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Mechanical degradation model of biodegradable scaffolds: A computational approach

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ABSTRACT

Biomaterial scaffolds are material used as a temporary supporting structure from bone growth. It should retain its strength during the recovery process whilst providing suitable condition for new cells to grow. Metallic base materials such as titanium, ferum and magnesium alloys are among potential candidates for this purpose. They are not only inert for human bodies but also can be tailored to be biodegradable according to the desired implantation period. Scaffolds surface from metal alloys also can be treated so that cell growth can be enhanced for a rapid recovery. A rapid development of several metallic biomaterial scaffolds requires relatively fast mechanical characterisation of the structure. *In vivo* approach demand a lengthy time in order to prepare samples for characterisation. This impediment can be overcomed by having a virtual model of the scaffolds and general properties of the structure can be predicted by numerical simulations. In present work, titania biomaterials are presented as a case study to show this capability. The nature of surface erosion on these scaffolds can be precisely simulated by means of calculating their mass at systematic intervals.

Keywords: Tissue engineering, Computational Method, Degradation

INTRODUCTION

Biomaterial scaffolds generally made of material that does not cause irriation to human bodies. These materials may be chosen from metallic allloys or polymer-based materials. They are preferred to have degradation ability are now used for bone-regenerative materials. This is a new concept of tissue engineering to replace the traditional practice of synthetic implants introduced in the early 1990's which has limitations from tissue grafting (Xie et al., 2010). These bio-compatible materials consist of two main parts, degradable polymers and scaffolds. Biodegradable polymers can be grouped into two: 1) natural-based poly-saccharides materials (i.e. starch, chitosan) and 2) proteins (i.e. soy, collagen, silk) (Rezwan, Chen, Blaker, & Boccaccini, 2006). Tissue engineering scaffolds can be made of bioceramics, bioactive glass, alumina (TiO_2) and calcium phosphates (Bretcanu, Verné, Borello, & Boccaccini, 2004). Bioactive glass and ceramic scaffolds are known to have a low compression load resistance and brittleness. Bioactive materials with bioinert particles such as TiO_2 and Al_2O_3 show a lack of bone bonding when the particulate form is in bulk or micrometer-size. The investigation described in this paper is to show the ability of numerical calculations to predict mechanical response of biomaterials even before an actual *in vivo* setup is ready. This can serve researchers and scientist to have some idea on how their samples will behave under a long immersion in simulated body solutions. Reconstructed models of biomaterial scaffolds under a systematic period of immersion is shown in Figure 1.

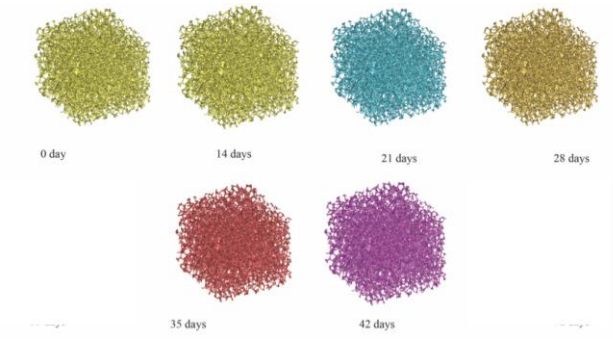

Fig. 1 Scaffolds virtually eroded in simulated body fluid.

MATERIALS AND METHOD

Materials

A rapid research interest is observed with the focus of developing cell-stimulative biomaterials, especially in the last two decades (Jones & Boccaccini, 2006; Rezwan et al., 2006). Recent biodegradable

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porous scaffolds not only act as a template for cell adhesion, but they also used to systematically deliver drugs for bone regrowth. Decomposition of biodegradable materials has been addressed by both *in vitro* (in lab with appropriate liquid solutions) and *in vivo* (engineering tissue scaffolds are directly implanted in living organism). Titania scaffolds (TiO_2) is used in the present study and virtual models are developed by means of micro-CT (μCT) approach. Surface erosion is done by Boolean plugin embedded in commercially available in μCT image processor, Materialise Mimics.

Vicker's Hardness

The above simulations can be divided into six stages of immersion in SBF: 1) 0 day, 2) 14 days, 3) 21 days, 4) 28 days, 5) 35 days and 6) 42 days. By using Vickers hardness and yield strength relationship (Tekkaya, 2000), the yield stress of decomposed titania samples can be determined. The relationship between Vickers hardness and yield stress is defined as, $HVN = 3.0\ Y$. *HVN* stands for Vickers hardness number, and *Y* is the yield stress of the material. The constant value of 3.0 for this relationship is justified by the usage of non-strain-hardening materials used in this work. The approximated yields stress values for porous sintered titania is summarized in the following Table 1.

Table 1: Yield stress of sintered pure titania scaffold before and after immersion in SBF solutions.

Sample	Immersion time (days)	Yield stress (MPa)	Vickers hardness (kgf/mm²)
Pure titania	0	134.7	449
	14	120.9	403
	21	106.5	355
	28	88.5	295
	35	70.5	235
	42	59.7	199

RESULTS AND DISCUSSION

Scaffold models but with different porosities is created to simulate imaginary degradation of titania scaffolds. Models with different porosities also represent a mass loss experienced by biomaterial scaffolds during 6 weeks immersion in SBF.

Sintered pure titania scaffold has 4.05% of mass reduction after 42 days in SBF liquids according to the work reported in (Menon, 2009). It is interesting to note that only sintered pure titania scaffold experiencing the mass loss from decomposition in SBF solutions (see Fig. 2). The phase-porous titania scaffolds increase in weight due to apatite layer formation on their surface after 6 weeks of immersion. The presence of apatite (see Fig.3) is confirmed by X-ray diffraction analysis (XRD). Apatite substance is indicated as spikes denoted as HAP in the figure 3. In this numerical analysis, apatite layer formation is ignored. The main reason for this simplification is already suggested in Fig.2, where only pure titania sample decreased in weight after immersion in SBF for 6 weeks. Several investigations suggest the likely explanation for the weak bioactivity of sintered pure titania is due to amorphous surface structure possessed by this biomaterial scaffold as can be seen in (Tadashi Kokubo, Miyaji, Kim, & Nakamura, 1996; Nagano, Nakamura, Kokubo, Tanahashi, & Ogawa, 1996; Uchida, Kim, Kokubo, Fujibayashi, & Nakamura, 2003). However, the bioactivity of titanium and its alloys can be improved by performing a chemical pre-treatment using (i.e.: alkali hydroxide solutions) on their surfaces (Tadashi Kokubo et al., 1996).

Simplifications of numerical model by omitting the apatite layer formation also supported by the relatively weak adhesive bonding onto the titania scaffold matrix. Adhesive strength of apatite layer on biomaterial matrix can be measured from detaching tests. Sample is ready to be harvested after 6-8 weeks of immersion in SBF liquids. Sample is prepared so that two cortices (outer layer of tissues) are remained for detaching test (T Kokubo, 1996). Adhesive strength of apatite to chemically treated titanium metals is reported in the range of 9.8 to 11.5 MPa (T Kokubo, 1996). The adhesive strength of apatite on polymer matrix (i.e.: increasing order of adhesive strength poly(methyl methacrylate) (PMMA), polyethylene terephthalate (PET), polyethylene (PE) and poly-ether sulphone (PESF)) is even lower with the range of 1.06 MPa to 4.4 MPa (T Kokubo, 1996). This is considered relatively low compared to the bonding of polymeric adhesives found in acrylic and epoxy which ranges from 65.2 MPa to 78.4 MPa (You, Yan, Zheng, Zhu, & Hu, 2010). Thus, the effect of the apatite adhesion on titania scaffold can be assumed negligible justified by the relatively low adhesive strength of apatite layer formation onto biomaterial matrix.

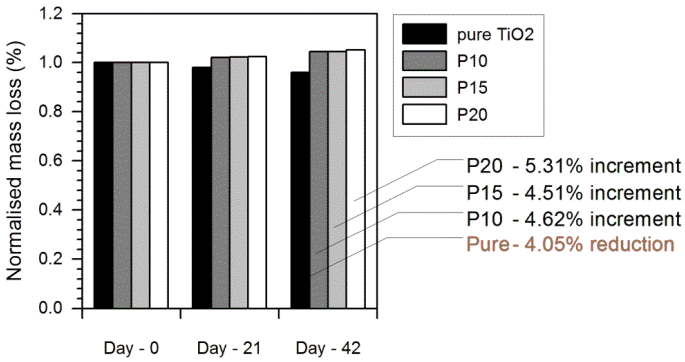

Fig. 2: Normalised mass loss of pure and porous titania scaffolds (Note: P10, 15 and 20 indicates the different wt% of pore former polyethylene glycol (PEG) added to the titania powder)

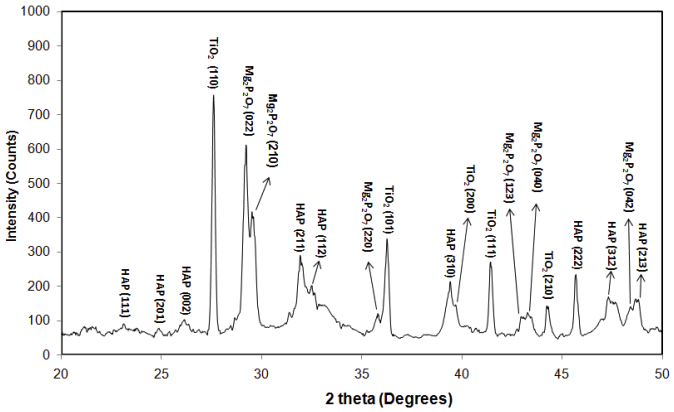

Fig.3: XRD pattern of porous scaffold placed in SBF for 42 days

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Adhesive strength of apatite layer on biomaterial matrix can be measured from detaching tests. Sample is ready to be harvested after 6-8 weeks of immersion in SBF liquids. Sample is prepared so that two cortices (outer layer of tissues) are remained for detaching test (T Kokubo, 1996). Adhesive strength of apatite to chemically treated titanium metals is reported in the range of 9.8 to 11.5 MPa (T Kokubo, 1996).. The adhesive strength of apatite on polymer matrix (i.e.: increasing order of adhesive strength poly(methyl methacrylate) (PMMA), polyethylene terephthalate (PET), polyethylene (PE) and poly-ether sulphone (PESF)) is even lower with the range of 1.06 MPa to 4.4 MPa (T Kokubo, 1996).. This is considered relatively low compared to the bonding of polymeric adhesives found in acrylic and epoxy which ranges from 65.2 MPa to 78.4 MPa (You et al., 2010). Thus, the effect of the apatite adhesion on titania scaffold can be assumed negligible justified by the relatively low adhesive strength of apatite layer formation onto biomaterial matrix.

Table 2 Yield stress of sintered titania scaffold before and after immersion in SBF solutions.

Sample	Immersion time (days)	Yield stress (MPa)	Vickers hardness (kgf/mm²)
pure	0	134.4	448
P10		66.9	223
P15		60.6	202
P20		54.6	182
pure	21	106.5	355
P10		59.4	198
P15		53.4	178
P20		48.0	160
pure	42	60.0	200
P10		51.9	173
P15		46.2	154
P20		41.4	138

CONCLUSION

Numerical models of a biomaterial scaffolds are virtually augmented in the present investigation. These models are developed with extremely high accuracy due to micro-computed tomography's ability to scan up to ~17 microns of precision. Surface erosion on actual *in vivo* samples can then be emulated by tagging a sample's mass during the degradation process. Porosity of eroded scaffolds also can be used as an input data to build the virtual models. Hardness value of a certain material can be used to project their yield strength to be used in finite element simulations.

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