up to ~98% of their corresponding original height after compressing 6% strain (Table 4). The elastic-plastic deformation can be found during the 2nd loading cycle (Fig. 8a, b, iv). However, once the load is completely released again, the OCT or HR_OCT totally recover their corresponding origin height (Fig. 8a, b, v), resulting from the 3D net-shaped architecture, large ductile behavior of photopolymer and local buckling. After the 2nd loading-unloading cycle, the OCT or HR_OCT exhibit a high recoverable ability, with samples recovering up to 96–97% of their origin height (Table 4). During the 3rd loading process, the deformation behaviors of OCT or HR_OCT are significantly influenced by the fracture at the nodes, thereby leading to a large number of intact phenomena (Fig. 8a, b, vi). Nevertheless, after completely releasing the load, the samples still remain an excellent recoverability (Fig. 8a, vii and Fig. 8b, vii–ix).

In Fig. 8c and d, during the loading process, the VHR_OCT and BVHR_OCT also demonstrate a relatively high deformability, where the majority of struts experience an elastic-plastic deformation, local buckling and even fracture at the nodes (Fig. 8c–d, ii, iv, vi, viii). However, during the unloading process, the VHR_OCT and BVHR_OCT also display a favorable recoverability and a 90–92% residual strain after the 4th compression to 30% strain (Table 4). Moreover, the loading-unloading deformation and recoverability of VHR_OCT and BVHR_OCT can be affected by the additional vertical struts. As mentioned in the static analysis (Section 5.3.2), the vertical struts have an effective contribution to enhance the load-bearing capacities of VHR_OCT and BVHR_OCT (Fig. 7c and d), whereas a relatively high-strength recoverability is still observed in Fig. 8c and d. In the VHR_OCT, during the 3rd loading process, the vertical struts significantly undergo a buckling

<table>
<thead>
<tr>
<th>Cellular material</th>
<th>Maximum loading strain</th>
<th>6%</th>
<th>14%</th>
<th>22%</th>
<th>30%</th>
</tr>
</thead>
<tbody>
<tr>
<td>OCT</td>
<td>Residual strain</td>
<td>1.4%</td>
<td>3.4%</td>
<td></td>
<td></td>
</tr>
<tr>
<td>HR_OCT</td>
<td>1.6%</td>
<td>3.6%</td>
<td>5.7%</td>
<td>8.0%</td>
<td></td>
</tr>
<tr>
<td>VHR_OCT</td>
<td>1.9%</td>
<td>4.1%</td>
<td>5.0%</td>
<td>8.8%</td>
<td></td>
</tr>
<tr>
<td>BVHR_OCT</td>
<td>2.0%</td>
<td>4.7%</td>
<td>7.0%</td>
<td>8.0%</td>
<td></td>
</tr>
</tbody>
</table>

Fig. 9. An Ashby chart exhibiting the Young’s modulus versus density for cellular materials studied in this work and other previously cellular materials.
deformation (Fig. 8c, vi), and the corresponding deformation is recovered after the 3\textsuperscript{rd} unloading process (Fig. 8c, vii). However, due to the local failure at the nodes (Fig. 8c, vi), the elastic response of VHR\_OCT is affected, which can be confirmed by the stress-strain curve during the loading-unloading process in Fig. 8c. In the BVHR\_OCT, an elastic-plastic deformation of vertical short struts is observed (Fig. 8d, ii–iv), resulting in a positive effect to reinforce the mechanical properties. Afterwards, the BVHR\_OCT still experiences a plastic deformation in the loading-unloading stress vs. strain curve (Fig. 8d), which attributed to the plastic deformation of struts (especially for vertical short struts), local buckling and failure at the nodes. After completely unloading load, the BVHR\_OCT is totally recovered its original architecture (Fig. 8d, ix).

Therefore, the net-shaped architecture, elastic-plastic deformation of struts, localized buckling and failure at the nodes are the primary mechanisms for the recoverability of the as-designed cellular materials.

6. Discussion

6.1. Modulus and strength scaling of cellular materials

Fig. 9 compares the Young’s modulus vs. density of the as-designed cellular materials studied in this work with other previously reported cellular materials. The as-designed cellular materials based on the optimization sizes are relatively stiffer than Parylene [39], open-cell [40], alumina [41], silica [42] and hollow OCT [43] cellular materials. Furthermore, by artificially designing the moduli of HR\_OCT, VHR\_OCT and BVHR\_OCT are gradually improved with the increase of density. From this view, the Young’s modulus of as-designed cellular materials can be reasonably guiled by optimization method. Furthermore, Fig. 10 shows the compressive strength plotted against the density in Ashby chart. Compared to simple nickel pillar [39], Parylene [39] and polymer [44] cellular materials, the as-designed cellular materials demonstrate a higher strength. The strength of four as-designed polymer cellular materials can be comparable to that of metal-based cellular materials, such as Al honeycombs and hollow Ni-P OCT cellular materials [43].

6.2. Energy absorption capacity of cellular materials

According to the results shown in Sections 5.3.2 and 5.3.3-In situ tests, the energy can be dissipated through elastic-plastic buckling and fracture at the nodes, which therefore gives an indicator of the high-efficient energy absorption. To elaborate the energy absorption capacity of four as-designed cellular materials, the absorbed energy efficiency (AEE), $\eta$, can be quantitatively expressed as [21].

$$\eta = \frac{\int_{\varepsilon_0}^{\varepsilon} \sigma(\varepsilon)d\varepsilon}{\sigma_p\varepsilon_d}$$

(5)

where $\sigma_p$ is the peak stress, $\sigma$ is the stress, $\varepsilon$ is the stress strain, increased up to $\varepsilon_0$ ($\varepsilon_0 = 0.3$, in this work). Fig. 11 presents the Ashby chart of AEE as a function of the corresponding density for various cellular materials designed in this work and other reported cellular materials. All the OCT, HR\_OCT, VHR\_OCT and BVHR\_OCT cellular materials exhibit a high absorption energy capacity from 60% to 81% of AEE. The AEE of the cellular materials are also higher than that of other reported ones [27,45–47], revealing that the as-designed cellular materials can be used as alternative cellular materials with the energy-absorbing characteristic.

6.3. Surface microstructure effect induced by 3D printing technique

Fig. 12 illustrates SEM images of microcracks at different local fracture points. Although there are some corrugation-like microstructures on the surface of 3D printed surface (Fig. 12c), the contributions to crack propagation and fracture of struts are fairly small (Fig. 12a, d). Compared with other cellular materials printed by using FDM [15] and SLM [48], the SLA 3D printing technique and the printed parameters used in this work provide an available route to precisely fabricate the as-designed cellular materials. Additionally, the fractured surface shown in Fig. 12e, f exists many ductile strips (lamellar damage), indicating that the fracture mode attributes to the tension-induced fracture mode. Simultaneously, the fracture point “1” occurs a small plastic deformation, attributed to the tension-induced stress.

7. Conclusion

In this work, the Kriging assisted Multi-objective Genetic Algorithm was successfully applied to deliberately design the OCT cellular materials with the objective of maximum specific modulus ahead of fabrication and experiment. Taking mechanical reinforcement into consideration, three categories of reinforced cellular materials, viz. HR\_OCT, VHR\_OCT and BVHR\_OCT, with an isotropic property were man-made designed based on the optimized OCT architecture. By using the SLA 3D printing technique, the cellular materials designed in this work provide an available route to precisely fabricate the as-designed cellular materials.
technique, the four as-designed cellular materials were successfully fabricated. Optimization design, in situ tests, FEA and SEM analysis were implemented to elaborate the mechanical behaviors of cellular materials. The main conclusions drawn from the optimized, experimental and simulated studies are:

1. Optimization result shows that competitors with the larger diameter of struts and smaller cell length have a higher modulus, giving an indication of better energy absorption characteristic.

2. The static mechanical responses of four as-designed cellular materials were in situ tested, and the experimental results show the moduli and strengths of them are steeply enhanced. Additionally, buckling-induced deformation and layer-by-layer failure were observed for all the cellular materials, resulting in the progressive failure.

3. The in situ loading-unloading experiments were also performed to investigate the recoverability of as-designed cellular materials. Due to the comprehensive effect of elastic-plastic deformation of struts and tensile-induced local damage, all the cellular materials exhibit a high recoverability and the absorbed energy efficiencies are steeply increased from 60% to 81%, which are higher than other cellular materials.

4. The influence of 3D printer on the damage propagation is relatively weak. Although the corrugation-like microstructure still exists on the surface of struts, the initial microcracks and the propagation path are irrelevant to the microstructure. It indicates that the SLA 3D printing technique is an effective route to print the ordered cellular materials.

Supplementary data to this article can be found online at https://doi.org/10.1016/j.matdes.2019.107773.

CRediT authorship contribution statement


Acknowledgements

The support for this research has been provided by the National Natural Science Foundation of China (Grant No. 51778373), Natural Science Foundation of Guangdong Province China (Grant No. 2017A030313286), the Knowledge Innovation Project of Shenzhen (Grant No. JCYJ20170302143625006), New Teacher Program of Shenzhen University (Grant No. 2016066), Shenzhen Science and Technology Innovation Committee (JCYJ20170818103206501) are gratefully acknowledged.

References


Fig. 12. SEM images of fractured sample. a, A completely failed beam as a result of loading. b, Local enlargement of failure point beyond the node. c, Surface microstructure of beam. d, A completely failed domain at the node. e, Local enlargement of failure point at the node. f, Ductile fracture morphology.