International Journal of Advance Research in Science and Engineering
Volume No. 10, Issue No. 08, August 2021
www.ijarse.com
IJARSE
ISSN 2319 - 8354

Factors Affecting the Compressive strength of Additive Manufactured Scaffolds: A Review

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ABSTRACT

Recent advances in biomaterials and the different manufacturing process are addressing the need for patient specific bone scaffolds. Additive manufacturing processes such as Fused Deposition, Selective Laser Sintering (SLS), and Stereolithography have evolved as benchmark for tissue engineering owing to their ability to replicate the compound structural formation of dense tissues such as bones. It is now possible to create novel bone tissue engineering scaffolds with beneficial macro-microstructure, custom designed form, structure, wettability, mechanical strength, and cellular responses. The weak mechanical strength is a major disadvantage of these scaffolds. Since the mechanical properties play a vital role in designing a scaffold it is required to understand the effect of various factors on the mechanical properties of the scaffold. In this review, we have systematically stated the influence of several factors such as cell shape, porosity, pore size and strut diameter on the compressive strength of a 3D printed metallic scaffold. This review will certainly open new avenues for optimizing the compressive properties of a 3D printed scaffold by determining factors having maximum and minimum influence on the mechanical strength.

Keywords: *Additive Manufacturing, Compression Properties, Porosity, Scaffolds, Scaffold Architecture*

1. Introduction

The bone, also known as osseous tissue, is made of two parts: cortical and cancellous bone. The cortical bone is the thick outer layer while cancellous is the inner spongy part of the bone. The bone undergoes remodelling, maturation, differentiation, and resorption which depends on the interaction between osteocytes, osteoclast cells and osteoblasts[1]. These cells are responsible for the self-healing ability of the bone and helps to repair defects; but large-scale defects cannot be healed completely[2]. Thus, certain intervention methods are employed to accelerate the healing process and one such process is bone tissue engineering.

Bone Tissue engineering comprises of various methods which seek to repair the bone inadequacy by replacement of the damaged bone with a scaffold material. Scaffold is an artificial and temporary extra cellular matrix which promotes the formation of new bone. A perfect scaffold should have the surface area ratio, right porosity, biocompatibility, mechanical support, surface activity and shape that complement the medical application, and can enhance the growth of blood vessels and nerves and promote cell adhesion [3]. These bone substitutes namely scaffolds must have a customized external shape, a porosity and pore structure in order to provide a microenvironment which is necessary for cell activity and reproduction. In order to manufacture these scaffolds various fabricating procedures, exhibit this ability along with certain limitations as well. Consequently, exploring a fabrication method that is not best limited to only attaining the characteristic external shape but also correctly managing the pore shape for scaffolds is of wonderful importance for further orthopaedic application[4].

One such technique is Additive Manufacturing (AM), which includes a large range of versatile methods such as Selective laser melting (SLM), Electron beam melting (EBM) and Selective laser sintering (SLS) are mostly used. Scaffolds manufactured via AM processes are cost effective, durable and of good service quality despite the geometric complexities. But in spite of the above advantages the material and specimen properties, particularly the mechanical properties is one of the principal factors to decide the feasibility of AM processes in manufacturing a successful scaffold[5,6,7,8].

From a mechanical perspective it is essential to maintain the mechanical strength of the scaffolds structure after implantation for the reconstruction of tough, load bearing tissues inclusive of bone and cartilages. To be used correctly in tissue engineering, it is far vital that a biomaterial scaffold briefly withstands and conducts the loads and stresses that the new tissue will ultimately bear[9]. The following three factors of the scaffolds must match for a successful scaffold design: strength, bone loading stiffness and fatigue strength. Bone failure is a result of the scaffolds' stiffness exceeding the value of the natural bone which causes stress concentration in the bone vicinity. Whereas, if the stiffness of the scaffold is less than that of natural bone, stress concentration in the scaffold can cause implant failure as well as bone atrophy. This impact of stiffness mismatch which gives rise to irregular load distribution among bone and implant, is known as stress shielding[10].

Design parameters such as pore size, strut diameter, porosity and cell shape influence the stress shielding effect by directly impacting the compressive properties of the 3D printed specimen. A mechanically sound design must consider the extent of the effect of these parameters so as to easily customize the scaffold based on patient requirement, site of implantation and other need. This review paper summarises the above factors specifically for a metallic scaffold namely $Ti_6Al_4V_5$, manufactured using various additive manufacturing techniques.

2. Factors affecting compressive properties of scaffolds

2.1 Strut Diameter

Struts in a scaffold are cylindrical connectors which form each individual cell. Struts are connected to each other at nodes, which are usually spherical. The node diameter is often similar to the strut diameter to prevent stress concentration in either the strut or the node. Strut diameters for scaffolds typically range from 0.2mm - 0.8mm.

For different variations of configurations of lattices, it is observed that load capacity is affected significantly with change in configurations. However, the compressive strength is highly dependent on strut diameter and architecture[11]. In a study done between Ti_6Al_4V specimens with 4 different configurations (Regular dodecahedron(Rd), Diamond(Di), Regular hexahedron(Rh),Cuboctahedron(Co)), it is observed there is a significant increase in compressive strength with an increase in strut diameter (TABLE.1). However, the

IJARSE
ISSN 2319 - 8354

relation between the compressive strength and strut diameter is not strictly linear.The increase in compressive strength varies for different lattice configurations.

Table.1 Compressive strengths of 4-unit cell types for different strut diameters: (a) Strut diameter: 0.2mm (b) Strut diameter: 0.5mm (c) Strut diameter: 0.8mm. [11].

Unit Cell Type	Max. Compressive Strength (MPa)		
	StrutDiameter0.2mm	Strut Diameter 0.5mm	Strut Diameter 0.8mm
Regular dodecahedron (Rd)	0.3	6	20
Diamond (Di)	0.05	1	5
Regular hexahedron (Rh)	0.6	7	40
Cuboctahedron (Co)	0.9	25	75

Similarly, another study compared the effect of strut diameters on elastic modulus of bone scaffold. The elastic modulus is directly proportional to the strut thickness as a result of lower porosity(Fig.1). Additionally, strut diameter determines fusion which affects the mechanical properties[12].

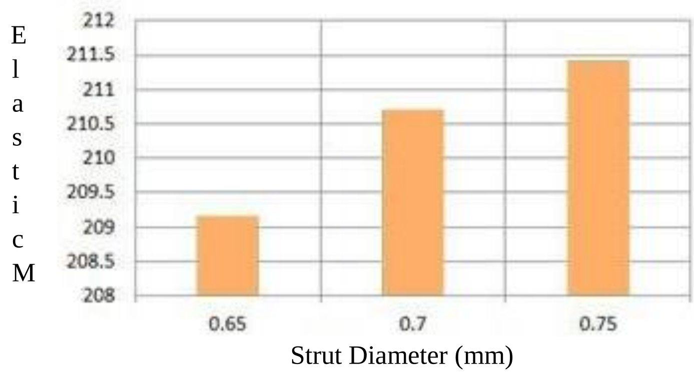

Fig.1: elastic modulus of a bone scaffold at different strut diameters[12].

Shrinkage of the struts as well as powder fusion also has a considerable effect, especially for strut diameters below 0.4mm. The shrinkage of the scaffolds needs to be measured in X, Y and Z directions. Linear shrinkage of ~13% is typically observed in all directions [13].

Another consideration to be taken is the effect of surface roughness and structure thickness distribution in complex lattice designs[14]. In certain designs such as a dodecahedron, the thickness distribution of the struts shows broader and asymmetric peaks with shoulders. These variations in thickness distributions can cause necking which can cause unexpected failure for lower strut diameters. This can be ignored for strut diameters above 0.4mm as even the thinnest sections can withstand the loads [15].

Strut orientation with respect to vertical loading also has an effect on the compressive strength and needs to be considered while designing a scaffold[16]. Diagonally oriented struts have the highest Ultimate compressive strength, followed by horizontal struts and finally, rectangular struts with shifted strut alignment[17,18].

For a printed octet truss structure, with an increase in strut radius if 0.1mm, the compressive strength approximately doubles. For low strut diameters, the struts deform mainly by stretching whereas for higher strut diameters fail due to bending. Additionally, samples with a higher height-to-width ratio have higher compressive strength for the same strut diameter, which needs to be considered while testing[19,20].While struts may prematurely twist at higher loading speeds during compression testing, loading speeds do not have a significant impact on mechanical properties[21].

There are several parameters regarding struts that affect compressive strength of a scaffold. However, strut diameter has the highest and most direct effect on the strength of the structure. For 3D printed scaffolds, strut diameter should ideally lie between 0.4-0.8mm. With an increase in strut diameter, there is an increase in compressive strength and a decrease in porosity. The optimum value of strut diameter observed in most tests is 0.4mm, regardless of other parameters.

2.2 Pore Size

Scaffolds must have pores which can serve as a medium for tissue regeneration. Pore size is the diameter of these (usually circular) pores in the cells of the scaffolds and is measured in micrometres. The shape of the pores varies and are dependent on the type of unit cell structure. However, for design considerations, a circular pore with a median diameter is assumed such that the circumference is in contact with most if not all sides of the actual pore.

The compressive properties of porous Ti Scaffolds can be similar to those of Bone and are variable with pore architecture[22,23]. Moreover, pore alignment can have an effect on the compressive strength of the scaffold [24.]. It is necessary to determine the pore size in the initial stages of scaffold design. Generally, the observed trend is an increase in compressive strength with the decrease in pore size, as it generally leads to an overall decrease in porosity. However, there are limitations on the minimum pore size, both because of the possibility of fusion with the struts and the minimum pore size required for tissue regeneration. A pore size higher than 0.1mm is considered ideal for rapid tissue regeneration in the scaffold, while a maximum pore size of 1.6mm is suggested for scaffold stability, as with the increase in pore size there is an increase in porosity, which reduces compressive strength. A bigger pore size also implies longer struts between nodes, causing overhangs which are weak to bending[25,26].Another study compared two levels of porosity (50% and 70%) and two pore size ranges

(250–500 μm and 500–1500 μm). The results concluded that for both levels of porosities, 500– 1500 μm was the most suitable pore size for tissue regeneration [27]. For scaffolds with cuboidal and gyroid lattice configurations, pore size of 0.3mm has the most ideal elastic modulus and yield strength, as seen in the below graph(Fig.2)[28].

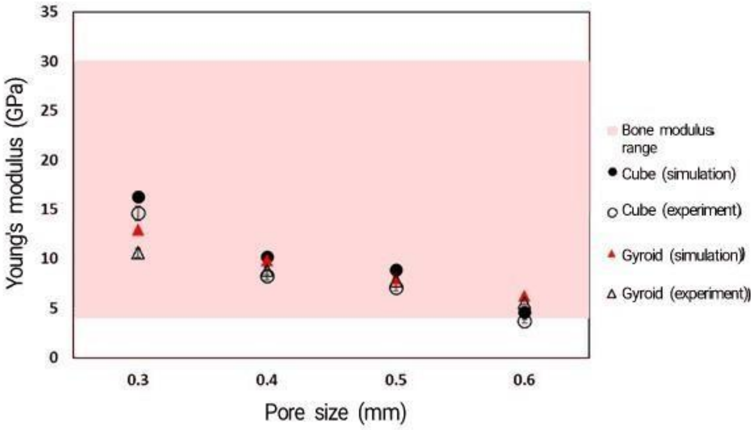

Fig.2:comparison between theoretical and experimental young's modulus for cube and gyroid samples[28].

In another study, for two samples with pore sizes of 0.7mm and 1.5mm, the strength and young's modulus decrease considerably because of an increase in porosity for quasi-static and cyclic compression tests[29.]. The mechanical properties of a group of Ti6 samples with pore size in the range of 0.1-0.5mm and a porosity of 55% reported a young's modulus in the range of 15-18GPa and compressive strength at yield point range 222-274MPa which is similar to that of the cortical bone[30].

As can be seen, pore sizes can vary anywhere from 0.1mm to 1.5mm. The elastic modulus of Ti6Al4V scaffold with a pore size of 1mm is very close to that of a human bone and stress shielding can be prevented to an extent [31]. However, considering the effect of other parameters such as cell type, strut diameter and porosity on the compressive strength for scaffolds, the optimal pore size probably lies somewhere between 0.2-1mm [32].

2.3 Porosity

Porosity is defined as a measure of the quantity of void spaces in a given component or material[33]. Porosity for this paper will be defined as a ratio of volume of pores to total volume. By definition, scaffolds imply that the component has a certain amount of porosity, by design. The purpose for a component having porosities is several. Porous components utilise lesser material and a compromise can be found so that the material has similar mechanical properties as compared to its solid counterpart. In several cases, such as the scaffold, porosity is an advantage. Ti64 scaffolds are used mainly for bone or tissue regeneration applications, where the scaffold is needed to allow and facilitate Osteointegration. Therefore, porosity is a very important parameter, and it also affects the compressive strength.

Increasing porosity is known to increase bone growth and osteoconduction. But on the other hand, we cannot keep increasing it as it also reduces mechanical properties[34,35,36,37]. Another factor to be considered is, Stress Shielding which refers to the reduction in bone density because stress is reduced from the bone by an implant. So increasing porosity reduces stress shielding but reducing porosity beyond a certain limit will not support the bone[38,39,40,41,42,43]. So, it can be said that for actual applications, porosity has a control range. Several studies have been reviewed to find a suitable trend between porosity and compressive strength.

A study compared the compressive strengths of scaffolds which had porosities ranging from 36-63%. These values of varying porosities were achieved by varying pore sizes and distribution.It did so by having two pore size distributions: Distribution A: 100-200μm and; Distribution B: 350-450μm. For Distribution A samples had porosities of 36%, 41%, 50% and 61%. For Distribution B samples has porosities of 39%, 45%, 55% and 63%.When compressive strength and porosities were plotted against each other, they showed a linear and inverse relationship to each other irrespective of Distribution A and B. That is to say, the values of compressive strengths kept decreasing with increase in porosities even for different pore sizes. Therefore, the specimen having the lowest porosity (35% of Distribution A) had the highest compressive strength and the specimen with the highest porosity (63% of Distribution B) had the lowest compressive strength. Thus, the study concluded that, compressive strength and porosity have a linear relation, completely independent of variations in pore size distribution[41].

A similar study showed that porosity and compressive strength remain linear nature, even if porosity is changed by varying cell size[21].However, when it comes to cell types, a study compared two-unit cells: Tetrahedron and Octet, each having porosities of 50%, 60%, 70% and 75%. On comparing the compressive strengths of both cell type against each of their porosities, they also showed a linear and inverse relation. However, for the specimens with octet cell types, the changes in compressive strength ended after 70% porosity. Thus, the study concluded that, for octet cell type, the decrease in compressive strength with increase in porosity ends for a porosity of 70% [42].

It is worth noting that, porosities in the range of 3% to 23% have been tested on rats and 23% proved to be the best among them for osteointegration and tissue regeneration [43]. Another study said that porosities of 75%–85% and a pore size of more than 100 μm are considered to be preferable only for rapid bone ingrowth[44]. In conclusion, porosity can be controlled in several ways. Irrespective of these ways, porosity mostly maintains a linear and inverse relation with compressive strength. However, through some of the studies mentioned above, smaller deviations are seen which gives rise to future scopes. There are chances that the linear relationship between porosity and compressive strength can change with different cell types.

2.4 Cell shape

Apart from porosity and pore size the cell shape also has a significant influence on the mechanical performance of porous scaffolds. A study evaluated the honeycomb structure of a Ti_6Al_4V scaffold and proved to match the mechanical strength of bone tissue. A Ti_6Al_4V confine with two rectangular square spaces was planned to accomplish required elastic properties of microstructure by Selective Laser Melting. Particular six-unit cells were planned using a software tool.

The unit cells were characterised according to shape (triangular (T), hexagonal (H) and rectangular (R)) and pore size in a 2-dimensional plane (1000 lm and 500 lm). The value of compressive stiffness ranged from 454 ± 39.9 MPa to 11.3 ± 2.8 GPa. The scaffolds compressive stiffness relied on the design and volume fraction of the scaffold. Designs with hexagonal pores had the maximum compressive stiffness[45]

Another study compared the mechanical properties of primitive, gyroid and BCC scaffold cell type. It was seen that the gyroid scaffold had the maximal strength, which was almost twice as that of BCC and primitive despite of all of them having the same porosity. The paper stated the large cross section area at the stress cross section as the reason for better stress transmission at loading direction leading to the largest compressive strength among the samples. These outcomes show that it is feasible to produce exceptionally permeable, high-strength and low stiffness frameworks for orthopaedic health applications by choosing the fitting unit cells of porous scaffolds. [46.] Cuadrado et al. studied the compressive strength of the following CAD designs: cubic, body-centred cubic and cross (Fig.3). The cubic structures which were manufactured at 45º displayed similar mechanical properties to the body centred cubic structures (BCC) though a somewhat greater value for BCC and fundamentally lesser value than the cubic structures was obtained. The main reason for this is because bending mechanism dominates both the above-mentioned structures[47].

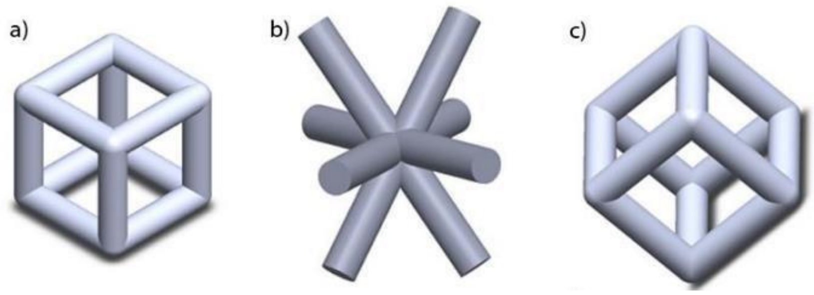

Fig 3: schematic representation of a) Cubic b) Body Centred c) Cross Structure [47].

The elastic modulus and compressive strength of four cell topologies namely Face centred cubic (FCC), Cubic closed packed (CCP), Body centred cubic (BCC) and Spherical hollow cubic (SHC) with different cell sizes were simulated using FE simulations.

The simulation results show that the modulus and strength of the FCC type was maximum and was minimum for CCP type despite of the cell size being constant. The main reason for FCC's maximum strength and modulus values is due to the presence of the horizontal arms which act as reinforcements. These reinforcements offer instantaneous resistance to deformation in case of FCC whereas they substantially decrease the pressure awareness inside SCR's. However, the ascending order on the strength and modulus of the cell topologies for constant porosity is SHC, BCC, FCC and CCP [48].

In another study 5 different cell types: Hexagon, Grid, Dodecahedron, W and X cell geometry were subjected to compression test. The strength strain curve was plotted to find the value of the Young's Modulus. The results were documented as seen (TABLE2). As seen the highest value of compressive strength was 81.04 MPa for Grid type of cell while the lowest was for W type[49].

Table 2: Compression test results [49]

Cell Type	Max. Compressive Stress (MPa)	Strain
Hexagon	51.34	0.08
Grid	81.04	0.045
Dodecahedron	21.96	0.75
W	20.11	0.55
X	30.06	0.82

Thus, different cell shapes provide varying compressive strength depending on their orientation, the manufacturing process and the cross-section area at the maximum stress area. The type of cell to be selected does not depend only on the compressive strength it also depends on the site of implantation and the design tolerance that can be achieved using a particular Additive Manufacturing Process.

3. CONCLUSION

The design and optimization of 3-dimensional scaffolds should be treated as a multidimensional problem which depends on the combination of various internal and external factors. The micro scale factors include manufacturing technologies and material properties while the meso scale factors include cell shape, pore architecture, topology, orientation and unit cell size. All these factors together affect the mechanical behaviour of scaffolds[49]. Apart from the above-mentioned factors, roughness and surface topography also lead to deviation in compressive strength[50]. The effect of stress shielding occurs mainly due to mismatch in the Young's Modulus between the scaffold and the bone. Thus, the need to completely understand all the parameters which may affect the structural properties of the scaffold arises. Since the inter dependency of various parameters is extremely complex, it is very important to understand the extent of the effect of these factors on compressive properties to optimize the scaffold design.

Declaration of Competing Interest

The authors declare no conflict of interest associated with this manuscript.

Funding

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

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