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3D PRINTING OF CUSTOMIZED BIOMEDICAL SCAFFOLDS AND IMPLANTS

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ABSTRACT: Powder-bed 3D printing (3DP) is a versatile additive manufacturing technique that uses ink binder deposition to bind loose powder particles together. This paper evaluates the work using powder-bed 3DP technology as a low-cost manufacturing system to print non-proprietary materials for biomedical scaffold and implant applications. The work presents development of suitable binder-powder feedstock coupled with the suitable printing and post-processing methodologies for respective biomaterial.

INTRODUCTION

Additive manufacturing (AM) is an alternative manufacturing process that is capable of producing near net-shape products directly from their 3D computer models without the use of a mold. The ease in design modification is earning this technology favorable notion to fabricate products that require customization or are produced in low volume, such as biomedical implants and scaffolds.

There are still engineering challenges and opportunities in provisions of 3D biomedical scaffolds with porous microstructures mimicking the tissue architecture, fabricated using suitable biomaterials (Griffith & Naughton, 2002). Both AM and non-AM techniques have been attempted to produce such scaffolds. Non-AM methods include gas foaming (Yoon et al. 2001), solvent casting and particulate leaching (Kawanishi et al. 2004, Sato et al. 2004), and freeze-drying (Whang et al. 1995). These methods are typically good to obtain random porosity, as the porosity is process-dependent or resultant from pore forming agents. Consequently, these methods typically produce pores in irregular shape and size, with limited interconnectivity of the porous network.

Porous scaffold and implant fabrication by AM have been reported in various publications (Yang et al. 2002, Yeong et al. 2004, Bártolo et al. 2009, Peltola et al. 2008), reporting scaffolds and implants with regularly structured pores and porosity, in customizable overall shape and size. Powder-based AM systems are considered as the most versatile AM technology, as various materials are directly available in powder form. Compared to other commercially available powder-based AM systems, such as selective laser sintering, selective laser melting and electron beam melting, powder-bed 3DP is regarded as a low-cost system, therefore reducing the initial investment cost of owning a powder-based AM system.

A powder-bed 3DP builds part based on inkjet printing deposition concept. The system consists of a part building chamber, a powder supply chamber, a roller and an inkjet printer depositor. The ink depositor mechanism selectively drops the ink binder onto the loose powder in the part-building chamber, corresponding to that particular layer's cross-sectional area. This action produces a layer of bonded powder material at the selected regions.

Parts printed by powder-bed 3DP are naturally porous. This is mainly due to the 3DP inkjet-printing manner, whereby the powder particles are bound together by the ink binder with minimal heating at temperature much lower than the powder's melting or glass-transition temperature. Therefore the bound powder particles still maintain their original size and shape, with highly porous printed parts produced. Combining the design flexibility that is a common feature in AM technologies, 3DP products can have dual porosity that is a combination of the inherent and pre-designed porosity (Maleksaeedi et al. 2013).

This paper elaborates the studies conducted using powder-bed 3DP to print non-proprietary biomaterials, namely poly(lactide-co-glycolide and polyglycolide biopolymers and titanium. The studies encompass analysis of selecting suitable ink binder material and development of suitable printing process. Exploration of post-processing methodologies, such as thermal debinding, sintering and infiltration, were conducted for the respective biomaterials to enhance the scaffold mechanical performance.

3D PRINTING OF BIOPOLYMER SCAFFOLDS

50/50 poly(DL-lactide-co-glycolide) (PLGA) and polyglycolide (PGA) (Purasorb PDLG, Purac Asia Pacific Pte Ltd) pellets were pulverized in cryogenic environment (SPEX CertiPrep 6850 freezer/mill) to obtain particles size less than 212 μm . Binder materials evaluated were poly(ethylene glycol) (PEG) (30,902-B, Sigma-Aldrich), poly(ethylene oxide) (PEO) (18,945-6, Sigma-Aldrich) and poly(vinyl alcohol) (PVA) (GL-05S, Nippon Gohsei). Ink binders evaluated were water, ethanol and acetone.

The compatibility between the polymer materials (PLGA, PGA and various binders) and ink binder was established by dissolving the polymer in the respective ink binders. The solubility of each material is provided in Table 1. Although ethanol was not able to dissolve the PLGA, it was included in the ink formulation to dilute the corrosive effect of acetone on the printer components.

Test printing (ZCorp Z402 system) was conducted to examine the solvent-binder-polymer compatibility by overprinting the same area of polymer/binder mixture with acetone/ethanol/water

ink binder. The printing experiments showed that polymer/PEG resulted in weak bonding, whereas polymer/PEO and polymer/PVA mixtures gave good bonding.

Table 1. Solubility of polymer materials and ink binder.

Polymer	Solubility in ink binder		
	Water	Ethanol	Acetone
PGA	No	No	No
PLGA	No	No	Yes
PEG	Yes	No	No
PEO	Yes	No	No
PVA	Yes	No	No

PVA material was finally selected as the binder material with the consideration that PVA was less costly than PEO and was commercially available in powder form. PEO was commercially available in flake form and thus required further pulverization process before using in the printer.

Degradation study was conducted to the biopolymer scaffolds to understand the time required for the scaffolds to fully degrade. Degradation rate also affect cellular processes, including cell growth, tissue regeneration and host response (Babensee et al. 1998). Both PGA and PLGA degrade by hydrolysis of the ester bonds into lactic and glycolic acid, which are then removed from the body through normal metabolic cycle (Göpperich, 1996). The biopolymer scaffolds were placed in phosphate-buffered saline (pH 7.4) prepared in-house, and kept in controlled temperature bath at 37°C up to 6 weeks.

Molecular weight, M_w , of PLGA in the scaffold decreased exponentially with degradation time. At week 2, M_w had decreased to 517,000 from 106,000 before degradation study (Figure 1(a)). The PLGA in the scaffold had degraded as expected. The scaffold strength reduced by 50% after week 1 and this level of strength was maintained through week 2 (Figure 1(b)). Another 50% reduction in strength appeared at week 3. Scaffolds on week 4 onwards were too brittle to be measured.

Figure 1. (a) Molecular weight and (b) strength of dried scaffold after degradation study.

Reasonably good shape retention was observed up to week 4, after which there was distortion in shape (Figure 2).

Figure 2. Physical appearance at: (a) week 1, (b) week 3, (c) week 4, (d) week 6, in PBS

3D PRINTING OF TITANIUM POROUS STRUCTURES

Stress shielding is a perennial problem with orthopedic implants as the modulus of implant materials such as titanium alloy, cobalt-chromium or stainless steel are much higher compared to bone modulus causing remodeling and loss of bone mass closer to the implant (Engh et al. 1987). This causes implant loosening in the long run necessitating revision surgeries to be performed. Therefore, it is desirable to have the implant's modulus similar to do the bone. A method to solve this issue is to introduce porosity to the implant's structure and thus reducing its modulus (Wiria et al. 2010).

Titanium is selected as a material of choice due to its biocompatibility and good corrosion resistance, making it being widely used in its bulk form for implant such as prosthetic joints and dental fixtures. Most importantly, bulk titanium's modulus of 102-105 GPa (Lütjering & Williams, 2007) is the lowest compared to other commonly used metallic implant materials such as stainless steel or cobalt-chromium alloy (Ramakrishna et al. 2001).

Commercially pure (CP) titanium (Grade 2, TLS Technik Spezialpulver) was mixed with PVA (NH-18S, Nippon Gohsei) binder in 5 to 30 wt.% binder concentration and was printed (ZCorp, 310Plus system) using water as ink binder. Post-processing consist of thermal debinding and further sintering of titanium green part (CM tube high temperature furnace) at sintering temperature from 900 to 1350°C.

The polymer binder functions to bind the raw material mixture during printing process. Presumably higher amount of binder would provide printed parts with better green strength for ease of handling. There should be a balance in binder concentration in the powder mixture. Higher concentration of binder causes several effects. First, the amount of porosity is expected to increase due to the voids left behind when the binder is removed. Secondly, excessive binder powder may restrict titanium particles to be in close contact with each other after the debinding cycle, hence higher temperature or longer sintering time would be required to achieve high densification.

The effect of binder concentration to the porosity and density of the final sintered titanium structure, when sintered at the same sintering temperature, are provided in Figure 3(a). Porosity was measured using pycnometer (AccuPyc II 1340, Micromeritics) and density was measured according to Archimedes principle (British Standard BS EN 623-2:1993). It was as expected that the porosity would increase and density would consequently decreases as the binder concentration

increases. However, it can be seen that the increase in porosity is not too significant between the lowest and highest binder concentration used. Typical microstructure of porous printed titanium is provided in Figure 3(b).

Figure 3. (a) Apparent density and porosity of printed titanium (sintered at 1200°C) with various binder concentration, (b) Typical microstructure of porous printed titanium (insert: printed titanium after sintering)

As the increase in porosity was not very significant with the increase in the binder concentration, further compression testing was conducted on samples built using 30 wt.% binder and sintered using various sintering temperature. Table 2 shows the elastic modulus of the printed titanium. The values of the elastic modulus fall within the range of modulus of natural bone, 3 to 20 GPa (Krishna et al. 2007).

Table 2. Elastic modulus of printed titanium, sintered at various temperature.

Sintering temperature (°C)	Elastic modulus (GPa)
1250	4.8
1300	10.8
1350	13.2

These results indicate that it is feasible to modify the porosity and mechanical property of porous titanium structures by engineering the raw material concentration and post-processing methodology.

CONCLUSIONS

These works have demonstrated the capability of commercial 3D printer system to be used to print various non-proprietary materials, namely PGA/PLGA biopolymer and titanium, by developing suitable material mixture and ink binder. The printed parts, coupled with suitable post-processing methodologies, have enabled the fabrication of porous scaffolds and implants that are favorable to be used in biomedical applications. Engineering the post-processing methodology of printed titanium has enabled the formation of porous parts with elastic modulus mimicking to that of natural bone.

REFERENCES

- Babensee, J.E., Anderson, J.M., McIntire, L.V., Mikos, A.G. (1998). "Host response to tissue engineered devices", *Advanced Drug Delivery Reviews*, 33(1-2), 111-139.
- Bártolo, P.J., Chua, C.K., Almeida, H.A., Chou, S.M. & Lim, A.S.C. (2009). "Biomanufacturing for tissue engineering: Present and future trends", *Virtual and Physical Prototyping*, 4(4), 203-216.
- British Standard BS EN 623-2:1993. "Advanced technical ceramics – Monolithic ceramics – General and textural properties – Part 2: Determination of density and porosity"
- Engh, C.A., Bobyn, J.D., Glassman, A.H. (1987). "Porous-coated hip replacement – The factors governing bone ingrowth, stress shielding, and clinical results", *The Journal of Bone and Joint Surgery*, 69B(1), 45-55.
- Göpferich, A. (1996). "Mechanisms of polymer degradation and erosion", *Biomaterials*, 17(2), 103-114.
- Griffith, L.G. & Naughton, G. (2002). "Tissue engineering – Current challenges and expanding opportunities", *Science*, 295(5557), 1009-1014.
- Kawanishi, M., Ushida, T., Kaneko, T., Niwa, H., Fukubayashi, T., Nakamura, K., Oda, H., Tanaka, S. and Tateishi, T. (2004). "New type of biodegradable porous scaffolds for tissue-engineered articular cartilage", *Materials Science and Engineering C*, 24, 431-435.
- Krishna, B.V., Bose, S., Bandyopadhyay, A. (2007). "Low stiffness porous Ti structures for load-bearing implants", *Acta Biomaterialia*, 3(6), 997-1006.
- Lütjering, G. & Williams, J.C. (2007). *Titanium*, Springer-Verlag.
- Maleksaeedi, S., Wang, J.K., El-Hajj, A., Harb, L., Guneta, V., He, Z., Wiria, F.E., Choong, C., Ruys, A.J. (2013). "Toward 3D printed bioactive titanium scaffolds with bimodal pore size distribution for bone ingrowth", *Procedia CIRP*, 5, 158-163.
- Peltola, S.M., Melchels, F.P.W., Grijpma, D.W., Kellomäki, M. (2008). "A review of rapid prototyping techniques for tissue engineering purposes", *Annals of Medicine*, 40(4), 268-280.
- Ramakrishna, S., Mayer, J., Wintermantel, E., Leong, K.W. (2001). "Biomedical applications of polymer-composite materials: a review", *Composites Science and Technology*, 61(9), 1189-1224.
- Sato, T., Chen, G., Ushida, T., Ishii, T., Ochiai, N., Tateishi, T. and Tanaka, J. (2004). "Evaluation of PLLA-collagen hybrid sponge as a scaffold for cartilage tissue engineering", *Materials Science and Engineering C*, 24, 365-372.
- Whang, K., Thomas, C.H., Healy, K.E. and Nuber, G. (1995). "A novel method to fabricate bioabsorbable scaffolds", *Polymer*, 36, 837-842.
- Wiria, F.E., Yong, J.M.S., Lim, P.N., Goh, F.C.W., Yeo, J.F., Cao, T. (2010). "Printing of titanium implant prototype", *Materials and Design*, 31(Supplement 1), S101-S105.
- Yang, S.F., Leong, K.F., Du, Z.H. and Chua, C.K. (2002). "The design of scaffolds for use in tissue engineering. Part II. Rapid prototyping techniques", *Tissue Engineering*, 8(1), 1-11.
- Yeong, W.Y., Chua, C.K., Leong, K.F. and Chandrasekaran, M. (2004). "Rapid prototyping in tissue engineering: Challenges and potential", *Trends in Biotechnology*, 22(12), 643-652.
- Yoon, J.J. and Park, T.G. (2001). "Degradation behaviors of biodegradable macroporous scaffolds prepared by gas foaming of effervescent salts", *Journal of Biomedical Materials Research*, 55, 401-408.