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1 Introduction

Emerging additive manufacturing and three-dimensional (3D) printing processes are enabling the design and fabrication of mechanically efficient complex structures [1-3]. Here we investigate the mechanics of beam-based lattice structures, using a case study of tissue engineering scaffolds that carry physiological loads while promoting targeted tissue regeneration [4-8]. Beam-based lattices consist of a network of beam structures that are organized to form a structure with potentially high mechanical efficiency and open pores to facilitate nutrient transport for growing tissues. Scaffolds are challenging to design due to the large number of conflicting mechanical and biological trade-offs that depend on a scaffold's geometry [9–11]. Computational design approaches are helpful in navigating these trade-offs and have recently demonstrated the merits in using 3D printed beam-based lattices that have an efficient scaling of stiffness with structural density [12-14]. Computational approaches are limited, however, because of the mismatch between computational predictions and experimental outcomes, partly due to limitations in 3D printed part fidelity [15,16]. In this paper, new mechanical testing experiments are

Mechanics of Three-Dimensional Printed Lattices for Biomedical Devices

Advances in three-dimensional (3D) printing are enabling the design and fabrication of tailored lattices with high mechanical efficiency. Here, we focus on conducting experiments to mechanically characterize lattice structures to measure properties that inform an integrated design, manufacturing, and experiment framework. Structures are configured as beam-based lattices intended for use in novel spinal cage devices for bone fusion, fabricated with polyjet printing. Polymer lattices with 50% and 70% porosity were fabricated with beam diameters of 0.4-1.0 mm, with measured effective elastic moduli from 28 MPa to 213 MPa. Effective elastic moduli decreased with higher lattice porosity, increased with larger beam diameters, and were highest for lattices compressed perpendicular to their original build direction. Cages were designed with 50% and 70% lattice porosities and included central voids for increased nutrient transport, reinforced shells for increased stiffness, or both. Cage stiffnesses ranged from 4.1 kN/mm to 9.6 kN/mm with yielding after 0.36–0.48 mm displacement, thus suggesting their suitability for typical spinal loads of 1.65 kN. The 50% porous cage with reinforced shell and central void was particularly favorable, with an 8.4 kN/mm stiffness enabling it to potentially function as a stand-alone spinal cage while retaining a large open void for enhanced nutrient transport. Findings support the future development of fully integrated design approaches for 3D printed structures, demonstrated here with a focus on experimentally investigating lattice structures for developing novel biomedical devices. [DOI: 10.1115/1.4042213]

> conducted to investigate the mechanical design of lattices with favorable topologies for tissue engineering applications, as informed by previous computational design studies. After investigating general lattice mechanics relevant to the tissue engineering design space, findings are applied toward configuring devices using lattices for bone tissue engineering.

> In our previous work, we have used computational design to investigate the performance of hierarchical lattices for tissue engineering and diverse lattice topologies, using finite element analysis incorporating beam elements to predict structural behavior [9,17]. These studies showed that the introduction of hierarchy added large voids throughout the lattices that could improve nutrient transport with the trade-off of reducing the overall stiffness of the structure. Cubic unit cells with orthogonal and diagonal beams were found to perform well in both compression and shear relevant to bone tissue engineering. Simulations of tissue growth demonstrated these topologies were additionally favorable for rapid biological growth [4]. This previous work has identified a favorable topology design for lattices; however, empirical validation in the form of mechanical testing has not yet been conducted. We therefore propose an approach with design, build, and test phases for mechanical characterization of beam-based lattices and biomedical devices [18], which may be used as a basis for developing fully integrated computational design and experimental approaches. The approach is suitable for characterizing biomedical devices such as spinal cages [19-22] that are implanted in

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Fig. 1 Design, manufacturing, and experiment approach, shown for 3D printed lattices in spinal cage devices

place of a removed intervertebral disk and facilitate bone tissue growth resulting in vertebral fusion (Fig. 1).

The design-build-test approach uses rapid prototyping to facilitate efficient iterations between design generation and testing [1,23]. The approach begins with outlining device specifications that inform subsequent decisions. In the design phase, structures are parameterized and generated with specified properties. In the build phase, structures are fabricated and part fidelity is assessed. During the test phase experiments are conducted to measure structural properties linked to performance. Outcomes inform future iterations or the tuning of a refined product. The approach is suitable for biomedical device development when multiple iterations are required to validate models and translate generalizable findings toward clinical applications [24-28]. Such iterations are essential in the context of 3D printing, where different materials, manufacturing processes, and lattice configuration strategies may influence final part performance uniquely [29], therefore requiring new experiments for each printing process and/or material considered.

The design-build-test approach may facilitate spinal cage development using an initial iteration to investigate how lattice design influences its mechanics. These general findings can focus on generic configurations of lattices as porous materials with properties suitable for bone tissue engineering, such as having at least 50% porous volume and a network of pores ranging from 0.2 mm to 1.0 mm in diameter [30,31]. General findings may then inform device-specific testing, such as developing spinal cages with macromechanical properties similar to the bone structures that they aim to bridge [32–35]. 3D printing enables the tuning of cages that use beam-based lattices as porous scaffolding with additional features, such as large pores included for enhanced nutrient transport or localized reinforcements for improved mechanics.

Our implementation in this paper uses two iterations of the Fig. 1 schematic, with a focus on experimental characterization. The first iteration investigates lattices generated with a plausible range of specified beam diameters and porosities for tissue engineering using polyjet printing [36]. Polyjet printing is favorable for scaffolds due to its capabilities for printing biocompatible materials at relevant resolutions, such as achieving beam diameters and pore sizes on the order of 0.5 mm [4]. The process additionally provides potential benefits in using polymers [37] to construct resorbable scaffolds with stiffnesses that facilitate bone growth, while avoiding stress shielding issues often caused by metal scaffolds. Only recently have biocompatible materials been introduced commercially for polyjet printing processes [38], and there is a need to measure their relevant properties when configured in lattices scaled for biological applications. Mechanical testing is used to measure properties and identify an appropriate range of favorable design parameters, such as a choosing a beam diameter that provides suitable stiffness for a spinal cage structure while retaining pore sizes suitable for tissue growth. Experimental characterization is employed to examine how the build process influences both fidelity and performance. For instance, there may be variances in beam diameters that influence a lattice's mechanics [16], but it is necessary to first measure these variances prior to including in a computational model, which has not been completed for polyjet printed lattices at these scales. This step accounts for the characteristics of the additive manufacturing process that influence the mechanical performance on a design that are difficult to predict.

In the second iteration of Fig. 1 schematic, a favorable point in the design space is identified through mechanical testing experiments, such as selecting an appropriate beam diameter for the porous lattice material that is used to investigate different strategies in configuring a novel spinal cage device. Configuration strategies that are considered include adding large voids for improved nutrient transport and adding localized supports for improved mechanics. These strategies build from previous studies that computationally demonstrated the inclusion of large voids reduced stiffness, yet lacked experimental validation [9]. Local structural reinforcement may mitigate this reduced stiffness and may be added toward the outside of the lattice structure where tissue is not expected to grow, since tissue only grows to fill concave pores within a lattice [4].

This iterative approach of first experimentally characterizing a base material followed by specific device configuration facilitates the development of biomedical products based on mechanical principles, where general characterization is necessary prior to using more expensive modeling and testing assessments as a product becomes more defined [28]. The approach is novel in that it uses a bio-inspired strategy of considering both small and large pores/voids throughout a structure, which is a feature of bone and many stochastic foam scaffold approaches, but is not commonly integrated with beam-based bone scaffolds [13].

In this paper, we focus on advancing the experimental aspects of a design-build-test approach, using previous computational design work as a foundation. Future work may then build on these results to demonstrate a fully integrative design-build-test framework that effectively utilizes both advanced experimental and computational design methods. An initial iteration of experiments are conducted for generally characterizing lattice mechanics and fidelity using compression testing and microscopy, followed by a second iteration for assessing spinal cages (i.e., application-specific devices). Cages are designed with varied strategies of local reinforcement for increased stiffness or material removal for improved nutrient transport and/or continuous bone bridging. Significant advances provided by this research are a general characterization of polyiet printing for microscale lattices, new mechanical experiments investigating a favorable topology for bone growth, and prototyping with mechanical assessment of novel spinal cage configuration strategies. Findings are expected to provide a better understanding of 3D printed lattice properties that are generalizable to a broad range of design and additive manufacturing applications, while also providing a foundation for integrating computational design, manufacturing, and experiments in biomedical device development.

2 Related Work

2.1 Design for Additive Manufacturing. Research and industrial applications for 3D printing have grown substantially over the past three decades, with a primary focus on improved manufacturing processes, while new design approaches have lagged [1]. New design approaches are necessary to take full advantage of 3D printing capabilities, such as enabling higher part complexity with customized configurations. Design feedback and iterations are particularly important, and are known to aid three-dimensional object development for traditional manufacturing approaches [39]. Feedback and iteration may facilitate design for additive manufacturing approaches since differences in achieved part performance depend on the printing processes used, with each process requiring specific guidelines for maximized performance [40].

Mechanical testing has been used to determine how manufacturing processes influence part performance, with a particular emphasis on stiffness and elastic modulus properties [36,41]. The layer-by-layer build processes of 3D printing introduce defects during fabrication due to the discretized layers being unable to precisely create part boundaries [15]. Further defects from printing processes emerge from the speed variation of machine tools and errors in positioning systems [42]. Errors in position may propagate from one layer to subsequent layers in the build process, therefore amplifying inconsistencies throughout the part. Objects may also move slightly during the build process, which is problematic since there is typically no feedback or process monitoring system to correct for these errors during printing. Material errors also occur and include shrinkage and stress-based distortions. Different printing processes will also have unique influences on error. When polyjet, digital light processing, fused filament fabrication, and stereolithography 3D printing processes were compared for printing dental models, polyjet and digital light process had the highest precision for cases when print time was similar across processes [43]. The differing levels of precision emerge from differences in the layer by layer fabrication for each process. For instance, stereolithography has errors emerge due to the movement of a mirror that directs a laser beam whereas errors for the polyjet process may occur due to how liquid resin is placed for curing in each build layer.

Fabrication defects make idealized predictions with computational models inaccurate [44]. The layers also introduce anisotropic mechanics for printed parts, which can further create difficulties in reconciling computational models with experimental outcomes. These mismatches are important to characterize empirically to determine appropriate modeling assumptions to use since they have trade-offs in computational time required and predictive power. For instance, less computationally demanding beam element simulations may be used to assess relative tradeoffs in exploring large numbers of alternate designs [17], but may overestimate stiffness in comparison with solid element models that better capture the effects of microscale deviations in part fidelity [16]. Due to these mismatches between expectations and actual part dimensions, it is essential to conduct experiments to characterize 3D printing processes for accuracy to fully understand how build processes influence performance.

2.2 Tissue Scaffolds. Scaffold mechanical and biological performance is informed by structural properties including porosity, stiffness, and pore size that are tunable with beam-based lattice design [12,13]. Porosity is the proportion of scaffold void area for tissue growth, stiffness is a scaffold's capacity to resist deformation under load, and pore size refers to the size of cavities within a scaffold where tissue grows. When beam-based lattices adhere to Maxwell's criterion for static determinacy, they are up to three times stiffer than stochastic foams of similar densities commonly used in scaffolds [2,14], offering significant potential for design optimization. The topological organization of beams can influence a lattice's mechanical response to varied loading conditions, and it is advantageous to orient beams in multiple directions to ensure a scaffold performs well in both compression and shear [17,45,46].

There are diverse strategies for organizing lattices, including periodic and pseudoperiodic organizations, with lattices often applied as lightweight core materials in sandwich structures [29]. Lattices have the potential to improve both compressive and shear strengths while suppressing buckling in comparison with common honeycomb strategies used for sandwiching. When used for bone tissue scaffolds, 3D printing is advantageous to fabricate complex structures, such as octahedral-based lattices. Previous studies have generated and investigated bone scaffolds using topology optimization, implicit surface modeling, image-based design, and CAD to create porous structures that carry mechanical load and have diverse strategies for nutrient transport and guided tissue growth [5]. Beam-based lattices with repeated unit cells are well-suited for providing both mechanical and biological functionality. When octahedral, pillar octahedral, cubic, and truncated octahedral unit cell topologies were compared empirically, the octahedral shape was found to provide greater stiffness and strength under compression as well as an increased rate of cell proliferation compared to other topologies [46]. Octahedral-based titanium scaffolds have been shown to have favorable performance compared to metal tantalum foam structures, with mechanical testing indicating that octahedral-lattices can be up to five times stronger than similar

foam structures when considering configurations suitable for bone tissue engineering [13].

When considering the classic Gibson–Ashby theory of lattice mechanics for bending and stretch-dominated structures [14], stretch-dominated structures will reach a higher stiffness for a given material density than bending-dominated foam structures. Bending-dominated structures have a strength that scales with porous material density $\rho^{1.5}$ while stretch-dominated structures scale with ρ [13]. A similar reasoning applies for comparisons in terms of stiffness, which motivates the selection of a stretch-dominated body-centric cubic topology used in this study as a base unit cell for lattices [47].

Mechanical testing of scaffolds with porosities relevant to tissue engineering has demonstrated that stiffer scaffolds tend to be less porous. Stiffness is related to a scaffold's effective elastic modulus that is dependent on lattice geometry, such as beam width to length ratio [48]. Effective elastic modulus refers to the ratio between stress and strain based on the nominal dimensions of a lattice, therefore taking into account its porous geometry. The effective elastic modulus is therefore lower than the elastic modulus of the base material used to construct the lattice. Scaffold effective elastic modulus is proportional to the elastic modulus of its base material, with titanium scaffolds having 100 times higher effective elastic modulus than scaffolds configured with similar topologies using sintered tricalcium phosphate [49]. Titanium and tricalcium phosphate represent two commonly used materials for bone scaffolds that have either high stiffness but are not biodegradable, as is often the case with metals such as titanium, or that have lower stiffness but with biodegradability, as is the case with ceramics such as tricalcium phosphate. The relative difference between a designed lattice's strength and its base material is influenced by the manufacturing process used to fabricate the lattice [29] and often requires experiments to fully characterize.

Accurate predictions for mechanical tests using finite element methods require solid element approaches, since Euler–Bernoulli and Timoshenko beam theory assumptions often do not hold [50]. Accurate imaging and mechanical testing measurements are required to validate these computational models so that they may be incorporated in computational design approaches to reduce the need for future time- and resource-intensive experiments. However, since each material and manufacturing process can influence 3D printed part performance uniquely [29], it is essential to first experimentally assess part performance prior to modeling to ensure relevant assumptions are incorporated.

2.3 Computational Design. Computational design is useful for biomedical device development since it facilitates efficient iteration between experiments, modeling, and design for complex and resource expensive applications [24–26]. Modeling may be used to evaluate both mechanical and biological scaffold functioning using finite element analysis and mechanobiological simulations, respectively [47,51]. Computational approaches also enable automation for generating diverse designs that may be evaluated and optimized to find favorable designs prior to fabrication and testing [9,52].

Computational design can aid in configuring spinal cage devices that include strategies for stand-alone porous cages, reinforced porous cages, cages with large central voids, and cages supported by pedicle screws in adjacent vertebrae [19–21]. Although compression is the primary loading experienced by cages, they are also subject to shear and torsion when considering all modes of spinal motion [22]. Mechanobiology simulations that model how loading influences tissue growth suggest that the cage geometry and stiffness may have substantial influences on resulting bone growth [53], and motivates the need for computational design methods to aid in tuning cages for optimized performance. Optimization approaches have produced 3D printed cage designs with stiffness of \sim 31.2 kN/mm for titanium [54] and \sim 7.5 kN/mm for poly(ϵ -caprolactone) mixed with hydroxyapatite [30]. The lower stiffness for the latter cage avoids stress shielding that is a

common problem for high stiffness titanium cages that result in weaker surrounding bone growth. These studies provide benchmarks for stiffness values useful for developing cages with alternate strategies.

Previous studies using computational design to investigate 3D printed cages have focused on the balance of mechanical stiffness of the cage with diffusivity for nutrient transport [30]. A limitation in these approaches is the focus on circular pores, rather than using beam-based lattices that create an open porous scaffold that enables the development of topologies with favorable scaling of stiffness with density. Additionally, previous studies used a laser sintering approach that constrained the minimum pore size, with minimum feature sizes from manufacturing being typically around $800 \,\mu\text{m}$. Polyjet printing allows smaller features such as pores around 500 μ m that encourage faster bone growth [4]. The polyjet printing process offers the use of polymers with potential advantages over titanium-based approaches, which have been used for similar beam-based lattices; however, the titanium scaffolds have a higher stiffness that may cause stress shielding and impede bone growth [13]. Although computational design approaches are essential to fully exploring and characterizing the complex design space of scaffold design, it is necessary to first experimentally characterize relevant print process and materials used so they may be accurately modeled, which is a focus of our present study.

3 Methods

3.1 Lattice Design. Lattices were designed with python code that automates ABAQUS software to construct unit cells from solid beams with octagonal cross section. This is adopted from our previous approach of comparing alternate topologies via finite element analysis using beam elements [17]. Cubic unit cells were constructed with beams along each edge and from each corner that meet in the center. The specific topology was chosen due to simulations suggesting it has favorable mechanics and bone tissue growth rates in comparison with alternate unit cell topologies constructed from beams [4]. Porosity *P* was determined by comparing the material volume of a lattice to its nominal volume. Lattices with P = 50% and P = 70% were generated by selecting a beam diameter ϕ and adjusting unit cell length *l* until a specified *P* is achieved (Fig. 2).

Figure 2 lattices have $\phi = 0.8$ mm with l = 2.6 mm for the P = 50% sample and l = 3.6 mm for the P = 70% sample, and act as control samples for a variety of mechanical tests and comparisons. Further samples were generated by specifying ϕ and rescaling l to achieve a desired P. Lattices for testing were generated with varied numbers of patterned unit cells, with lattices for beam diameter measurements having a $2 \times 3 \times 4$ configuration, meaning they have a cross-sectional area consisting of two unit cells by three unit cells and a height of four unit cells. In Fig. 2, the P = 50% sample has a $3 \times 3 \times 3$ unit cell patterning. Preliminary testing has demonstrated that repatterning unit cells of the same design to



Fig. 2 Designed lattice samples



Fig. 3 Support material removal for samples (a) as-printed, (b) after external support removal, (c) in chemical bath, and (d) cleaned (10mm length indicator)

alter overall lattice volume has a minimal influence on elastic modulus [18], with a $3 \times 3 \times 3$ and $11 \times 7 \times 4$ lattice having no significant difference in elastic modulus measurements. The preliminary testing suggested print batch variability had a larger influence of sample performance, therefore motivating all relative comparisons between samples groups to consist of samples printed in the same batch.

3.2 Build Process. Lattice samples were fabricated (Fig. 3) using a Stratasys Objet500 Connex3 polyjet printer with MED610 biocompatible polymer and SUP706 support material fully surrounding each sample (Fig. 3(a)) [36,38]. The printer builds structures by depositing and curing liquid resin in a layer-by-layer fashion, with a temporary support material used to facilitate construction of complex geometries. Post-processing is required to remove support material from each sample. External support material was removed using a razor blade (Fig. 3(b)), followed by sample immersion in a chemical bath of 2% NaOH:1% Na₂SiO₃ solution intermittently stirred while rocked at 60rpm (Fig. 3(c)). For $3 \times 3 \times 3 P = 50\%$ samples, approximately 4 h were required for internal support material removal while the $5 \times 5 \times 5 P =$ 70% samples required approximately 6h. Once samples were removed from the chemical bath, they were rinsed with water and left to dry overnight (Fig. 3(d)).

Once samples were cleaned, build accuracy was assessed using calipers to measure nominal sample dimensions. Porosity was determined by weighing samples and comparing their density to the 1.09mg/mm³ density of the base material. Sample faces were imaged with an Olympus IX51 microscope from the top of the structure relative to its original build direction (referred to as an "in-plane" face) and from the side of the structure relative to its original build direction (referred to as an "out-of-plane" face). Beam diameters were measured using IMAGEJ software [55]. Diameters were measured using P = 50% and P = 70% samples designed with $\phi = 0.4 \text{ mm}$, $\phi = 0.6 \text{ mm}$, $\phi = 0.8 \text{ mm}$, and $\phi = 1.0$ mm. Lattices were not reliably fabricated with open pores for beam diameters less than $\phi = 0.4$ mm. Each reported measurement represents the mean of at least ten samplings of beam diameters from multiple beams. Beam diameters were categorized based on beam orientation and whether they were aligned with in-plane or out-of-plane faces. In-plane beams were differentiated as 0 deg/ 90 deg orthogonal or 45 deg diagonal; 0 deg/90 deg beams were grouped together since they have no differentiable in-plane build artifacts. Out-of-plane beams were differentiated as 0 deg orthogonal with build layers aligned with their length, diagonal at 45 deg,



Fig. 4 Lattice in-plane and out-of-plane compression testing, with indicated build direction (0.5mm scale bars), sample height, and circled region indicating a broken beam from the build/clean process

or 90 deg orthogonal with build layers aligned with a beam's width.

3.3 Compression Testing. Mechanical properties were measured in quasi-static compression with an Instron E10000 Electro-Puls. Samples were tested with build layers parallel to loading plates for in-plane tests or rotated 90 deg for out-of-plane tests. (Fig. 4). Prior to compression testing, a small proportion of beams on lattices did not survive the building and cleaning process, as indicated in Fig. 4.

Load and displacement data were interpreted with Python code to calculate the effective elastic moduli for lattices and stiffnesses for cages in the linear region of the load–displacement curves and to find the yield point. Sample height *h* is always defined according to a sample's final orientation relative to loading plates during compression testing, regardless of their build direction as indicated in Fig. 4. Sample cross-sectional area always refers to the nominal area of each sample in contact with the loading plates. For comparing lattice samples to the fully dense material, solid 10 mm length cubes were fabricated and measured to have approximately 1.09mg/mm³ density, in-plane elastic modulus of 2115 \pm 32.3 MPa, and out-of-plane elastic modulus of 1860 \pm 6.8 MPa.

Three sets of mechanical tests were conducted to compare factors influencing lattice effective elastic moduli. The first tests compared designed samples used for beam diameter measurements to determine whether beam diameter and porosity influenced effective elastic modulus. The second tests compared Fig. 2 samples for in-plane and out-of-plane orientations to determine whether build direction influenced effective elastic modulus.

The final tests compared Fig. 2 samples subject to four different environmental conditions to determine how these influenced effective elastic modulus after two weeks and four weeks, compared to a control sample. Previous studies have demonstrated that the cleaning process does not influence mechanical properties if parts are removed from the chemical bath within three days [18]. However, part storage after printing and before/after cleaning or when exposed to solutions representative of conditions in the human body could influence mechanics [46], which informed the choice of environmental conditions. These conditions included dry storage at room temperature after cleaning (labeled "Cleaned"), storage in support material shown in Fig. 3(a) prior to cleaning a day before testing (labeled "Support"), immersion in water (labeled "H2O"), and immersion in phosphate-buffered saline solution (labeled "PBS"). Phosphate-buffered saline is a commonly used water-based salt solution representative of the physiological environment a scaffold experiences in vivo.

3.4 Cage Design. Spinal cages were designed using 50% and 70% porous lattices with localized material addition/removal to improve mechanical strength or provide a central void for improved nutrient transport and bone bridging. Lattices were rescaled for cages by patterning unit cells to achieve lateral dimensions of approximately 22 mm (about 500mm² cross section) with 12 mm height that are appropriate for placement between vertebrae [56]. Four spinal cages for each porosity were designed with the first configuration being a base "Lattice" that was modified by adding a central void ("+Void") for improved nutrient transport and bone bridging [9], adding a reinforced shell ("+Shell") for improved stiffness [17], or both ("+Both"), as shown in Fig. 5 for lattices with beam diameter $\phi = 0.8$ mm.

Voids for cages were formed by removing half or full unit cells through the lattice cross-sectional area to create a large planar pore of \sim 45mm². To mitigate the loss in stiffness introduced by the void, the hole was reinforced with material thickness equal to the beam diameter on unit cells bordering the hole. Shells were formed by adding solid reinforcements with half unit cell thicknesses to the edges of unit cells. Unit cell edges were modified on one set of opposing faces on the side of the cage and on top/bottom faces, which accounts for the asymmetry in thicknesses of side cage faces when viewing from the top perspective in Fig. 5(b). Figure 5(b) illustrates this added thickness for a structure that is made up of two layers of unit cells with added material above and below unit cells to form a structure with a height of about three unit cells. Beams connecting unit cells on these side faces and the top/bottom faces were reinforced by adding material thickness equal to one beam diameter.

3.5 Statistical Analysis. All results are reported as means from four printed lattice samples and cage devices, with significance between measurements determined as p < 0.1 when using a student's *t*-test. All plotted error bars represent the standard error of the mean.

4 **Results**

4.1 Build Accuracy. Samples with porosities P = 50% and P = 70% were fabricated with beam diameter ø ranging from 0.4 mm to 1.0 mm, in 0.2 mm intervals for the purposes of determining the accuracy of the printing processes (Table 1).

Mean nominal dimensions of samples were within 0.2 mm of their designed dimensions, with a bias of being too large. Measured P for the P = 50% samples were between 46% and 50%. Measured P for the P = 70% samples were between 65% and 68%. Images of in-plane and out-of-plane faces were used to assess build accuracy and investigate remaining support material presence, as shown in Fig. 6 for $P = 50\%/\phi = 1.0$ mm and $P = 70\%/\phi = 0.4$ mm samples (Fig. 6), that have unit cell lengths l = 3.3 mm and l = 1.8 mm, respectively. Images suggest most support material was removed, except a small layer clinging to beam surfaces and a few larger pieces on the corner of pores in the P = 70% sample.

Beam diameter measurements are presented in Table 2, with measurements grouped based on the orientation of beams on each imaged face. Measurements for beam diameter ϕ had a distribution of values based on variances in local beam diameters, therefore providing a standard deviation for each Table 2 measurement of mean diameter. Table 2 also contains the minimum and maximum local beam diameters measured for each lattice design when considering all beam orientations. For lattices designed with $\phi = 0.4$ mm, ϕ measurements ranged from a minimum of 0.28 mm to a maximum of 0.58 mm. Lattices designed with $\phi = 1.0$ mm had ϕ measurements ranging from 0.77 mm to 1.10 mm. The Range measurement in Table 2 represents the difference in



Fig. 5 (a) Designed spinal cage devices with varied configuration strategies and (b) schematic indicating labeled faces, build direction, and height h

Table 1 Lattice designed with varied beam diameters and unit cell lengths with respective measurements after fabrication; all lattices were generated in a $4 \times 3 \times 2$ unit cell configuration

	Lattice design							Mean measurements				
	Beam diameter (mm)	Unit cell length (mm)	Length (mm)	Width (mm)	Height (mm)	Porosity	Length (mm)	Width (mm)	Height (mm)	Weight (mg)	Porosity	
P = 50%	0.4	1.32	5.7	4.4	3.0	0.48	5.85	4.60	3.06	41	0.46	
	0.6	1.98	8.5	6.5	4.6	0.51	8.61	6.70	4.53	120	0.50	
	0.8	2.64	11.4	8.7	6.1	0.51	11.51	8.81	5.93	300	0.46	
	1.0	3.30	14.2	10.9	7.6	0.51	14.20	11.02	7.57	590	0.46	
<i>P</i> = 70%	0.4	1.80	7.6	5.8	4.0	0.68	7.74	5.96	3.98	59	0.65	
	0.6	2.70	11.4	8.7	6.0	0.68	11.41	8.89	6.07	180	0.68	
	0.8	3.60	15.2	11.6	8.0	0.68	15.25	11.70	7.94	430	0.67	
	1.0	4.50	19.0	14.5	10.0	0.68	19.06	14.53	9.90	880	0.65	



Fig. 6 Top and side faces of samples (0.5mm scale bar)

minimum and maximum local beam diameters for each lattice design and varied from 0.19 mm to 0.33 mm when considering all measured lattices.

When considering mean measurements of each beam grouping for both porosities, in-plane 0 deg/90 deg beams measured between 0.03 mm smaller to 0.12 mm larger than designed, while in-plane 45 deg beams measured between 0.06 mm smaller to 0.08 mm larger than designed. Out-of-plane 0 deg beams measured between 0.15 mm and 0.03 mm smaller than designed, out-of-plane 45 deg beams measured between 0.03 mm smaller to 0.09 mm larger than designed, while out-of-plane 90 deg beams measured between 0.06 mm smaller to 0.12 mm larger than designed. Some accuracy dependencies were observed with respect to designed ø, such as inplane 0 deg/90 deg beams for $P = 70\%/\phi = 0.4 \text{ mm}$ measured as $\phi = 0.52 \,\mathrm{mm}$ in comparison with more accurately fabricated P = $70\%/\phi = 1.0$ mm beams measured as 1.01 mm. The standard deviation of beam measurements ranged from 0.01 mm to 0.05 mm, and demonstrates there was a small range of deviances in local diameter measurements for each sample. Results from Table 1 are plotted in Fig. 7, to demonstrate deviances in mean measured beam diameters from their intended design for each case.

4.2 Lattice Mechanics. The measured effective elastic modulus for samples with varied porosity P and beam diameter ϕ is presented in Table 3.

Table 2 Designed lattices from Table 1 with mean beam diameter measurements for each beam orientation with standard deviation, with included minimum and maximum local diameter measurements

	Lattice design		Mean beam diameter measurements w/ standard deviation						Local beam diameter		
	Beam diameter (mm)	Unit cell length (mm)	Top 0 deg/ 90 deg (mm)	Top 45 deg (mm)	Side 0 deg (mm)	Side 45 deg (mm)	Side 90 deg (mm)	Minimum (mm)	Maximum (mm)	Range (mm)	
P = 50%	0.4	1.32	0.47 ± 0.026		0.30 ± 0.011	0.35 ± 0.024	0.39 ± 0.042	0.28	0.51	0.23	
	0.6	1.98	0.71 ± 0.045	0.66 ± 0.035	0.52 ± 0.009	0.56 ± 0.025	0.65 ± 0.029	0.51	0.75	0.24	
	0.8	2.64	0.84 ± 0.029	0.86 ± 0.019	0.73 ± 0.013	0.75 ± 0.028	0.72 ± 0.047	0.63	0.88	0.25	
	1.0	3.30	0.98 ± 0.025	1.05 ± 0.022	0.94 ± 0.019	0.93 ± 0.034	0.84 ± 0.047	0.77	1.10	0.33	
<i>P</i> = 70%	0.4	1.80	0.52 ± 0.022	0.55 ± 0.024	0.31 ± 0.010	0.38 ± 0.043	0.50 ± 0.024	0.30	0.58	0.28	
	0.6	2.70	0.66 ± 0.014	0.69 ± 0.025	0.54 ± 0.011	0.57 ± 0.026	0.64 ± 0.033	0.53	0.72	0.19	
	0.8	3.60	0.82 ± 0.032	0.86 ± 0.034	0.74 ± 0.015	0.74 ± 0.042	0.68 ± 0.040	0.62	0.94	0.32	
	1.0	4.50	1.01 ± 0.029	0.97 ± 0.025	0.85 ± 0.045	0.93 ± 0.024	0.89 ± 0.016	0.79	1.05	0.26	



Fig. 7 Measured beam diameters on (a) top and (b) side faces for porosity P = 50% samples and (c) top and (d) side faces for P = 70% samples; dotted lines represent ideal match between design and measurement

Table 3 Designed lattices from Table 1 with mean measurements of effective elastic modulus with standard error

	Lattice	e design	Mean measurements		
	Beam diameter (mm)	Unit cell length (mm)	Elastic modulus (MPa)	Standard error (MPa)	
P = 50%	0.4	1.32	60.6	10.5	
	0.6	1.98	66.4	4.4	
	0.8	2.64	126.4	2.7	
	1.0	3.30	180.4	5.2	
P = 70%	0.4	1.80	27.9	6.0	
	0.6	2.70	28.1	0.7	
	0.8	3.60	47.6	1.7	
	1.0	4.50	81.8	0.7	

Trends suggest that effective elastic modulus of the lattice *E* increases as ϕ increases, but decreases as *P* increases. Linear regressions were fit to results, with a best fit of $E = 0.21\phi - 38$ ($R^2 = 0.92$) for P = 50% samples and $E = 0.09\phi - 12$ ($R^2 = 0.85$) for P = 70% samples (Fig. 8).

Lattices of $\phi = 0.8$ mm were tested for in-plane and out-ofplane orientations with results shown in Fig. 9, using sample designs from Fig. 2. Out-of-plane samples had approximately twice as high an *E* as in-plane samples, which was 1.80 times higher for P = 50% and 2.15 times higher for P = 70% samples. Samples of $\phi = 0.8 \text{ mm}$ with E = 127.9 MPa and E = 37.7 MPa for P = 50% and P = 70% designs were used as a control for determining the influence of "Cleaned," "Support," "H2O," and "PBS" environmental conditions on E after two and four weeks of exposure (Fig. 10), using sample designs from Fig. 2.

Four-week measurements for all conditions significantly differed from the control, with Cleaned samples having an increase in E over time while all other conditions had a decreased E. Twoweek measurements for Cleaned, H2O, and PBS conditions



Fig. 8 Effective elastic modulus as porosity and beam diameter varies, with linear regression fits



Fig. 9 Effective elastic modulus for in-plane and out-of-plane compression orientations

differed significantly from the control, while samples in the Support condition did not differ significantly from the control. For the Cleaned samples, the two-week measurement was significantly different from the control but not from the four-week measurement, which suggests the largest increase in effective elastic modulus occurs soon after cleaning. Two-week measurements of Support samples showed no significant difference from the control, but did differ significantly from the four-week measurement. H2O and PBS samples after two weeks were significantly different from the control. All two-week measurements also differed significantly from four-week measurements for H2O and PBS conditions, with the exception of P = 50% samples in PBS.

4.3 Spinal Cage Assessment. Cages were fabricated for each porosity with 0.8 mm beam diameters, which was chosen as a

balance between having smaller beam diameters to reduce pore size for the scaffold while retaining large enough beams to reach a suitable stiffness. Cages were configured as base lattices with no additions or with added central voids and/or shells. All cages, regardless of configuration strategy, had an approximate height of 12 mm and cross-sectional area of 500 mm². Fabricated cages (Fig. 11) retained build accuracies similar to lattice samples reported in Tables 1 and 2. All cages were fabricated using the build direction orientation presented in Fig. 5, thereby representing an out-of-plane testing case.

Force-displacement curves were measured for each sample (Fig. 12(*a*)), with yielding occurring between 0.45 mm to 0.48 mm (approximately 4% strain) for P = 50% cages and 0.36 mm to 0.39 mm (approximately 3.3% strain) for P = 70% cages. The P = 70% Lattice and +Void cages were the least stiff cages and only required 1.4 kN for 0.4 mm displacement, while the highest stiffness cages were the P = 50% +Shell and +Both cages that required about 3.6 kN for 0.4 mm displacement. P = 50% Lattice and +Void cages, that all required about 2.2 kN for 0.35 mm displacement, prior to the P = 70% cages yielding. These relative differences among cages were also reflected by stiffness measurements (Fig. 12(*b*)).

Stiffness *k* ranged from 4.1 kN/mm to 9.6 kN/mm. Comparisons showed no significant difference between Lattice and +Void designs for P = 50% cages that had $k \sim 7.4$ kN/mm and for P = 70% cages with $k \sim 4.1$ kN/mm. There was also no significant difference between +Shell and +Both designs for P = 50% cages with $k \sim 9.3$ kN/mm and for P = 70% with $k \sim 7.7$ kN/mm. When comparing designs of different porosities, there was no significant difference between +Shell and +Both for P = 70% cages and Lattice and +Void for P = 50% cages. Results suggest that adding the large central void with local reinforcements to the lattice does not significantly decrease stiffness.

5 Discussion

A design-build-test approach was proposed for investigating the mechanics of beam-based lattices for tissue scaffold applications using 3D printing. In this paper, we focused on experimental design-build-test phases to first investigate lattice structures as porous materials in an initial iteration. Lattice elastic moduli measurements were then used to design a suitable lattice configuration for experimentally investigating design-build-test phases for spinal cages. Future work may pursue a fully integrated approach that more deeply explores both experimental and computational design, build, and test phases.

Lattices were designed with specified beam diameters and porosities by adopting previous computational design methods [17] that enabled the fabrication of lattice samples with relevant properties for tissue engineering. Build accuracy measurements showed that lattices were within 5% of their intended porosity and



Fig. 10 Effective elastic modulus after two and four weeks in varied conditions for (a) porosity P = 50% and (b) P = 70% samples compared to control (solid line)



Fig. 11 Fabricated cage devices (10mm length indicator)



Fig. 12 Mechanically tested cages for (a) mean force-displacement and (b) stiffness

0.2 mm of their nominal dimensions, and suggest the polyjet process provides a suitable degree of accuracy and reliability at these scales. The discrepancy in porosity between design and printed parts is potentially due to some beams being broken during the cleaning process (Fig. 4) and inconsistencies in printed beam diameters.

Beam diameters were up to 0.15 mm different from their intended design on average, with greater variances locally. For instance, beams designed with 0.4 mm diameters were measured with local diameters as small as 0.28 mm. These deviations from the intended design could have large influences on part performance due to weaknesses in beams where the local diameter is thinner than the mean diameter. In cases when beams were measured as larger than intended on average, such as for 1.0 mm designs, these local fabrication deviations could act as points of weakness with the local diameter potentially being smaller than the designed diameter. Local alterations of a given size would have a larger relative effect on smaller beam designs. As an example, if local fabrication differences resulted in a 0.1 mm smaller diameter than expected, it would result in a 25% smaller diameter than expected for a 0.4 mm beam but only a 10% smaller diameter than expected for 1.0 mm designed beams. In our measurements, the difference between local minimum and maximum measured diameters ranged from 0.19 mm to 0.33 mm when considering all lattice designs. Future statistical analyses, such as factor analysis or clustering, may be employed to further investigate potential defect thresholds and accuracy/error ranges.

Lattices were demonstrated to increase in stiffness as beam diameters increased while porosity was held constant. A linear regression was used for describing the elastic modulus change with beam diameter and represents the simplest case to accurately describe the empirical data. These findings correspond to previous finite element models that demonstrate a linear scaling of elastic modulus as the mean diameter of a lattice is varied when manufacturing defects are considered for cubic unit cells [57]. Higher order fits could be used to describe the data with a better accuracy. For instance, a second-order polynomial fitting of Fig. 8 data provides $E = 0.0001\phi^2 - 0.073\phi$ ($R^2 = 0.95$) for P = 50% lattices and $E = 0.00005\phi^2 - 0.025\phi$ ($R^2 = 0.91$) for P = 70% lattices that gives higher R^2 values in comparison with the linear fit. However, without collecting a broader range of data it is unclear whether the linear or another model is more appropriate for extrapolations for designs with diameters beyond the range of collected data.

The elastic modulus for the 70% porous lattice was assessed using ABAQUS with beam elements [17], which calculated its elastic modulus as approximately 120 MPa, based on its relative elastic modulus being about 0.065 for the fully dense material's measured value of 1860 MPa. The elastic modulus for lattices was found in these simulations by applying a small displacement to one face of the lattice, determining the reaction forces, and then the solving for the elastic modulus based on a lattice's height and nominal cross-sectional area. The relative elastic modulus was determined by taking the ratio of the lattice's elastic modulus to that of a solid structure of the same dimensions and material. When the modeled lattice was mechanically tested, it was demonstrated to have an elastic modulus of 28–82 MPa, depending on its beam diameter (Fig. 8), thus suggesting an overestimation with the beam element model.

The ABAQUS model also incorrectly suggests the elastic modulus remains constant if porosity is fixed via rescaling unit cell sizes as beam diameter is altered as a design parameter. This mismatch between experiment and results suggests the need for more advanced finite element analysis that incorporates solid elements [50]. There are multiple sources of error for the ABAQUS model that limits its agreement with the measured data. For example, the cross-sectional area of fabricated beams is not uniform which requires adjustment for finite element analyses to accurately predict mechanics in lattices [16]. The inconsistent cross-sectional area of beams is supported by the measurements in Table 2 showing numerous cases with mean beam diameter measured lower than designed in addition to local beam diameter fluctuations informed by the standard deviation of measurements. In contrast to fused deposition printing processes that accounted for beam diameters varying up to 0.9 mm for a given design [16], the polyjet printing process has smaller absolute local variances, as evidenced from the 0.78 mm to 1.10 mm range of local diameters for 1.0 mm beam diameter designs collected for Table 2 data. This smaller range suggests the modeling of distributed local beam diameters may have a greater relative influence for fused deposition modeling parts in comparison with more smoothly printed polyjet printed parts. However, since the polyjet printing process can reach lower resolutions, the stochastic effects may have a larger relative influence on smaller beam diameter designs. For instance, 0.4 mm designed beam diameters range locally from 0.28 mm to 0.58 mm and have a high relative variance in local diameter even though the absolute difference is not as large as fused deposition modeling. The ABAQUS model also does not incorporate the increased stiffness of joints that has been shown to improve fits of beam-based finite element models to experiments for the same unit cell topology used in this study [58]. Finally, the model does not consider deformation mechanisms of the lattice, such as beam bending that may occur during mechanical testing.

Models must also consider anisotropy introduced by the build process, since lattices loaded out-of-plane to their original build direction had approximately twice the effective elastic modulus to those compressed in-plane. Interestingly, the polymer material's compressive modulus was measured as 2115 MPa for in-plane fully dense solid cubes and 1860 MPa for out-of-plane fully dense solid cubes, thus suggesting the anisotropic properties of the lattice are due to its beam-based nature and not just the base printed material and printing process. To elaborate, our results show that an inplane solid cube that is rotated for out-of-plane compression has a 12% drop in stiffness, whereas an in-plane 70% porous lattice that is rotated for out-of-plane compression has a 115% increase in stiffness. These opposing influences of anisotropy for solids and lattices suggest that optimization methods lattices at these size and length scales should model beam mechanics explicitly [59], rather than assuming anisotropic effects of lattices are strictly proportional to those of the base material. With the exception of the anisotropy test samples tested out-of-plane, all lattice samples in this study were tested in the in-plane direction. Due to the increased stiffness of out-of-plane lattices, the eight cage designs were tested in the outof-plane direction as indicated in Fig. 5.

Lattices were subject to varied storage and environmental conditions to determine how they perform in scenarios relevant to biomedical devices. On average after four weeks, dry storage resulted in a 30% effective elastic modulus increase, storage in support resulted in a 47% decrease, soaking in water resulted in a 60% decrease, and soaking in PBS resulted in a 38% decrease. Findings suggest that cleaning parts immediately is preferred to storing parts in support material. The increase in effective elastic modulus over time may occur due to polymer crystallization for dry storage conditions, since it is common for polymerization (cross-linking) to continue for photosensitive resins over time. The influence of water and PBS suggests that parts should be removed from the chemical bath as soon as support material is removed, since prolonged exposure to water may negatively influence part performance. Our findings agree with studies of 3D printed polymer lattices soaked in PBS for 1 h prior to testing that demonstrated a significant drop in stiffness, halving material elasticity in some cases, that was attributed to relaxation of the

polymer [46]. The drop in stiffness over time when lattices were stored in support material contrasts with findings for other polyjet printed materials that did not lose stiffness when stored in support material for up to 20 days [36]. Differences suggest that both the material choice and printing process should be taken into account when storing parts, different versions of support material may also influence performance after storage. These considerations suggest further testing is required to determine how environmental factors influence a lattice's functioning over longer durations, such as swelling that may occur due to water absorption or part deterioration. Preliminary measurements, however, suggest external lattice dimensions and lattice porosity do not change significantly after soaking and drying prior to mechanical testing. Further work is required to fully investigate how environmental influences affect lattices and individual beams in the structure. The need for further characterization highlights the importance in utilizing safety factors to account for mechanical properties changing over time.

Findings from mechanical experiments were used to inform design decisions in developing spinal cages, such as using a beam diameter of 0.8 mm and out-of-plane orientation. The beam diameter size was chosen since it provides the highest effective elastic modulus when controlling for porosity that retains reasonable pore sizes for tissue growth since larger beam diameters increase pore size, while smaller beam diameters reduce lattice stiffness. Internal planar pore sizes were approximately 1.1 mm for the cages with 70% porous lattices and 0.7 mm for cages with 50% porous lattices. The lattice porosity represents a local porosity each cage retains for tissue ingrowth, but overall cage porosities differ based on additional voids or shells that take up volume but do not provide porous space for tissue to grow. When lattices configured as spinal cages were compared to lattices with the same beam diameter in Fig. 8, there was a maximum difference of 4 MPa, suggesting that general effective elastic modulus measurements from lattice samples are applicable to rescaled lattices sized for similar applications.

Rescaled lattices have differing numbers of unit cells that may influence overall mechanical behavior of the structure since unit cells near the outside of the lattice have different boundary conditions than those toward the center of the lattice. In foam lattices, differing scaling of the overall structure to unit cell size has demonstrated an increase of about 12% in shear modulus that occurs as overall volume is reduced when evaluating with a numerical model [60]. Finite element analyses that have investigated unit cells with similar topology as this study found that the elastic modulus increases as the lattice is designed with more unit cells. Their findings showed that a single unit cell has a 15% lower elastic modulus than a $3 \times 3 \times 3$ lattice [61]. However, as more unit cells are added the differences in elastic modulus diminish. For example, a $3 \times 3 \times 3$ lattice has less than a 1% difference in elastic modulus when compared a $5 \times 5 \times 5$ lattice. The consistency in elastic modulus occurs as the lattice is rescaled to a larger size since there is a diminishing of the effects caused by unit cells at the boundaries of the lattice structure. An increase in volume therefore leads to a homogeneity in structural deformation and stress distribution as if the structure were part of a homogeneous porous solid. These results agree with our previous mechanical testing findings that show rescaling lattice volume by patterning different numbers of unit cells does not significantly influence elastic modulus measurements for the relevant cases studies [18]. For example, we rescaled a 50% porous $3 \times 3 \times 3$ lattice to a $7 \times 11 \times 4$ configuration that represents a volume suitable for cage applications and found a less than 1% difference in their mean elastic modulus measurements with standard error of 131.6 ± 3.1 MPa and 130.1 ± 1.6 MPa, respectively.

Cages were designed with strategies including a large reinforced central void for improved nutrient transport and/or additional material to form a reinforced shell. Cage stiffness significantly improved with the shell, but did not significantly decrease with the addition of the void, thus suggesting the void as an improvement for designs when increased porosity is desired for increased nutrient transport. The downside of adding the void is a decrease in surface–volume ratio that results in less locations for cells to attach. The shell is beneficial in cases when a lattice needs a higher stiffness or safety factor. The 50% porous cage with both a void and shell had a stiffness of 8.9 kN/mm, that compares favorably to 3D printed cages using poly(ε -caprolactone) mixed with hydroxyapatite that have 7.5 kN/mm stiffness and avoids stress shielding issues of titanium cages that have 31.2 kN/mm stiffness [30,54]. The maximum spinal load typically experienced on a daily basis has been reported as 1.65 kN [62]. This load would displace the cage by 0.19 mm, which suggests that the cage would operate well below its yield point of approximately 0.45 mm. The safety factor for the design may be further improved through additional pedicle screws and supporting hardware.

The out-of-plane cages had higher stiffness than their in-plane counter-parts collected in an earlier preliminary study [18]. For instance, the cage with both shell and void had a stiffness of 8.9 kN/mm when tested out-of-plane but 6.5 kN/mm for in-plane samples. However, precise comparison of samples is limited since data were collected with different batches of parts that were not controlled for testing time after printing. The strategy of including the large void in the cage aimed to remove the portion of a scaffold that typically has the slowest growth due to low nutrient transport. Although removing this aspect of the structure should result in a decreased stiffness [9], local reinforcements were provided along the boundaries that mitigated the loss in stiffness (Fig. 12). These reinforcements were strategically placed on the outside boundaries of the scaffold such that overall tissue growth should not be impeded, since tissues grow to fill concave cavities within the scaffold [4].

The strategy of adding a large void is based on hierarchical generation strategies that pattern unit cells across multiple length scales. Findings have suggested that an optimal degree of hierarchy for resilience consists of replicating the base unit cell across two levels of recursion [63], which generates a structure consisting of unit cells and large voids, similar to the cage with the reinforced void. Future work could investigate different ratios between unit cells and voids to find an optimal tuning of the structure, which may be dependent on unit cell type and could benefit from mimicking bone's hierarchical structure to provide both mechanical and biological benefit.

A limitation in this study is that only a small portion of the design space was experimentally investigated, in comparison with the very large design space of lattices that could be fabricated using polyjet printing [36]. Although experimental characterization is necessary prior to creating computational models, future investigations to characterize a larger portion of the design space could be performed once a computational model is validated for the collected data. Although this study focused on a specific application, there are general conclusions that may apply broadly to biomedical devices using designed lattices. Some of the most widely applicable principles include the notion that larger beam diameters increase lattice stiffness, anisotropy increases lattice stiffness, and patterning different numbers of unit cells does not significantly alter lattice elastic modulus. These conclusions, however, must be taken with caution as they may not extrapolate to test cases beyond the design space investigated in this study.

Future experiments may further develop an integrative computational design, build, test approach that investigates how local reinforcements and voids influence cage performance. Limitations of the current study could also be addressed, such as increasing the number of design variations considered or exploring the use of alternative additive manufacturing processes, materials, and length scales [64,65]. The methodology could be streamlined with design of experiments approaches for tuning cages based on experimental results from small samplings of alternate designs. Findings can inform new modeling approaches in finite element analysis that account for how manufacturing defects and anisotropy influence modeled performance [29,59]. Further computational design and scientific investigations could lead to discoveries in advantageous lattice configurations for diverse mechanical applications and increased accuracy of computational models.

6 Conclusion

Beam-based lattice structures were characterized for use in 3D printed biomedical devices and applied in the mechanical design of novel spinal cage devices using polyjet printing. Lattice samples were designed with 50% and 70% porosity and beam diameters of 0.4-1.0 mm. Fabricated lattices were printed within 5% of their designed porosity while beam diameters ranged from 0.15 mm smaller to 0.12 mm larger than designed. Effective elastic moduli ranged from 28 MPa to 213 MPa. Cages were designed and built with lattices informed from earlier measurements, with localized reinforcements around a central void, on lattice faces, or both. Tested cages had stiffnesses from 4.1 kN/mm to 9.6 kN/mm, with yielding after 0.36 mm to 0.48 mm displacement (3-4% nominal strain). The 50% porous cage with reinforced shell and central void was a particularly favorable design that had a stiffness of 8.4 kN/mm so it could potentially function as a stand-alone spinal cage for supporting bone fusion with favorable nutrient transport. Findings support the future development of integrated design, manufacturing, and experiment approaches for characterizing complex 3D printed structures, and provide a foundation for developing mechanically efficient lattice structures for diverse applications.

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