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Title: Evaluating the effect of increasing ceramic content on the mechanical properties, material microstructure and degradation of selective laser sintered polycaprolactone/β-tricalcium phosphate materials.

Author Names: Heather Doyle\textsuperscript{a}; Stefan Lohfeld, Dr.\textsuperscript{a}; Peter McHugh, Prof.\textsuperscript{a}

Author Affiliations: \textsuperscript{a}Biomechanics Research Centre (BMEC), Mechanical and Biomedical Engineering, National University of Ireland, Galway

Corresponding Author: Heather Doyle

Email: heatherdoyle@nui.ie

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Highlights:

- Polymer-ceramic orthopaedic scaffold materials were fabricated using selective laser sintering.
- The effect of increased wt% of β-TCP on mechanical properties and microstructure were evaluated.
- Increasing wt% of β-TCP resulted in reduced strength and a significant increase in stiffness relative to no β-TCP.
- The effect of β-TCP content on the accelerated degradation of materials was evaluated.
- Degradation resulted in reduced elastic modulus and increased strength with increased wt% β-TCP.
1 Abstract

Orthopaedic scaffold materials were fabricated from polycaprolactone (PCL) and composite PCL- β-tricalcium phosphate (PCL/β-TCP) powders using selective laser sintering (SLS). Incorporating β-TCP particles is desirable to promote osteogenesis. The effects of increasing β-TCP content on the material’s mechanical properties and microstructure were evaluated. The wt% of β-TCP and PCL particle sizes were found to influence material microstructure and mechanical properties, with increasing ceramic content causing a small but significant increase in stiffness but significant reductions in strength. Degradation of materials was achieved using accelerated ageing methods. The influence of β-TCP content on degradation at seven weeks was evaluated through changes in mechanical properties and microstructure, and the ceramic particles were found to reduce elastic modulus and increase strength. The results of this study highlight the influence of ceramic content on mechanical properties and degradation behaviour of PCL/β-TCP SLS materials, and indicate that these changes must be considered in the design of scaffolds for critical-sized defects.

Keywords: selective laser sintering; polycaprolactone, B-tricalcium phosphate; bone tissue engineering; mechanical properties; degradation; accelerated ageing.
2 Introduction

Biodegradable materials are of significant interest in the field of bone tissue engineering scaffolds. In the case of large bone defects in particular, such scaffolds can provide structural support during the initial stages of healing, and degrade over time allowing the body to replace the scaffold with new, native bone tissue. This is advantageous as in the long-term the defect is filled with bone produced naturally by the body. Synthetic calcium phosphates such as hydroxyapatite and β-tricalcium phosphate (β-TCP) are biodegradable ceramics that are similar in composition to natural bone [1] and are commonly used in orthopaedic applications [2–10]. Incorporating ceramic particles in polymer-based scaffold materials has been shown to have significant osteogenic benefits and to encourage new bone formation on scaffold surfaces [2,11,12]; however fully ceramic scaffolds can be quite brittle. Selective laser sintering (SLS) can be used to fabricate scaffolds from biodegradable polymer-ceramic composites, such as polycaprolactone (PCL) with β-TCP, in virtually any geometry, although scale is a limitation, making the fabrication of scaffolds to fit patient-specific defects a reality [13]. This type of composite scaffold also has the advantage of being less brittle than a fully ceramic scaffold.

For orthopaedic scaffolds designed for critical-sized defects where a large portion of the body-weight is supported by the scaffold, the stiffness of the scaffold material is a critical design factor. Given the relatively high stiffness of ceramic materials in general, it is expected that increasing the ceramic content of a polymer-ceramic composite scaffold will result in an increase in stiffness. This is true for polymer-ceramic composite materials where a homogeneous cross-section can be achieved [14–18]. However, the stiffening effect of ceramic particles in polymer-ceramic SLS materials has been shown to be less than for solid materials with the same polymer-ceramic ratio [19], which is a significant concern for the design of load-bearing scaffolds. Increasing the ceramic volume impedes sintering of polymer-ceramic materials due to settling of ceramic particles on the surface of the polymer particles [10]. It is important to fully understand the effects of increasing ceramic content...
on the sintering of scaffold materials, and their mechanical properties, to ensure effective scaffold design and in-vivo performance.

The change in mechanical properties of biodegradable scaffold materials during degradation is another critical design factor. The influence of the in-vivo environment on the scaffold material mechanical properties needs to be fully understood, particularly for load-bearing scaffolds. Few studies report on the degradation behaviour of PCL/β-TCP materials fabricated using SLS, however it is known that the ceramic content of composite materials [20] and surface area [21] both influence the in-vivo material degradation rate.

Retrieval studies are the most accurate way to evaluate the in-vivo performance of biodegradable materials, however in reality this is not always practical. Laboratory testing (room temperature, dry conditions) is generally acceptable to evaluate mechanical properties, but does not necessarily accurately evaluate in-vivo material performance. In-vivo conditions can be replicated in-vitro by exposing materials to simulated physiological conditions by immersion in phosphate buffered saline (PBS) at 37°C and pH 7.4. PCL has a long degradation time (2-4 years in-vivo [22–24]); long degradation times can be prohibitive for the timely development of new medical implants and devices. Accelerated ageing methods have been developed to increase degradation reaction rates so that polymer degradation can be assessed at reduced length scales through the use of temperature [25], pH [20,26] and the addition of enzymes to degradation media [26–28]. Temperature-based accelerated ageing methods can be characterised by assuming that the chemical reactions of the material degradation are described by the Arrhenius reaction rate function, which predicts that a 10°C increase in temperature doubles the reaction rate (ASTM F1980-02, Hukins et al. [25]). A major advantage of this approach is the clear relationship between the increase in temperature and degradation time, and data from accelerated ageing studies can relate back to real-time (degradation at 37°C) data.
The objective of this study was to evaluate the impact of increasing ceramic content in polymer-ceramic SLS materials on the structure and mechanical properties of the resulting materials and their suitability for orthopaedic applications. SLS scaffold materials were fabricated from PCL and from PCL with two volume fractions of β-TCP. Firstly, the effect of increasing ceramic content on material microstructure and elastic properties was assessed. Secondly, degradation of these materials was achieved using accelerated ageing to replicate in-vivo degradation at 14 weeks, and changes in mechanical properties and material microstructure were evaluated. Discussion of the implications of the results for the performance and suitability of the material for the intended application is also included.

3 Materials and Methods

3.1 Materials and Fabrication

PCL powder (CAPA6506, Solvay, UK) (average molecular weight (Mn) 60,000, particle size 600µm) was cryogenically ground at Noll Aufbereitungstechnologie GmbH (Bobingen, Germany) to produce particle sizes in two ranges: < 50µm and 50 - 110µm. Particles were separated into two particle size ranges using sieving. β-TCP (Fluka) (molecular weight = 310.18g/mol, particle size = 3-5µm) was mixed with the PCL powder until well blended to produce PCL/β-TCP powder blends for sintering with ratios of 90/10wt% (3vol% β-TCP) and 50:50wt% (25vol% β-TCP).

Individual scaffold struts were fabricated using SLS [29] in three ratios of powder blends: 100% PCL, 90/10wt% PCL/β-TCP and 50/50wt% PCL/β-TCP. The PCL and 90/10wt% PCL/β-TCP struts were fabricated using the larger PCL particle size to replicate the scaffolds used in Lohfeld et al. [10]. The 50/50wt% PCL/β-TCP struts were fabricated using the smaller PCL particle size as the inclusion of large quantities of β-TCP particles were found to inhibit sintering [10]. The outline scan method [30] was used to achieve a small strut cross section with a laser power of 7W, laser spot diameter of 410µm, part bed temperature of 49°C and powder layer thickness of 0.11mm with the Sinterstation
2500plus system (DTM, USA). The intended dimensions of each strut were: 13mm length; cross section 0.55mm x 0.55mm.

Strut tensile samples (Figure 1) were prepared by gluing each strut end inside plastic tubes (inner diameter 1.6mm) with a cyanoacrylate adhesive (Loctite® Power Easy™) for gripping during tensile testing [29]. Sample mass (m₀) was measured and recorded.

Strut tensile sample assembly fixation was significantly more challenging for the PCL struts compared to the composite struts, and frequent strut pull-out during tensile testing was observed due to incomplete glue bonding (corresponding data consequently not used). Further to extensive efforts to address fixation, improved results were found for smaller diameter plastic tubes (inner diameter 1.3mm), reducing the bond gap between the strut and the inside of the tubes; bond gap is an important factor in the bond strength of cyanoacrylate glues [31]. These fixation issues resulted in a variation in successful tensile test sample numbers (n) at each time point: PCL n = 4, 3, 5, for tests at weeks 1, 4, 7, respectively; 90/10wt% n = 4, 8, 9, for tests at weeks 1, 4, 7, respectively; 50/50wt% n = 7, 8, 10, for tests at weeks 1, 4, 7, respectively.

3.2 Accelerated Degradation

Accelerated degradation was achieved using a degradation temperature of 47°C (i.e. +10°C) to double the reaction rate relative to a baseline body temperature of 37°C, assuming the Arrhenius reaction rate function (ASTM F1980-02, [25]). Although this assumption is employed here for practicality, note that this is a generalised correlation that might not necessarily relate exactly to the degradation rate of PCL. A real-time degradation time of 14 weeks was chosen to compare with the in-vivo performance of 90/10wt% PCL/β-TCP scaffolds in a study by Lohfeld et al. [10], corresponding to 7 weeks accelerated testing time. Note that the “10°C rule” is intended for use below the polymer glass transition temperature (Tₐ), and is most commonly used with poly(l-lactide) (Tₑ of between 60 and 65°C [32]). PCL, however, has a Tₑ of -60°C and a melt temperature (Tₘ) of 60°C [32], so although the degradation conditions presented here are above the Tₑ of PCL, so too are physiologically
representative degradation conditions at 37°C, and this method of accelerating degradation has been used successfully for PCL for temperatures of 45-50°C [33,34].

To simulate in-vivo degradation conditions, tensile samples of each material were submerged in one litre of PBS in sealed containers (pH 7.4), with one container per material. The containers were placed in an oven at 47°C and both temperature and pH were monitored and remained constant throughout the course of the study, with no adjustments required.

3.3 Mechanical Testing

The primary tensile testing reported here is for degraded PCL, 90/10wt% and 50/50wt% struts at 1, 4 and 7 weeks under accelerated degradation conditions, or 2, 8 and 14 weeks real-time degradation. Samples were removed from PBS and blotted with paper towels prior to testing at each time point. Tensile testing of undegraded PCL struts was also carried out. Additionally, previously reported tensile data for undegraded 90/10wt% [29] and 50/50wt% [35] struts are reported, including stress-strain and elastic modulus (E) data.

All tensile testing was carried out on a Zwick biaxial testing machine as per ISO 527-1. Strain and sample cross-sectional area (CSA) were measured using a calibrated video extensometer camera. The overall strut elastic modulus (E) was calculated from the linear portion of the tensile curve up to a strain of 0.005 using a trend line function. The tensile strength (σ_s) was calculated as the maximum stress in the tensile curve and the corresponding strain at this value (ε_s) was recorded. The strain to failure was also recorded. Following tensile testing, each sample was dried to a constant weight in an oven at 47°C and the mass following degradation (m_d) was recorded. Percentage mass loss (%m) was calculated for each sample using the following formula: %m = (m_0 - m_d)/m_0*100.

3.4 Structural and Morphological Characterisation

Scanning electron microscopy (SEM) was carried out on undegraded and degraded (seven weeks) dried samples of each material post tensile testing to evaluate changes in surface morphology during
degradation. Strut segments were gold-coated before imaging at magnifications of x100 and x1000 using a Hitachi S-4700 SEM (Hitachi, UK) at a power of 10kV. EDX was used to identify regions of PCL and β-TCP. High resolution images of the strut materials (approximate dimensions 0.1mm x 0.1mm x 0.1mm) were obtained posttensile testing, undegraded and degraded, using micro-CT scanning (µ-CT) with a voxel size of 3μm (MicroCT 100, Scanco Medical, Switzerland).

4 Results

4.1 Mechanical Testing

Individual scaffold struts of each material were successfully fabricated using SLS. Average tensile stress-strain curves for undegraded materials are presented for full curves and the linear portion of curves in Figure 2. A trend of decreasing $\sigma_s$ and $\varepsilon_s$ with increasing ceramic content is visible (Figure 2A). Data for $E$, $\sigma_s$, $\varepsilon_s$ and CSA for each strut type (undegraded) is presented in Figure 3. There is negligible difference in the $E$ for the two composite struts but PCL struts are less stiff (Figure 3A). A trend of decreasing $\sigma_s$ and $\varepsilon_s$ with increasing ceramic content is visible (Figures 2A, 3B, 3C). No significant difference in CSA was observed for the PCL and 90/10wt% struts while the 50/50wt% strut CSA was greater than the PCL strut CSA.

Averaged tensile curves for day zero and week seven are shown in Figure 4 for each material for comparison purposes and to illustrate the scatter observed in test data. The full set of average tensile curves are shown for PCL (Figure 5A), 90/10wt% (Figure 5C) and 50/50wt% struts (Figure 5E), with undegraded tensile curves included for reference. Linear portions of the tensile curves are shown in Figure 5(B, D, and F). Variation in sample sizes (Figures 2 to 5) is due to the strut pull-out issue described above. No increase in test data scatter was observed at week seven in comparison to undegraded test data (Figure 4).

The change in $E$ during degradation is shown in Figure 6A. No reduction in $E$ was observed at seven weeks for the PCL struts, however a 63% and 54% drop in $E$ was observed for the 90/10wt% and
50/50wt% struts respectively. The largest reduction in $E$ occurred in the first week of degradation for the 90/10wt% material (57%) and in the first four weeks for the 50/50wt% material (52%).

A reduction in $\sigma_s$ was observed for all materials, with reductions of 78% for PCL, 68% for 90/10wt% and 66% for 50/50wt% strut at seven weeks, Figure 6B. The largest decrease in $\sigma_s$ occurred in the first week, with little further decrease observed in the remaining six weeks. A similar increase in $\epsilon_s$ was observed for the 90/10wt% and 50/50wt% struts (95% and 94% respectively), however an 18% decrease in $\epsilon_s$ was observed for the PCL struts, Figure 6C. No significant change in CSA or mass loss were observed during degradation for any of the strut types, indicating that no swelling due to water update or reduction in CSA due to degradation of the materials occurred.

The failure strain of PCL samples (0.585 ± 0.250) was greater than for the 90/10wt% and the 50/50wt% struts (0.0696 ± 0.0107 and 0.062 ± 0.011, respectively). Failure strain in PCL struts decreased by 25% after seven weeks degradation, whereas failure strain in the composite struts increased by 74% and 32% for 90/10wt% and 50/50wt%, respectively. This difference in ductility is likely due to the potential change in degradation kinetics that could occur with the incorporation of ceramic particles, relative to the pure PCL.

### 4.2 Statistical Analysis

The elastic modulus and CSA of each strut type were compared using a one-way ANOVA test (Figure 3). No statistical difference in $E$ was observed for the 90/10wt% and 50/50wt% struts ($P > 0.05$). No statistical difference in CSA was observed for the PCL and 90/10wt% struts ($P > 0.05$) while the CSA of the 50/50wt% struts was significantly greater than the PCL struts ($P < 0.001$). No statistical difference between the undegraded and degraded CSA was observed for PCL struts ($P=0.5308$), 90/10wt% struts ($P=0.9527$) or for 50/50wt% struts ($P=0.0820$) when compared using a two-tailed unpaired T-test. Mass loss at seven weeks was not significant for any material when compared to dried, undegraded control samples for PCL ($P = 0.1790$), 90/10wt% struts ($P = 0.7044$) or for 50/50wt% struts ($P = 0.8437$) when compared using a two-tailed unpaired T-test.
4.3 Structural and Morphological Characterisation

SEM images of each material are shown in Figure 7 (magnification x100) and in Figure 8 (x1000). β-TCP particles are visible as small bright regions, indicated by arrows in Figure 8, whereas the larger PCL particles have smoother surfaces. 50/50wt% struts have a finer particulate nature than the PCL and the 90/10wt% struts due to differences in PCL particle size, indicating that a greater degree of partial sintering occurred at the surface in the PCL and 90/10wt% struts than in the 50/50wt% struts (Figure 7). β-TCP particles are visible on the surfaces of the PCL particles in Figure 8, verified by EDX. Slightly less individual PCL particles attached at the surface of each strut are visible post-degradation (Figure 7) indicating a small degree of removal of surface particles.

The material microstructures are visible in the µ-CT scan images (Figure 9), where the bright pixels are β-TCP and the less bright pixels are PCL. Larger regions of fully sintered material are visible in the PCL struts than the composite struts, and more continuous material is visible in the 90/10wt% struts than the 50/50wt% struts, in which larger pores are visible. An increase in porosity in all three materials is visible in the µ-CT images at seven weeks degradation compared to the undegraded materials. Regions of β-TCP are present at the outer edges of PCL particles in the 90/10wt% struts, both for individual partially sintered particles and through the cross-section of fully sintered regions of material indicating the original boundaries between particles.

5 Discussion

The results show that good sintering was achieved for each material type, and the elastic properties indicate that all are suitable for scaffold fabrication, even with the inclusion of relatively large volumes of ceramic particles (50/50wt%). This is significant as previous studies have indicated difficulties in sintering materials with large volumes of ceramic particles [10]. Increased porosity in composite struts compared to PCL struts (Figure 9) is a result of partial sintering which occurs when the sintering energy is below a minimal threshold required for full sintering. When this occurs, only
partial sintering of the outer regions of the particle is achieved, creating a highly porous structure with particles connected at contact points by ‘necks’ between particles, Figure 10A.

The volume of ceramic in each sintered material affects the microstructure of the resulting material (Figure 10B). The smaller β-TCP particles can be seen coating the outside of the PCL particles (Figures 7 to 9). In a two-phase polymer-ceramic composite, sintering is carried out above the Tm of the polymer phase but below the sintering temperatures of the ceramic phase. Therefore the ceramic particles do not get sintered and act to inhibit the flow and consolidation of the polymer melt. As a result greater porosity is visible in the 90/10wt% struts than the PCL struts (Figure 9). Increasing the volume of β-TCP particles in the material accentuates this effect resulting in a lower degree of sintering, and as a result greater porosity is visible in the 50/50wt% struts than the 90/10wt% struts (Figure 9).

A reduced PCL particle size of 0-50μm was required for the fabrication of the 50/50wt% struts due to the increased β-TCP volume and the resulting difficulties in sintering [10]. The CSA of the PCL and 90/10wt% struts were the same, but the 50/50wt% struts had a greater CSA suggesting that CSA is controlled by the PCL particle size used. This is useful for planning the fabrication of parts using SLS materials; smaller particle sizes can be used to obtain parts with lower porosity, however reduced cross-section (i.e. a larger degree of shrinkage) should be expected. In general, this can be related to the ratio of PCL particle size (ØPCL) to laser spot diameter (ØL) used during fabrication, as illustrated in Figure 10C. When ØPCL is less than ØL, multiple particles are sintered directly by the laser (Case A), whereas when ØPCL is greater than ØL, only single particles in the laser path are directly sintered (Case B). Heat energy is transferred through contact points between particles forming partially sintered regions of material connected by a “neck” of sintered material. For Case A, a large number of particles are directly sintered and a large number of neighbouring particles are partially sintered, resulting in a greater CSA, but also in increased porosity.
The results presented here indicate that incorporating β-TCP particles has a stiffening effect, with $E$ increases of 52.6% and 42.5% for β-TCP inclusions of 10wt% and 50wt%, respectively, relative to the PCL struts (Figure 3). However, increasing ceramic content from 10wt% to 50wt% did not result in a corresponding increase in measured $E$ (Figure 3, $P=0.6254$, two-tailed unpaired T-test). For the 50/50wt% struts, the reinforcing effect of the greater volume of the β-TCP particles is offset by the greater degree of microporosity (Figure 9), resulting in a similar overall strut $E$ to the 90/10wt% struts.

The inclusion of β-TCP particles has a more significant effect on $\sigma_s$ and $\varepsilon_s$ than on $E$. For the 90/10wt% and 50/50wt% struts respectively, reductions in $\sigma_s$ of 41.2% and 68.9% and reductions in $\varepsilon_s$ of 67.9% and 81.0% were observed in comparison to the PCL struts (Figures 2 and 3). The PCL struts are more ductile than the composite struts with failure strains 5.3 times greater than $\varepsilon_s$, compared to 179% and 216% for 90/10wt% and 50/50wt% struts, respectively (Figure 2A). This is due to the greater number of pores and the lower amount of fully sintered material in the loading direction in the composite struts. Increasing the ceramic content has the effect of further reducing $\sigma_s$ and $\varepsilon_s$ due to the corresponding increase in porosity (Figures 2 and 3). When the sintered struts are loaded in tension, forces are transferred through the bonds or necks between individual sintered particles, and failure of these bonds is evident from reductions in stress at high strains (Figure 2). The failure behaviour of these sintered materials is therefore dictated by the degree of sintering between particles, which in turn is reduced by increasing ceramic content. Similar results were observed by Eshragi et al. [19], where the strength and yield strain of polymer-ceramic SLS scaffolds were not seen to improve with increasing ceramic content.

In this work, the strut mechanical properties were determined through tensile testing, as compression testing would not have been practical due to a buckling risk (as verified by initial trials). This results from the sample geometry, which was selected to most accurately replicate the individual struts in the full SLS scaffolds used in an animal trial by Lohfeld et al. [10]. Greater stiffness
and ultimate strength would be expected if the material properties of these SLS materials were evaluated using compression testing, as a result of material compaction during compression. It is important to emphasise that stiffness and strength are both independently important for these implantable scaffold materials, and can to a certain extent be controlled independently through variations in material processing. The stiffness controls how stress is distributed to the adjacent tissue, either loading the tissue if the stiffness is low, or (stress) shielding the tissue if the stiffness is high, while the strength relates primarily to the implant itself and the point at which it will mechanically fail.

For optimal new bone formation, scaffold stiffness mimicking that of developing bone encourages osteogenic differentiation of mesenchymal stem cells (MSCs) [36], whereas scaffold stiffness similar to mature native bone encourages the infiltration of bone-forming cells from the surrounding native tissue [37]. While good bone formation has been observed with softer scaffolds [38,39], they are too soft for use in critical-sized defects; extensive external fixation would be required in order to support loading in-vivo and as a result little or no loading would be transferred through the scaffold itself. As it is known that local strains are required to encourage new bone formation [40,41], this situation is not desirable. Conversely, excessively stiff scaffolds can cause stress concentrations in the surrounding healthy tissue resulting in bone resorption [42]. Stiffness values in the range 3.6 to 13.5 MPa have been measured for ovine callus tissue [43]. The $E$ values for the materials reported here are for single strut samples, whereas a full 90/10wt% scaffold with designed macropores as presented in Lohfeld et al. [10] is less stiff, with macroscale compressive $E$ values of 6.6 to 8.7 MPa, which are within the above suggested range.

The elastic modulus of the PCL struts was unchanged after seven weeks degradation relative to the undegraded material, as expected from the long degradation times reported for PCL, ranging from 2 to 4 years in-vivo [22–24]. An initial drop in $E$ for the PCL struts (Figure 6A) is possibly due in part to initial water absorption on immersion in PBS (and resulting material plasticisation). However, it is
not possible to make definitive conclusions in relation to this based on data generated here; the observed effect is actually more likely due to the limited sample numbers at this specific time point which resulting from difficulties in tensile sample gluing, as discussed above. Reductions of 63% and 54% in $E$ were observed for the 90/10wt% and 50/50wt% struts, respectively, at seven weeks. However, the resulting lower $E$ values are still within the suggested range of values required to stimulate bone formation, indicated above. This greater reduction in $E$ for composite struts, relative to the PCL struts, could be due to a number of factors including the inclusion of $\beta$-TCP particles, which have a shorter degradation time [44–48] compared to PCL. The composite struts also have greater porosity than the PCL struts (Figure 9) therefore greater surface area is exposed to the degradation medium, which is known to influence degradation [21]. Degradation of the bonds between PCL and $\beta$-TCP particles may be occurring, which would also influence degradation in mechanical properties. However, this would be difficult to determine because of the size scale of the $\beta$-TCP particles.

Reductions in $\sigma_s$ were observed for all materials at seven weeks, with a 10% greater decrease for PCL struts than for composite struts, where $\beta$-TCP particles are still visible embedded in the PCL matrix in µ-CT scans at seven weeks. While degradation of the PCL matrix is likely occurring (as evidenced by the reduction in $\sigma_s$ for the PCL struts, Figure 6) the $\beta$-TCP particles still provide reinforcement. This could be confirmed by evaluating changes in molecular weight, however this was outside of the scope of this study. As no further reduction in $\sigma_s$ occurred after one week, the idea of hydrolytic degradation of the polymer is not supported.

Similar trends of increasing $\varepsilon_s$ were observed for the composite struts, however a decrease of 18% was observed for the PCL struts. The trend of reduced $\varepsilon_s$ with increasing ceramic content is visible after seven weeks. Note that it is difficult to make conclusions regarding the effect of increasing ceramic content on the mechanical properties or degradation behaviour of the materials due to differences in PCL particle sizes between the 90/10wt% and 50/50wt% materials.
Changes in mechanical properties of the SLS scaffold materials due to degradation are important considerations for the design and performance of orthopaedic scaffolds. The reduction of material properties during degradation must be taken into account to ensure that loads are supported even after a period of degradation, in particular for critical sized defects. Degradation of the scaffold itself is desirable to allow for new, native bone to replace the scaffold and to fill the defect space over time, eventually replacing the scaffold. Newly-formed bone contributes to the overall stiffness of the scaffold and defect [49] which, depending on volume and quality, would offset the resulting reduction in stiffness.

Changes in scaffold material properties and the resulting changes in the local tissue mechanical environment will influence tissue healing and new bone formation; it is known that scaffold stiffness influences the differentiation of MSCs [50] and mechanical strain influences the regeneration of bone tissue [41,51]. Therefore the changes in \( E \) and \( \varepsilon_s \) in each of the scaffold materials considered here will play a role in native bone tissue adaptation. Changes in material properties during degradation were found to be dependent on the constituent materials; material property changes were similar for the composite struts, but differences between PCL/\( \beta \)-TCP struts and PCL struts were observed. In all cases (degraded and undegraded), increasing \( \beta \)-TCP content was found to decrease the strength of the sintered materials, and struts with \( \beta \)-TCP were less stiff than PCL struts.

Performance of an orthopaedic scaffold with the same material composition as the 90/10wt% struts presented here has been demonstrated in a pre-clinical study by our group [10]. As presented here, the incorporation of a 50wt% volume of \( \beta \)-TCP particles in this type of scaffold material resulted in only a slight reduction in the mechanical properties, and only a slight increase in degradation, compared to the 90/10wt% material. As the purpose of the increased \( \beta \)-TCP content in the 50/50wt% strut materials is to improve osseointegration, the results here suggest that the small reduction in mechanical properties with increased ceramic content is more than compensated for by the expected increase in bone formation [2,10–12]. Therefore, we recommend that in-vivo performance of this type of scaffold material should be further investigated.
6 Conclusions

For PCL/β-TCP SLS scaffolds it can be concluded that β-TCP wt% and PCL particle size influence the resulting material microstructure and mechanical properties. The β-TCP resulted in a slight increase in stiffness compared to pure PCL, however increasing ceramic content from 10wt% to 50wt% did not further increase stiffness. Composite struts had significantly reduced strength and strain at maximum stress relative to PCL struts, with increasing ceramic content causing a further significant reduction in both strength and strain. The incorporation of β-TCP influenced the change in material properties during degradation. The reduction in stiffness due to degradation was greater for composite struts than for PCL struts. Materials containing β-TCP became more ductile due to degradation; conversely, PCL struts became less ductile.

These results illustrate the complexity of polymer-ceramic SLS material microstructures, and the influence of material composition on mechanical properties. The resulting mechanical properties, and changes during degradation, must be considered for the design of orthopaedic scaffolds for critical-sized defects.

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Conflicts of Interest: None declared

Ethical Approval: Not required
8 References


9 Figure Captions

Figure 1: Strut ladder and tensile sample assembly.

Figure 2: Average tensile curves of PCL struts (grey, n=19), 90/10wt% (blue, n=16) and 50/50wt% (red, n=7) PCL/β-TCP struts with standard error presented as shaded areas (A). The linear portion of each curve used for the calculation of the material elastic modulus is presented in B. Tensile curves for 50/50wt% struts to maximum stress have previously been published elsewhere [29] and an average curve is presented here for comparison purposes.

Figure 3: Comparison of elastic modulus (A), tensile strength $\sigma_s$ (B), strain at $\sigma_s$ (C) and average cross-sectional area (D) for undegraded PCL, 90/10wt% and 50/50wt% PCL/β-TCP struts. The elastic modulus of 90/10wt% and 50/50wt% struts published elsewhere (in [29] and [35] respectively) have been included in (A) for comparison purposes. *$p > 0.05$.

Figure 4: Average tensile curves for (A) PCL, (B) 90/10wt% struts and (C) 50/50wt% struts before (D0, solid lines) and after seven weeks of degradation (dashed lines). The shaded region in each plot represents the standard deviation of the full set of tensile curves. Tensile data for 50/50wt% struts at D0 have previously been published elsewhere [29] and an average curve is included here for comparison purposes.

Figure 5: Average tensile curves for PCL struts (A, B), 90/10wt% struts (C, D) and 50/50wt% struts (E, F) after 0 days (D0) and one, four and seven weeks degradation. Full curves are shown in A and B; linear portions of the tensile curves are shown in C and D. Tensile data for 50/50wt% struts at D0 have previously been published elsewhere [29] and an average curve is included here for comparison purposes. Standard deviation is not shown in these plots for visual purposes.

Figure 6: Summary of degradation study mechanical testing data for PCL (grey), 90/10wt% (blue) and 50/50wt% struts (red). The change in elastic modulus for each material is shown in A, the change in
tensile strength is shown in B and the change in strain at the tensile strength is shown in C. Elastic modulus data for undegraded 90/10wt% PCL/β-TCP struts and 50/50wt% struts have been published elsewhere (in [35] and [29] respectively) and are included here for comparison purposes. PCL test sample sizes were $n = 4$, 3 and 5, for tests at weeks 1, 4 and 7, respectively. 90/10wt% test sample sizes were $n = 4$, 8 and 9, for tests at weeks 1, 4 and 7, respectively. 50/50wt test sample sizes were $n = 7$, 8, 12 and 10, for tests at weeks 1, 4 and 7, respectively.

**Figure 7**: SEM images of PCL, 90/10wt% and 50/50wt% PCL/β-TCP struts under magnification of x100 with zero degradation (D0) and after seven weeks degradation (Wk7).

**Figure 8**: SEM images of PCL, 90/10wt% and 50/50wt% PCL/β-TCP struts under magnification of x1000 with zero degradation (D0) and after seven weeks degradation (Wk7). Arrows point to β-TCP particles in 90/10wt% and 50/50wt% struts.

**Figure 9**: CT scan images of PCL, 90/10wt% and 50/50wt% PCL/β-TCP struts (left to right) with zero degradation (top) and after seven weeks degradation (bottom). White arrows point to β-TCP particles.

**Figure 10**: (A) Sintering of individual polymer particles to form partially sintered and fully sintered material with time and temperature. (B) Illustration of the effect of increasing β-TCP content on the sintering of PCL/β-TCP composite materials. A lower degree of sintering is achieved with increasing β-TCP content. (C) Schematic of sintering of powder particles with different particle sizes. Regions directly under the laser path are fully sintered, and heat is transferred to surrounding particles through contact points to form partially sintered regions. Full sintering occurs in a greater number of particles when the particle size is less than the laser spot diameter ($\phi_0$) than when particles are larger than $\phi_0$. Greater partial sintering and lower porosity occurs with particle sizes less than $\phi_0$. 


Figure 1:
Figure 2:
Figure 3:
Figure 4:
Figure 5:
Figure 6

(A) Effective Elastic Modulus (MPa)

(B) Tensile Strength σ1 (MPa)

(C) Stain at σ1
Figure 8:
Figure 9:
Figure 10: