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Trabecular scaffolds created using micro CT guided fused deposition modeling

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Abstract

Free form fabrication and high resolution imaging techniques enable the creation of biomimetic tissue engineering scaffolds. A 3D CAD model of canine trabecular bone was produced via micro CT and exported to a fused deposition modeler, to produce polybutylene terephthalate (PBT) trabeculated scaffolds and four other scaffold groups of varying pore structures. The five scaffold groups were divided into subgroups ($n=6$) and compression tested at two load rates (49 N/s and 294 N/s). Two groups were soaked in a 25 °C saline solution for 7 days before compression testing. Micro CT was used to compare porosity, connectivity density, and trabecular separation of each scaffold type to a canine trabecular bone sample. At 49 N/s the dry trabecular scaffolds had a compressive stiffness of 4.94 ± 1.19 MPa, similar to the simple linear small pore scaffolds and significantly more stiff ($p < 0.05$) than either of the complex interconnected pore scaffolds. At 294 N/s, the compressive stiffness values for all five groups roughly doubled. Soaking in saline had an insignificant effect on stiffness. The trabecular scaffolds matched bone samples in porosity; however, achieving physiologic connectivity density and trabecular separation will require further refining of scaffold processing.

Keywords

Scaffolds; Rapid prototyping; Fused deposition modeling; Micro CT; Bone

1. Introduction

Over twenty million Americans suffer pain and disability from cartilage injuries [1]. The poor natural healing response of cartilage injuries has led to the development of a number of approaches that attempt to repair tissue within cartilage defects. These techniques include soft tissue grafts and osteochondral transfer (Mosaicplasty®), as well as cell transplantation techniques such as autologous chondrocyte implantation (ACI). Recently, cartilage tissue engineering on porous three-dimensional scaffolds [2] has shown promise as a potential technique for cartilage defect repair. Scaffolds can provide a reproducible, biocompatible, and biodegradable structure capable of mimicking the mechanical characteristics of surrounding tissue while delivering cells and bioactive molecules [3–5]. Depending on the scaffold material and manufacturing technique, it is possible to develop scaffolds having one

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section upon which cartilage can be grown in vitro, and another section possessing a pore structure that closely resembles that of trabecular bone. This paper focuses on the production of scaffolds designed to facilitate bone in-growth which have a biomimetic pore structure.

Useful scaffold properties include an interconnected pore network that allows cell growth, transport of nutrients and waste, and controllable degradation and resorption rates that may potentially match tissue replacement rates [6,7]. A suitable surface chemistry [8–10] and microstructure [11–13] is necessary for cell attachment, proliferation, and differentiation [14–16]. In addition, the mechanical properties of the scaffold should be sufficient to protect new tissue from excessive loading [4,5,17], including wound contraction forces [18].

Scaffolds with interconnected pores have been produced by a variety of fabrication techniques such as solvent casting/particulate leaching [19–21], fiber bonding [22–24], and thermally induced phase separation (TIPS) [25,26]. However, concerns over incomplete evaporation of toxic organic solvents, limited reproducibility, and long preparation times using these techniques [27] have spurred interest in alternative methods. Free form fabrication (FFF) is a solvent-free technique capable of creating highly reproducible scaffold structures whose interconnected pores may vary in size and orientation across the matrix. This technique is useful for engineering scaffolds to be used at the interface between two or more tissues [17,28]. As such it can be useful for applications in which both bone and cartilage require repair. Free form fabrication techniques include stereolithography (SLA) [29–31], selective laser sintering (SLS) [29–32], 3D printing (3DP) [29–33], and fused deposition modeling (FDM) [4,29–31].

The SLA process, which consists of scanning a UV laser beam over a bath of photopolymerizable liquid polymer to build a 3D object, is currently used for the production of anatomical models for surgical planning [34]. Until recently, extensive post-processing and a lack of photopolymerizable, biocompatible/biodegradable polymers with the proper mechanical properties, have limited the use of SLA for scaffolds [35]. With the recent development of photocurable biodegradable copolymers such as poly- ϵ -caprolactone-*co*-trimethylene carbonate [36] and μ SLA devices with micrometer resolution [37], SLA has been used in a broader range of applications. Stereolithography still has its drawbacks, including the required post-processing of “green” parts by flooding with ultraviolet light, the possible formation of cavities in the part that may trap liquid resin, and the occasional need of a solvent to cleanse insufficiently drained resin from the surface [38].

SLS uses a CO₂ laser directed at a polymer or ceramic particles to raise the temperature of the powder to just above the material's glass transition temperature, fusing the particles into a solid mass [30,31]. SLS appears to be useful primarily for the preparation of scaffolds to be used in bone and it has been used to manufacture calcium phosphate ceramic bone implants from particles [39]. These scaffolds were biocompatible and osteoconductive in a canine model [40]. Disadvantages of SLS include a process more mechanically complex than most other FFF techniques, finished parts having an absorbent, powdery surface that may complicate sterilization, and that porosity may vary across the body of the finished part due to variations in particle size [41].

In 3DP, an inkjet print head sprays a binding agent over a layer of powder, after which a fresh layer of powder is rolled over the first, merging the two layers [30,32]. The process is repeated until the entire object is built, which must be fully dried before being removed from the loose powder. Although the 3DP process takes place at ambient temperature, there are several drawbacks, including a significant processing time to convert biocompatible polymers to a suitable powder, and difficulty in removing loose powder from within a porous construct [28].

Like the other free-form fabrication techniques, the fused deposition modeling (FDM) process utilizes 3D objects produced in a CAD program or from data sets created by imaging systems, to produce highly reproducible interconnected porous scaffolds. A thermoplastic polymer is heated to a semi-liquid state and extruded through a fine extrusion tip a few thousandths of an inch in diameter. The fine extruded filament is called a raster. The extrusion head assembly is guided as it moves in the X–Y plane by software that divides the 3D object into 2D slices [42]. Moving either the extrusion head or the stage and repeating the process builds the remaining layers. Pore structures are determined by the values entered into the control software for parameters such as raster thickness, raster gap width (space between rasters), and raster angle [43], while layer thickness is largely determined by the extrusion tip diameter [30]. Although “man-made” pores can be designed by adjusting software values, imaging techniques such as micro computed tomography (μ CT) can be utilized to produce a scaffold whose pore morphology mimics bone [44]. This is useful because the micro architecture of a given tissue is believed to be vital to the coordination of cellular processes, the simulation of appropriate mechanical properties, and the integration of the scaffold with surrounding tissues and microcirculation [3]. Using a CAD program, it is possible to create scaffolds produced from 3D sets with combinations of micro CT imaged data and geometric patterns. The inclusion of features, such as domes to fit the curvature of a tissue surface and cylindrical rings and struts designed to add stability and strength to the scaffold’s “trabeculated” structure, is also possible.

It was the goal of this study to develop and create polymeric scaffolds via FDM based on data obtained from a μ CT scan of canine trabecular bone and to compare the mechanical stiffness of the trabeculated scaffold to that of simple and complex porous scaffolds created using a CAD program. Morphological properties such as porosity, connectivity density, and trabecular spacing of each scaffold type and canine trabecular bone were also compared utilizing μ CT analysis software.

2. Methods

2.1. Micro computed tomography

Two samples were collected from the bones of an adult male hound. One was 12 mm in diameter and 4.5 mm long and was taken from the medial femoral condyle and one 19.1 mm in diameter and 25.4 mm long from the femoral head. The bone samples were placed into the sample holder of a desktop fan-beam micro CT scanner (μ CT-20, Scanco Medical AG, Switzerland) and scanned at medium resolution (600 X-ray projections with 512×512 pixels bitmap image, 18 μ m resolution). Integration time was set to 200 ms. Four hundred eighty-nine slices of the femoral head sample and 150 slices of the medial condyle sample were scanned, with each slice representing 30 μ m-thick segments of the sample. Once the scans were complete, 2D slices of each sample were viewed. The larger specimen collected from the femoral head was manufactured into a scaffold. To segment the slices into a 3D model, a volume of interest (VOI) was selected in the “Evaluation” program of the scanner’s software. A circle (5.6 mm diameter) was drawn onto a section of the image of the bone in the central region of the sample. Using the Morph function, the circle was copied over a range of 112 slices creating a 3D cylinder approximately 3.4 mm in length and 5.6 mm in diameter. A 3D segmentation of the VOI was done with a Sigma-Gauss value of 1.2, a support value of 2, and a threshold value of 200. Once segmentation was complete, a 3D image of the bone was created and saved into a seg.aim file.

Scanco’s image processing language (IPL) commands were used to first read the seg.aim file into memory and then convert it to an ASCII stereolithography (STL) file and export it via FTP in order to build the scaffold in the fused deposition modeler. Multiple sections of the femoral head sample were later segmented to obtain a range of values for the

morphometric properties porosity, connectivity density, and trabecular separation for comparison to those of the manufactured trabecular scaffolds.

2.2. Building the “trabecular” scaffold

Unfilled polybutylene terephthalate (Valox 315, GE Plastics, North America) pellets were melted at 250 °C and formed into a 1.70 mm diameter filament using a Haake single-screw polymer extruder (Thermo Electron Corporation) and loaded into the FDM. A commercially available support filament (P-400 Release, Stratasys, Inc., Eden Prairie MN, USA) was also loaded.

The trabecular.stl file was opened in QuickSlice, the STL-processing software accompanying the rapid prototyper (FDM 1600, Stratasys, Inc., Eden Prairie MN, USA), a device whose tolerance is ± 0.005 in. for parts less than 5 in. in diameter when using a manufacturer-provided polymer filament. The STL file was scaled up by a factor of 2.54 before being converted to slices to improve modeler resolution. After the 3D trabecular model was divided into slices, a support base (created by deposition of the P-400 Release) was added to create a barrier between the trabecular scaffold and the foam stage within the modeler. The extrusion road paths were then set using the road width 0.51 mm for the rasters and the perimeter, a raster gap width and perimeter-and-roads gap width of 0.00 mm so that the rasters would overlap each other by 0.003 in., and raster angles of 0 and 90°. The words raster and perimeter refer to the fine filament of melted polymer whose diameter is defined by the diameter of the extrusion tip (0.010 in.) and the rate of extrusion. The perimeter is the filament that is laid down along the outer edge of each slice of the object being built, while the raster is the linear filament, several of which are laid down to fill in the area of each slice. The file was exported as a Stratasys Machine Language (SML) file for upload into the FDM. The extrusion temperatures used for the model material and support material were 255 °C and 265 °C, respectively. The extrusion temperature used for the support material was recommended by the manufacturer, while the 255 °C PBT extrusion temperature was arrived at after a trial and error process of visual inspection of scaffolds produced at different PBT temperatures. After the scaffold was built, the support layer was peeled away. A photograph of the FDM in action is shown in Fig. 1. Eighteen trabecular scaffolds were created and this group was labeled 2p54trab (Fig. 2A).

2.3. Design and manufacturing of the simple linear pore and complex interconnected pore scaffolds

QuickSlice was used to create a series of slices of a cylinder 13.5 mm in diameter and 10 mm in height. Four scaffold types were created by setting different values for the raster gap width, road width, and raster angles as shown in Table 1. The simple linear pore scaffold groups, containing 18 scaffolds each, had cube-shaped pores formed by orthogonal rasters (Fig. 2B and C). The complex interconnected pore groups, containing 18 scaffolds each, had a honeycomb-like pore structure (Fig. 2D and e) achieved by giving alternating sets of two sequential slices raster angle values of 0°, 90° and 45°, -45° respectively (Table 1).

2.4. Preparation for mechanical testing

All 90 scaffolds (18 scaffolds in each of the five groups) were surface ground with a GP-25 Grinder Polisher (LECO Corp., St. Joseph, MI USA) using a very fine 1200 grit carbide paper (LECO Corp., St. Joseph, MI USA) to produce smooth, flat surfaces for compression testing. A bull's eye circular level was used to assess planeness. Following grinding, the height and diameter of each scaffold was measured using a vernier caliper (Kanon, Japan).

2.5. Mechanical testing

Eighteen scaffolds of each type were divided into three subgroups ($n=6$). One subgroup of scaffolds was tested in compression first at 49 N/s and then at 294 N/s. The second subgroup's scaffolds were soaked in 0.09% NaCl solution at 25 °C for seven days and tested in compression at 49 N/s. The third subgroup's scaffolds were soaked in 0.09% NaCl solution at 25 °C for seven days and tested in compression at 294 N/s.

Scaffolds were compressed axially on a mechanical test system (Model 810, MTS, Minneapolis, MN, USA) to assess stiffness. For each compression cycle, load (kgf) and stroke (mm) were collected simultaneously with data collection software (LabView 5.0.1, National Instruments, Austin, TX) through a Ni-DAQ data acquisition board into a Macintosh G4 computer and saved into a spreadsheet.

Maximum load, minimum load, maximum stroke, and minimum stroke of each test run on a scaffold were used for calculation of the average compressive modulus of each group of scaffolds.

2.6. Morphological analysis using micro CT

Four scaffolds from each of the five groups were placed into the sample holder of the μ CT and scanned at standard resolution (250 X-ray projections, 512×512 pixel bitmap images, with resolution of 18 μ m), with an integration time of 160 ms.

On the scanned image of the scaffold, an outline was drawn around the outer edge of the scaffold with the free-hand drawing tool. This outline was copied onto successive slices until a VOI was created encompassing the entire scaffold. A Sigma-Gauss of 1.2, a support of 2, and a threshold of 50 were used for segmentation. Porosity, connectivity density, and trabecular spacing were determined and averaged. Porosity is defined as the ratio of the volume of open space to the total volume of the sample. Connectivity density is the maximum number of trabecular connections that must be broken in order to break a trabecular sample into two parts [45]. Trabecular spacing, or separation, is the average distance between trabecular struts.

2.7. Statistical analysis

In all tests, ANOVA was performed to determine whether a significant difference between the scaffold groups was present. A Tukey HSD post hoc test was then used to determine where the differences lay, with a p -value =0.05 being statistically significant.

3. Results

3.1. Morphological analysis using micro CT scanning

Average porosity values for the five scaffold groups ranged from 62.11±0.36% for the trabecular 2p54trab group to 79.58±0.35% for the group that had the largest simple linear pores, 90deg50 (Fig. 3). There was no significant difference ($p<0.05$) between the porosity of the trabecular bone sample from which the 2p54trab group was created (57.92±0.04%) and the trabecular scaffolds (62.11±0.36%). The two scaffold types whose rasters were laid 1.27 mm apart, 90deg50 and 9045deg50 had the highest porosity values.

In comparison to the connectivity density value of 5.68±0.37 mm⁻³ for the trabecular bone sample, the connectivity density values of the five scaffold groups ranged from 0.521±0.013 mm⁻³ for the 2p54trab group to 2.505±0.084 mm⁻³ for the simple linear pore scaffold group 90deg30 (Fig. 4).

The trabecular spacing value of the bone sample was 0.43 ± 0.03 mm. The average spacing for the five experimental scaffold groups ranged from 0.628 ± 0.032 mm for the complex interconnected pore group 9045deg30 to 1.132 ± 0.009 mm for the simple linear pore scaffold group 90deg50 (Fig. 5). The average trabecular spacing value of the trabecular 2p54trab group was 1.085 ± 0.009 mm, approximately 2.3 times greater than that of the original bone sample.

3.2. Mechanical testing results

3.2.1. Dry scaffolds—When compressed at 49 N/s, the stiffness of the scaffold groups ranged from 2.46 ± 0.55 MPa for the complex interconnected pore structured 9045deg50 group to 5.11 ± 1.89 MPa for the simple linear pore structured of the 90deg30 scaffold group (Fig. 6). For both the simple and complex scaffold groups, compressive stiffness decreased as pore size increased. The 2p54trab group had a compressive stiffness of 4.94 ± 1.19 MPa, which was most similar to that of the simple linear pore group 90deg30 and significantly stiffer ($p < 0.05$) than either of the scaffold groups with complex pore structures.

For the dry scaffolds compressed at 294 N/s, the compressive stiffness values nearly doubled for most scaffold groups and ranged from 3.86 ± 1.08 MPa for the complex interconnected pore structured 9045deg50 group to 10.44 ± 2.09 MPa for the simple linear pore 90deg30 group (Fig. 7). The trabecular scaffold group had the second highest compressive stiffness value at 9.43 ± 0.93 MPa.

3.2.2. Soaked scaffolds—At the 49 N/s load rate, the compressive stiffness for scaffolds soaked in 0.09% NaCl solution for 7 days ranged from 2.05 ± 0.77 MPa for the 9045deg50 complex interconnected pore scaffold group to 5.66 ± 2.19 MPa for the simple linear pore 90deg30 scaffold group. The 2p54trab group had the second highest compressive stiffness value at $4.46 \text{ MPa} \pm 1.4 \text{ MPa}$ (Fig. 6).

For most soaked scaffold groups, the compressive stiffness values roughly doubled at the 294 N/s compression rate (Fig. 7). They ranged from 2.93 ± 1.27 MPa for the complex interconnected pore structured 9045deg50 group to 12.47 ± 1.73 MPa for the simple linear pore structured 90deg30 group, with the 2p54trab group having the third highest compressive stiffness value of 9.32 ± 2.76 MPa.

4. Discussion

The quality of trabecular bone can be characterized by its mechanical properties, chemical composition, and microarchitecture [46]. When endeavoring to create a biomimetic bone scaffold, a synthetic duplication of the chemical composition of bone is not yet possible. However, utilizing high resolution imaging technologies, free-form fabrication techniques, and a polymer or ceramic formulation with appropriate mechanical properties it is possible to closely simulate the mechanical properties and microarchitecture of trabecular bone.

The volume fraction (and porosity) of human cadaveric bone samples from the femoral neck have been reported as 0.29 ± 0.05 (porosity $71 \pm 5\%$) by Bayraktar et al. [47] and 0.16 ± 0.07 (porosity $84\% \pm 7\%$) by Kabel et al. [48]. Bayraktar et al. calculated the volume fraction for their samples using the Archimedes principle, while Kabel et al. did so for their samples using a 3D morphometric program. Porosity values for trabecular bone were also reported [49] using cylindrical transiliac crest trabecular bone samples obtained from 70 patients (32 women, 38 men, with a mean age 68 ± 16 years). These samples underwent 3D morphometric analysis via μ CT, then were sectioned for 2D histomorphometric analysis to allow comparison of the two techniques. The mean BV/TV ratio reported by Muller et al. [49] was $14.48 \pm 5.34\%$ for the μ CT-analyzed samples, corresponding to a mean porosity of $85.52 \pm$

5.34%. These published porosity values [47–49] are on the same order as the porosities of the five scaffold types produced for this study. The mean porosity of the trabecular scaffold group was less than that of human cadaveric samples reported in the literature, but nearly identical to that of the porous canine femoral neck sample from which it was produced.

In a study assessing the impact of the connectivity density of cancellous bone on its elastic properties [48] fifty-five trabecular specimens from six anatomical regions (calcaneous, proximal tibia, distal and proximal femur, proximal humerus, and lumbar vertebral bodies) were collected and microtomed to a slice thickness of 20 μm to 25 μm . Using digital photos of the slices, they were digitally reconstructed into a 3D voxel model for micro finite element analysis and morphological analysis. The connectivity density for this series was reported as $3.36 \pm 1.50 \text{ mm}^{-3}$ and the volume fraction for these specimens was 0.16 ± 0.07 with a corresponding porosity of $84 \pm 7\%$. Though the 2p54trab scaffold group was created from a μCT scan of a canine trabecular bone sample, it was the four scaffold groups with designed pore structures whose connectivity density values fell within the range of these published observations [48]. The mean connectivity density value for the trabecular-like scaffold group ($0.521 \pm 0.013 \text{ mm}^{-3}$) was lower. This may have been the result of the resolution limitations of the fused deposition modeler used in this study because the modeler was originally designed to build larger models. The 3D scaffold CAD model was scaled up 2.54 times to gain better resolution, but it is apparent that the process must be further refined or that a modeler with better resolution must be utilized to achieve scaffold connectivity densities that better match natural bone.

Muller et al. [49] reported mean separation values for the μCT -evaluated samples of $0.77 \pm 0.35 \text{ mm}$, which are on the same order as those of the scaffolds in the trabecular scaffold group in this study. However, the mean calculated trabecular spacing was significantly less than that of the canine sample from which the 3D CAD model was created (0.47 mm). This may also be due to the aforementioned modeler resolution issue.

Fused deposition modeling has been used to produce tissue engineering scaffolds with designed, interconnected pore structures like the simple and complex porous scaffolds produced for this study for both in vitro and in vivo studies. Several investigators [4,50–52] have used fused deposition modeling to produce porous polymer scaffolds with interconnected pore structures. In this study, FDM was utilized to produce scaffolds made of polybutylene terephthalate (PBT), a semi-crystalline polyester that has been used in copolymer formulations in other biomedical applications [52] and which can be readily formed into a suitable filament for the modeler used. Hutmacher et al. [4], Cao et al. [50], and Darling and Sun [51] used fused deposition modeling to produce polycaprolactone (PCL) scaffolds. Hutmacher et al. reported porosity values, measured by an ultrapycnometer, as 55% and 56% for their two complex scaffold designs. However, using porosimetry, scaffold porosity was measured as $61 \pm 1\%$ for both scaffold designs [4]. That was nearly identical to the porosity measured for the trabecular scaffolds produced in this study. Hutmacher et al. also performed mechanical testing of the polycaprolactone scaffolds reporting a compressive stiffness of $41.9 \pm 3.5 \text{ MPa}$ for dry scaffolds with a honeycomb-like raster angle pattern of $0^\circ/60^\circ/120^\circ$ and $29.4 \pm 4.0 \text{ MPa}$ for the same scaffolds that had been soaked in a phosphate-buffered saline solution at 37°C for 1 day prior to compression testing [4]. The scaffolds produced with a more complex pore structure (raster angle pattern $0^\circ/72^\circ/144^\circ/36^\circ/108^\circ$) had a lower dry compressive stiffness of $20.2 \pm 1.7 \text{ MPa}$. The compressive stiffness value for the saline-soaked scaffolds was essentially unchanged when compared with unsoaked scaffolds at $21.5 \pm 2.9 \text{ MPa}$ [4]. Like the more complex structured scaffolds produced by Hutmacher et al., the 9045deg30 and 9045deg50 scaffold groups in our study also had significantly lower stiffness values than their simpler counterparts. Overall, the dry compressive stiffness values for the polycaprolactone scaffolds were

substantially higher than the values observed for the PBT scaffolds tested in the present study. However, unlike the PCL scaffolds, soaking the scaffolds in saline at 25 °C for seven days did not cause a change in stiffness for the five PBT scaffold groups.

Darling and Sun [51] produced porous polycaprolactone scaffolds via FDM and used micro computed tomography to analyze multiple structures produced with two different extrusion tip sizes (10 mils and 7 mils). 2D analysis of CT slices was performed to assess strut width and pore width of the samples. Pore width, which is analogous to trabecular separation, for the 10 mil samples ranged from $196 \pm 56 \mu\text{m}$ ($0.196 \pm 0.056 \text{ mm}$) to $292 \pm 40 \mu\text{m}$ ($0.292 \pm 0.040 \text{ mm}$) and from $120 \pm 62.8 \mu\text{m}$ ($0.120 \pm 0.0628 \text{ mm}$) to $772 \pm 33.5 \mu\text{m}$ ($0.772 \pm 0.0335 \text{ mm}$) for the 7 mil samples [51]. The trabecular separation measurements reported in the present study were generally higher than the pore width values reported by Darling and Sun [51]. This could be due to the fact that a 12 mil extrusion tip was used and a larger raster gap width setting. Darling and Sun did not report raster gap width settings for their scaffold designs.

An alternative rapid prototyping technique was used in conjunction with micro CT to produce biomimetic trabecular bone scaffolds by Peng et al. [53]. 3D gel lamination, a process similar to 3D printing, was used to translate 3D data sets created from micro CT scans of canine femoral head samples into biphasic calcium phosphate ceramic scaffolds. The authors claim that morphometric analysis revealed no significant difference ($p < 0.05$) between the trabecular bone samples and trabecular scaffold samples with respect to BV/TV (porosity), trabecular thickness (Tb.Th), trabecular number (Tb.N), or trabecular bone pattern factor (Tb.Pf) [53]. Tb.Th, Tb.N, and Tb.Pf, were not assessed in the present study, but the reported 62% porosity [53] was nearly identical to that of the polymer trabecular scaffolds produced for the present study. The biphasic calcium phosphate scaffolds likely had a more similar radio-opacity to trabecular bone, enabling use of more similar micro CT threshold values between the bone samples and scaffold samples. This may have aided comparison of morphometric features between their structures. The ceramic material accounted for the large disparity in average compressive stiffness values reported by Peng et al. [53] ($0.464 \pm 0.036 \text{ GPa}$) and those of the trabecular bone scaffolds reported in the present study, which were also significantly lower than the compressive stiffness of trabecular bone, which has been reported as $0.616 \pm 0.707 \text{ GPa}$, $0.174 \pm 0.084 \text{ GPa}$, and $0.063 \pm 0.007 \text{ GPa}$ [54] for femoral neck samples compressed along the x , y , and z axes respectively and as $18.0 \pm 2.8 \text{ GPa}$ by Bayraktar et al.

Fused deposition modeling shows promise as a versatile, solvent-free tissue-engineering scaffold manufacturing technique that requires virtually no post-processing. More work done to match the resolution of scaffold manufacturing processes to that of the imaging techniques that have made possible the creation of biomimetic structures will be valuable, but it has already been shown that the current process is capable of matching trabecular bone porosity. In combination with free form fabrication, micro computed tomography is useful for generating 3D scaffold structures for rapid prototyping use, as well as morphological analysis of manufactured structures.

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Fig. 1.
Photograph of the FDM-1650 building a scaffold.

Fig. 2. Digital photographs and micro CT 3D segmentations of polybutylene terephthalate scaffolds. The scaffolds are photographed on top of a penny for reference. Scale bars in 3D segmentations represent 1 mm. A: 2p54trab, B: 90deg30, C: 90deg50, D: 9045deg30, and E: 9045deg50.

Fig. 3. Scaffold group porosity comparison. “k9femhead” denotes the femoral head trabecular bone sample.

Fig. 4.
Comparison of connectivity density of the five scaffold groups with the femoral head bone sample.

Fig. 5.
Comparison of trabecular separation among the five scaffold groups and the femoral head bone sample.

Fig. 6.
Scaffold stiffness at the 49 N/s compression load rate.

Fig. 7.
Scaffold stiffness at the 294 N/s load rate.

Table 1

QuickSlice parameters for the designed simple linear pore and complex interconnected pore scaffold groups

Scaffold group	Scaffold type	Raster gap width	Road width	Raster angles
90deg30	Simple small	0.76 mm (0.030")	0.51 mm	0°, 90°
90deg50	Simple large	1.27 mm (0.050")	0.51 mm	0°, 90°
9045deg30	Complex small	0.76 mm (0.030")	0.51 mm	0°, 90°, 45°, -45°
9045deg50	Complex large	1.27 mm (0.050")	0.51 mm	0°, 90°, 45°, -45°

“Small” and “large” refer to the relative pore sizes of each design