

Invited Contribution: 2007 AIChE Alpha Chi Sigma Award

From Material to Tissue: Biomaterial Development, Scaffold Fabrication, and Tissue Engineering

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The need for techniques to facilitate the regeneration of failing or destroyed tissues remains great with the aging of the worldwide population and the continued incidence of trauma and diseases such as cancer. A 16-year history in biomaterial scaffold development and tissue engineering is examined, beginning with the synthesis of novel materials and fabrication of 3D porous scaffolds. Exploring cell-scaffold interactions and subsequently cellular delivery using biomaterial carriers, we have developed a variety of techniques for bone and cartilage engineering. In addition to delivering cells, we have utilized growth factors, DNA, and peptides to improve the in vitro and in vivo regeneration of tissues. This review covers important developments and discoveries within our laboratory, and the increasing breadth in the scope of our work within the expanding field of tissue engineering is presented. © 2008 American Institute of Chemical Engineers AIChE J, 54: 3048–3067, 2008

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Introduction

The field of tissue engineering, loosely defined as the use of engineering principles and technologies to regenerate living tissues, has roots in many branches of science and engineering. Following its introduction to the broad scientific

community over 15 years ago,¹ the field has rapidly expanded and advanced, as evidenced by publication trends.² Founded at the intersection of chemistry, materials science, systems level and molecular biology, and chemical and mechanical engineering,^{3,4} the viability of technologies and products emerging from tissue engineering is apparent,^{5–11} and with continued expansion, tissue engineering-based therapeutics will play a large role in advancing medicine in the 21st century.

Most iterations of the traditional tissue engineering paradigm describe the combination of cells, scaffold materials, and bioactive factors towards the de novo growth or induced regeneration of living tissues following damage or in conditions under which regeneration would not normally

This invited contribution presents a 16-year perspective of the research conducted in the laboratory of Prof. Antonios G. Mikos of Rice University in the biomaterials and tissue engineering fields. Prof. Mikos was the recipient of the 2007 Alpha Chi Sigma Award for Chemical Engineering Research of the American Institute of Chemical Engineers.

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occur. For the last 16 years, our laboratory has investigated mainly orthopaedic and dental tissue engineering, focusing primarily on the regeneration of bone and cartilage. In doing so we have formulated new tissue engineering techniques, investigated key parameters for tissue growth within synthetic matrices, and developed novel biomaterials for use as tissue engineering scaffolds and bioactive factor delivery vehicles.

In 2005, over 2,300,000 procedures were performed in U.S. hospitals involving the partial excision of bone, treatment of fractures, or joint replacement.¹² Many of these procedures were likely necessitated by or will result in a bony defect that will not regenerate. Most commonly due to trauma or neoplasm, these nonhealing or nonunion bone defects are costly and can adversely affect patient quality of life. Bone tissue engineering is a potential source of treatments for these defects. If successfully implemented, bone tissue engineering strategies will allow for the complete functional and morphological regeneration of healthy bone tissue without the need for residual or permanently indwelling synthetic materials or large amounts of donor tissue, the procurement of which typically involves either a risk of transmitted disease from allo- or xenogeneic tissues^{13,14} or the necessity for additional surgeries¹⁵ and potential morbidity at the donor site for autologous tissues.^{16,17}

Regenerating bone tissue *in vivo* requires the consideration of a number of critical elements. First, bone regenerates or heals preferentially when under mechanical stimulation,^{18–20} possibly due to the differentiation of stem cells in response to their mechanical microenvironment.²¹ Thus, in addition to providing a three-dimensional template for tissue growth, a material used as a scaffold must be able to withstand the mechanical loading necessary to facilitate bone growth. Second, diffusional limitations on the delivery of oxygen and nutrients from the blood stream and the removal of waste products affect the size of defects that can be addressed by tissue engineering.^{22,23} Appropriate material porosity and the allowance or induction of vascular ingrowth can mitigate these limitations.^{24–27} Finally, for the regenerated bone to be identical to natural bone, the scaffold material must degrade *in vivo* but must do so at a rate so as not to compromise the mechanical stability of the scaffold prior to sufficient bony ingrowth. Along with these key elements, cyto- and biocompatibility must obviously be addressed.

The requirements for engineering other tissue types are similarly specific, and thus as the field of tissue engineering progresses, it is unlikely that a single material will be capable of meeting the criteria necessary for successful application towards engineering many tissues. There is a distinct need for biomaterials and combinations of biomaterials, processing techniques, bioactive factors, and cells tailored for tissue specific applications.²⁸ Early work in tissue engineering and within our laboratory focused predominantly on applications using the now FDA-regulated material poly(D,L-lactic-co-glycolic acid) (PLGA); however, as our laboratory and the field in general have evolved, more experimental work is being performed on novel biomaterials with parameters rendering them appropriate for specific applications.

Biomaterial and Scaffold Technology Development

Early work

Building upon prior work investigating preparation of polymer scaffolds,^{29–32} at its inception our laboratory began studying the interaction between osteoblasts and polymer matrices in collaboration with Dr. Rena Bizios, now of The University of Texas at San Antonio, first investigating the effect of varying the comonomer ratio in PLGA films on the expression of alkaline phosphatase (ALP) and collagen synthesis.³³ Similar work investigating polymer composition and cell-polymer interactions was performed using poly(L-lactic acid) (PLLA) and PLGA films to culture human retinal pigment epithelial (RPE) cells.^{34,35} The release of lactic acid from degrading PLLA membranes was modulated by varying the polydispersity of PLLA blends made from monodisperse PLLA of high and low molecular weights, allowing for controlled release of acidic degradation products to minimize fluctuations in pH.³⁶ Biodegradable polymer particles fabricated from both PLLA and PLGA on the order of 1 μm in diameter were found to inhibit cell proliferation and matrix mineralization at high particle concentrations, possibly due to diffusional limitations imposed by the particles and cell culture; however, established cell cultures were less effected than cultures exposed to particles on the first culture day.³⁷

Investigating the effect of pore size on fibrovascular tissue ingrowth, we found that varying the pore size in a polymer scaffold made of PLLA could drastically affect the rate at which surrounding tissues invaded the matrix²⁵ and later the rate at which materials degrade both *in vitro* and *in vivo*.^{38,39} Although rapid tissue ingrowth and vascularization can modulate diffusional limitations within large scaffolds, the presence of this granulation tissue within the pores of the scaffold limits space available for guided tissue regeneration and was thus recognized as being potentially detrimental to success in some applications.

Noting the role pore size, morphology, and interconnectivity would no doubt have in abrogating diffusional limitations and facilitating tissue regeneration within defects of clinically relevant size, we developed a novel method of particulate leaching to create PLGA foams around degradable gelatin microspheres.⁴⁰ Similar to other work involving fiber bonding²⁹ and phase separation,⁴¹ this technique allowed control over pore sizes and overall porosity but did so through a process involving no additional organic solvents, making it particularly attractive for biomedical applications. This particulate leaching method was later employed using salt crystals as a porogen, and simple leaching of the crystals resulted in PLGA/poly(ethylene glycol) (PEG) blended scaffolds with varied shear moduli, porosities, and pore diameters.⁴² Solvent casting, salt leaching, and extrusion were used to create porous PLLA and PLGA conduits to investigate peripheral nerve regeneration.^{43–45} Continued work with our collaborators has applied the particulate leaching technique for bone tissue engineering with calcium phosphate (CaP) biomaterials.^{46–48} In addition to expansion into other areas of scaffold fabrication and tissue engineering, our laboratory and collaborators have continued to investigate processing techniques for creating scaffold porosity (Figure 1).^{49–51}

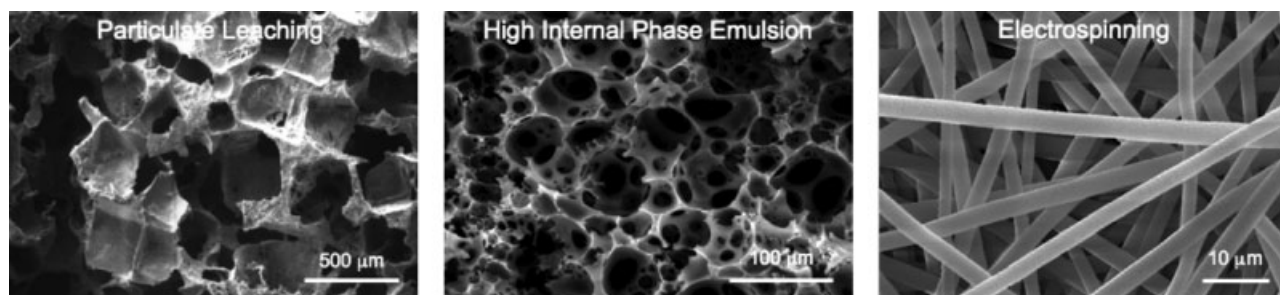


Figure 1. Examples of porous scaffolds.

Techniques employing particulate leaching (left), high internal phase emulsion (middle), and electrospinning (right) have all been used to fabricate porous biodegradable polymer matrices for use as tissue engineering scaffolds.

Poly(propylene fumarate)

In 1995, we began developing novel biodegradable scaffolds based on poly(propylene fumarate) (Figure 2), an unsaturated linear polyester based on fumaric acid, a nontoxic intermediate in the citric acid cycle.⁵² Early work focused on characterizing PPF both alone⁵³ and as a composite with the ceramic β -tricalcium phosphate (β TCP)⁵⁴ and later with PLGA.^{55–57} Subsequent refinements in the synthesis of PPF were made,⁵⁸ resulting in a step polymerization of diethyl fumarate and propylene glycol in the presence of zinc chloride (Figure 3).⁵⁹ This one pot method yielded PPF with molecular weights up to 4600 after 12 h.

In addition to optimizing the synthesis of PPF, a series of studies investigating the development and characterization of cross-linked PPF scaffolds were performed. Using benzoyl peroxide (BP) as an initiator, cross-linked PPF networks were synthesized using PEG-dimethacrylate (PEG-DMA)⁶⁰ or propylene fumarate-diacrylate (PF-DA).⁶¹ Using PPF/PF-DA networks as a model system, later work characterized the

participation of acrylate and fumarate groups in network formation⁶² and found continual reactivity of unreacted fumarate groups at physiological temperatures, resulting in an increased compressive modulus after 6-weeks incubation.⁶³ This unique process can lead to mechanical reinforcement of scaffolds with concurrent degradation, providing a possible mechanism for shape retention and maintained mechanical support with time.

We also investigated photocross-linking using bis(2,4,6-trimethylbenzoyl) phenylphosphine oxide (BAPO) to cross-link PPF networks both with⁶⁴ and without^{65,66} the addition of porogen. Porous and nonporous formulations of photocross-linked PPF maintained their structure, strength, and porosity as the material degraded, even after 32 weeks and as mass losses approached 30%.⁶⁷ A comparison of thermal- and photocross-link of PPF networks found photocross-linking using BAPO yields a higher cross-linking density and higher

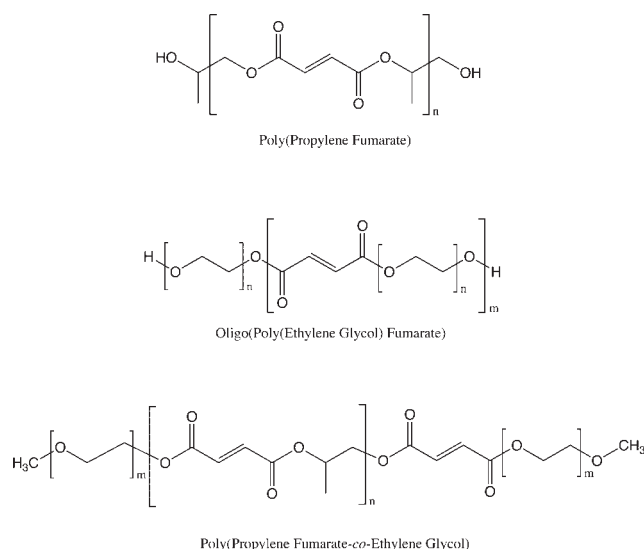


Figure 2. Structure of fumarate-based polymers, poly(propylene fumarate) (PPF), oligo(poly(ethylene glycol) fumarate) (OPF), and poly(propylene fumarate-co-ethylene glycol) (P(PF-co-EG)).

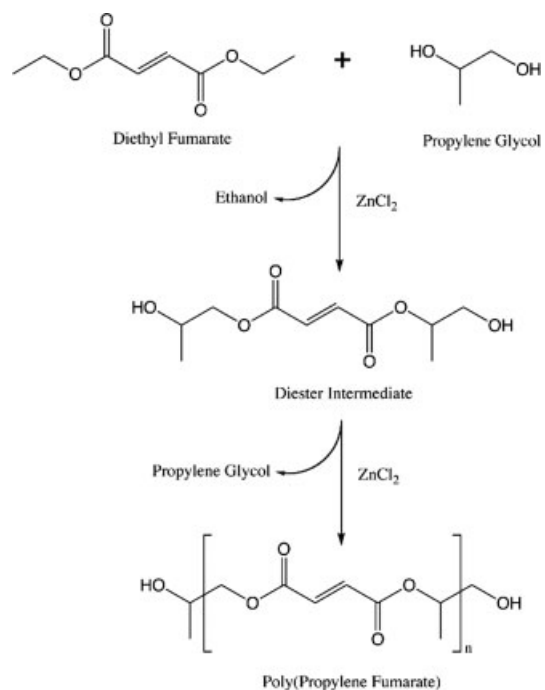


Figure 3. Reaction schema showing the synthesis of poly(propylene fumarate) from diethyl fumarate and propylene glycol.

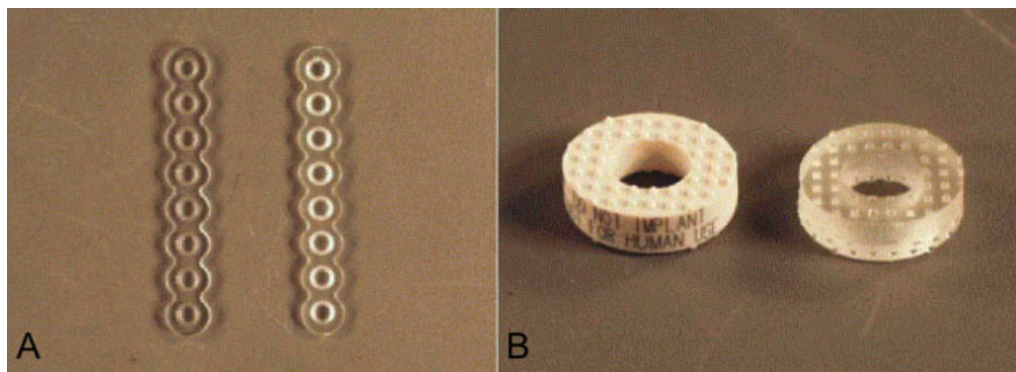


Figure 4. Biodegradable orthopaedic implants (A) 1.5 mm 8 hole adaption plates manufactured with 70:30 P(L/DL-LA) (left) and PPF/PF-DA with a double bond ratio of 0.5 (right).

The PPF/PF-DA plate was fabricated with a transparent silicone mold formed with a P(L/DL-LA) master. (B) Plastic model (left) and PPF/PF-DA with double bond ratio 0.5 replicate (right) of a 5-mm lordotic anterior cervical fusion spacer. The plastic model has identical geometry as the bone allograft implant and was used to produce the silicone molds for the PPF/PF-DA device.⁷¹ Reproduced from Ref. 71, with permission from Elsevier. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

double bond conversion than thermal cross-linking in the presence of BP;⁶⁸ however, both methods offer utility in different applications. Networks that undergo thermal cross-linking close to physiological temperature hold potential as in situ cross-linkable materials for injectable applications,^{69,70} while photocross-linking PPF/PF-DA networks within silicon molds⁷¹ or PPF/diethyl fumarate composites during stereolithography⁷² was successfully used to fabricate biodegradable orthopaedic implants (Figure 4). Using a rabbit model, photocross-linked PPF implants were also found to elicit only a mild inflammatory response 2 weeks after implantation in both soft and hard tissues, and this inflammatory response was largely resolved with surface degradation evident by 8-weeks post-implantation.⁷³

Other fumarate-based materials

While developing PPF, we also investigated other fumarate-based materials. Poly(propylene fumarate-co-ethylene glycol) (P(PF-co-EG), Figure 2), a block copolymer hydrogel of PPF and PEG, was synthesized and found to have tunable mechanical properties controlled by varying the PPF molecular weight and PEG content.⁷⁴ In vitro and in vivo degradation studies of P(PF-co-EG) hydrogels, performed in collaboration with Dr. James M. Anderson of Case Western Reserve University, determined that increasing the weight percent of PEG decreased material degradation rates, while changes in PEG molecular weight had only minimal effects on degradation.⁷⁵ Similarly, the biocompatibility of the hydrogels increased with increasing weight percent of PEG,⁷⁶ and both PEG weight percent and molecular weight influenced platelet attachment at the material surface.⁷⁷ The ability to tune the mechanical, degradative, and biointeractive characteristics of this injectable material make it an attractive substrate for engineering a variety of tissues. Later work using this copolymer included the introduction of a novel method for creating in situ cross-linked macroporous hydrogels using generated carbon dioxide as a porogen.^{78,79} Substitution of methoxy poly(ethylene glycol) for PEG yielded biodegradable copolymers that undergo both physical and chemical gelation,⁸⁰ a concept that has continued to be investigated in our laboratory.⁸¹

In addition to P(PF-co-EG), hydrogels based on oligo(poly(ethylene glycol) fumarate) (OPF, Figure 2), a novel oligomer developed by our laboratory,⁸² were investigated as tissue engineering and drug delivery substrates. A PEG-based macromer with unsaturated double bonds along its macromolecular chain, OPF synthesized with PEG of different molecular weights allows for modulation of tensile mechanical properties, swelling characteristics, mesh size, and cell attachment.^{83,84} Cross-linking of OPF hydrogels using redox radical initiators such as ammonium persulfate (APS) with a reducing agent was found to be a feasible method for fabricating cross-linked hydrogel networks in the presence of mesenchymal stem cells (MSCs);⁸⁵ a subpopulation of bone marrow stromal cells or adherent cells found within the bone marrow space, these adult progenitor cells hold great promise for tissue engineering and other medical applications. OPF would later be used for a number of tissue engineering and drug delivery applications, most notably in the area of cartilage regeneration.

3D Cell-Scaffold Constructs

Beyond material characterization such as cytotoxicity evaluation, the development of materials for tissue engineering applications requires application specific investigation of cellular function on or within the material being developed. As we sought to develop materials for bone tissue engineering, the function and interaction of osteoblasts and osteoblast progenitors with the biomaterial scaffold was critical to the success of the material. Our early work found that osteoblasts proliferated and deposited a mineralized matrix on PLGA foams both in vitro^{86,87} and in vivo,⁸⁸ even when implanted into an ectopic anatomical site (one in which bone growth does not naturally occur). Although this early work found no correlation between scaffold pore size and mineralized matrix deposition, subsequent work relating matrix deposition and cell culture period to pore morphology found that highly porous scaffolds seeded with osteoblasts tended to collapse after only 1 week in culture.⁸⁹ This work was important in shaping future studies, as a balance is needed between highly porous scaffolds that allow rapid tissue ingrowth and minimize diffu-

sional limitations and less porous materials that retain both construct shape and the ability to bear mechanical loads in a complex biochemical and mechanical environment.

In continuing to investigate interactions between cells and scaffolds, a study of the attachment and proliferation of bone marrow derived osteoblasts on end-capped and non-end-capped poly(D,L-lactic acid) (PLA) and a diblock copolymer of PEG-monomethyl ether and PLA found that cells on PLA adhered and proliferated equally well, while cells on the copolymer did not proliferate but were more highly differentiated and more metabolically active than cells on PLA alone.⁹⁰ This was attributed to decreased protein adsorption on the copolymer and provided an early example of the influence of material properties on cell differentiation and activity. Comparing OPF and PPF within a rabbit tooth socket defect, we found that implantation of the hydrophilic OPF constructs inhibited bone healing as compared to PPF and control groups.⁹¹ Immunohistochemistry during the wound healing process found that the presence of OPF relative to PPF blunted the response of fibroblast growth factor-2 (FGF-2), an important factor in bone and wound healing, indicating that the interaction of biomaterials and growth factor cascades may be critical to wound healing and tissue regeneration.

In addition to OPF and PPF, P(PF-co-EG) was studied as a carrier for endothelial cells. In situ cross-linking of P(PF-co-EG) did not blunt the wound healing response in vivo and in vitro cross-linking with encapsulated endothelial cells confirmed the viability of the copolymer as an injectable cell carrier.⁹²

3D Composite Scaffolds

Building upon the knowledge that bulk material properties and surface characteristics are critical in biological applications, our laboratory has continually explored various composite materials as tissue engineering scaffolds. Initial work primarily focused on polymer composites and polymer-ceramic composites, but as new advances have been made in chemistry and materials science, we have incorporated new materials into tissue engineering constructs.

Polymer-ceramic composites

As previously described, PPF/ β TCP composites were among the earliest materials investigated in our laboratory. PPF/ β TCP composites were found to have increased compressive moduli and strength compared to PPF alone, and the composites degraded in vivo with a mild inflammatory response following implantation.⁹³ In collaboration with Dr. Michael J. Yaszemski of the Mayo Clinic in Rochester, Minnesota, we investigated moldable PPF/ β TCP paste as an alternative to the currently used poly(methyl methacrylate) (PMMA) bone cements. PPF/ β TCP pastes had comparable mechanical properties to PMMA and, in contrast to PMMA, were biodegradable and cross-linked below 50°C, well below 94°C, the potentially toxic curing temperature of PMMA.⁹⁴ Layered composites were also fabricated,⁹⁵ and adherent marrow stromal cells attached and proliferated well on the composites.⁹⁶

Our more recent studies of polymer-ceramic composites have been performed with our collaborator, Dr. John A. Jansen of Radboud University Nijmegen Medical Center in the Neth-

erlands. Most of this work has focused on CaP cements augmented with PLGA microspheres. These materials exhibit excellent biocompatibility, have improved biodegradability over CaP cements alone, and can be used as injectable materials.^{48,97,98} In vivo bone regeneration using these materials has been promising,⁹⁹ and more recently our laboratories have been investigating both release of the osteogenic growth factor bone morphogenetic protein 2 (BMP-2) from PLGA microspheres¹⁰⁰ as well as the inclusion of gelatin microparticles in CaP cements as porogens or drug delivery vehicles.^{46,101} With the importance of biological interactions at the nanoscale becoming more apparent, we have also developed methods for the dispersion of calcium phosphate nanocrystals within OPF hydrogels.¹⁰²

Nanocomposite scaffolds

Commensurate with the need for scaffolds with enhanced compressive mechanical properties, our laboratory has investigated methods to reinforce polymer scaffolds. Earlier attempts in this direction used short hydroxyapatite fibers to reinforce PLGA foams and succeeded in enhancing the compressive yield strength;¹⁰³ more recent attempts at composite scaffold reinforcement have focused on nanoscale scaffold reinforcement. A variety of surface modified carboxylate alumoxane nanoparticles were dispersed within PPF/PF-DA scaffolds and found to cause a threefold increase in flexural modulus over polymer resin reinforced scaffolds.¹⁰⁴ Following an accelerated degradation protocol, alumoxane reinforced PPF scaffolds degraded faster than nonreinforced scaffolds, and the inclusion of alumoxane nanoparticles did not affect scaffold cytotoxicity or biocompatibility.¹⁰⁵ After 12 weeks of in vitro degradation, porous PPF/alumoxane composites maintained their compressive mechanical properties, pore morphology, and overall scaffold size despite mass losses of over 5% due to degradation.¹⁰⁶ Following 12-weeks implantation within a goat hindlimb condylar defect, PPF/alumoxane composites were found to degrade and regenerate bone similar to control PPF/PF-DA scaffolds, indicating that the improved mechanical properties of PPF/alumoxane nanocomposites are gained without a concomitant decrease in degradation or bone healing in vivo (Mistry et al., submitted).¹⁰⁶ Despite these positive results, other efforts in our laboratory involving nanoreinforced scaffolds aimed to achieve superior scaffold mechanical properties with corresponding improvements in in vivo bone regeneration.

Carbon nanotube composite scaffolds

We have also investigated the incorporation of single-walled carbon nanotubes (SWNTs) dispersed within PPF scaffolds. With respect to enhancement of mechanical properties, an ideal concentration of 0.05 wt % resulted in a 74% increase in compressive modulus and a 69% increase in flexural modulus.¹⁰⁷ At higher than 0.05 wt %, nanotube aggregates were observed. Functionalized SWNTs were synthesized and interacted with PPF chains, increasing cross-linking densities of PPF networks and facilitating load transfer between the polymer and the nanotubes thereby resulting in threefold increases in compressive and flexural moduli and twofold increases in compressive and flexural strengths over nonreinforced PPF.¹⁰⁸ Incorporation of nanotubes did not

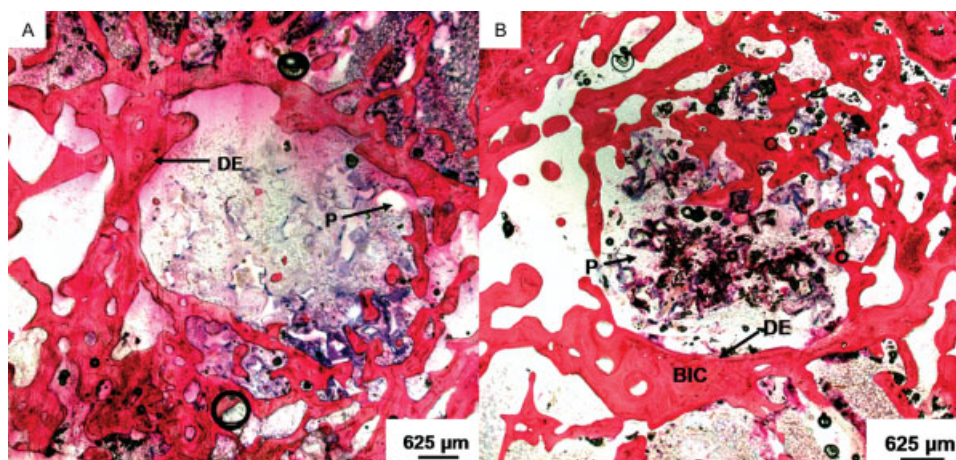


Figure 5. Representative histological sections of scaffolds implanted in femoral condyle defects: (A) a PPF/PF-DA scaffold after 12 weeks, and (B) an ultrashort carbon nanotube/PPF scaffold at 12-weeks post-implantation. The images are presented at $\times 1.6$ magnification.

The PPF scaffold (P) appears as white areas in all images. The original defect edge (DE) is visible in the low magnification images. Bone-like tissue appears red; direct bone-implant contact (BIC) and bony ingrowth occurred with the nanocomposite scaffold 12 weeks after implantation.¹¹² Reproduced from Ref. 112, with permission from Elsevier. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

increase the cytotoxicity of the material,¹⁰⁹ and marrow stromal cells attached and proliferated on porous scaffolds augmented with ultrashort nanotubes.¹¹⁰ Additionally, these materials have the potential to be used as injectable biomaterials.¹¹¹ Twelve weeks following implantation in a rabbit condylar defect, PPF/ultrashort nanotube composites displayed a threefold greater bone tissue ingrowth than control PPF/PF-DA scaffolds, and additionally, markedly fewer infiltrating inflammatory cells and more highly organized connective tissue were observed around the nanocomposites (Figure 5).¹¹² Because of their greatly increased mechanical properties combined with improved *in vivo* capacity for bone regeneration compared to PPF alone, PPF/SWNT composites are a promising material for bone and other hard tissue engineering applications. Additionally, the development of methods for functionalization and dispersion of nanotubes within biomaterial matrices could be used to augment other biomaterials or to make materials with limited mechanical strength appropriate for new applications within tissue engineering and biology.

Injectable Carriers for Cellular Delivery

An area of research that holds great promise in biomaterials science and tissue engineering is the development of injectable materials and systems for clinical applications. Injectable systems can be delivered to a patient using minimally invasive methods and can fill complex tissue defects without the need for extensive imaging and prefabrication. Perhaps most importantly, injectable systems that use water as a solvent can potentially be used to deliver cells and water soluble growth factors or drugs for therapeutic purposes.¹¹³ In biology, cell-based technology is rapidly expanding and cellular therapeutics are moving closer to becoming clinical reality, making the need for delivery vehicles and injectable matrices increasingly apparent.

We investigated PPF/gelatin microparticle composites as an injectable system for tissue engineering. Marrow stromal cells, a subpopulation of which are osteoblastic precursors or MSCs, maintain their viability and differentiation capacity when encapsulated within gelatin microparticles.¹¹⁴ This encapsulation process protects MSCs during PPF cross-linking,^{115,116} and amelioration of the potentially toxic effects of being exposed to uncross-linked macromers and initiators renders the system viable for injectable cell delivery.

Cell adhesion to P(PF-co-EG) hydrogels has also been investigated along with cell viability during *in situ* cross-linking using the water soluble APS initiator system described previously.^{117,118} Cells adhered to the cross-linked gels and the viability of cells present during cross-linking was above 30% for some formulations, making cell delivery for *in vivo* applications possible.

OPF has been extensively studied as a carrier for cell delivery. OPF incorporating low molecular weight PEG chains was originally found to elicit a minimal inflammatory response *in vivo*¹¹⁹ and had cytotoxicity below 25% after 24-h exposure *in vitro*.¹²⁰ These promising results led to a series of studies investigating cell encapsulation within OPF hydrogels.

Because of their high water content, hydrogels are often thought of as ideal carriers for cell encapsulation and delivery; however, a number of parameters must be considered and optimized to ensure appropriate material properties in combination with maintained cell viability and differentiation. Our early encapsulation studies focused on identifying radical initiators and reducing agents as well as proper concentrations of chemicals from both classes that would achieve ideal gelation kinetics without large fluctuations in pH or decreased encapsulated cell viability. The effects of two persulfate oxidizing agents and three reducing agents derived from ascorbic acid were measured with respect to encapsulated rat marrow stromal cell survival, and initiator

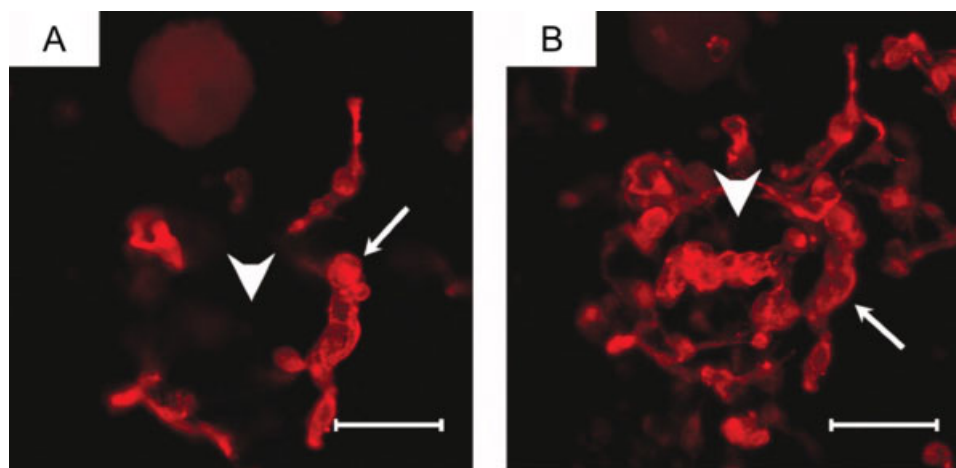


Figure 6. 2D image (A) and z-axial projection stack image (B) of rabbit marrow MSCs in OPF hydrogel composites with both IGF-1-loaded microparticles and TGF- β 1-loaded microparticles at Day 14.

Samples were incubated with rhodamine phalloidin solution for 1 h and imaged with confocal fluorescence microscopy. Scale bar represents 100 μ m. Small arrows indicate rabbit marrow MSCs and large arrows indicate microparticles in hydrogel composites.¹⁴⁰ Reproduced from Ref. 140, with permission from Wiley Interscience. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

concentration and final pH were recognized as key parameters affecting cell survival.⁸⁵ Encapsulation of rat marrow stromal cells in the presence of cell culture media containing osteogenic supplements demonstrated that encapsulated cells retained the ability to differentiate into osteoblast-like cells,¹²¹ and OPF hydrogels with greater swelling due to incorporation of higher molecular weight PEG support greater osteoblastic differentiation of encapsulated cells than OPF hydrogels that undergo less swelling.¹²²

Native cartilage consists of a largely acellular matrix composed primarily of water with interspersed chondrocytes, a structure that is well-approximated by encapsulating chondrocytes within a hydrogel matrix. OPF hydrogels have been used to deliver encapsulated chondrocytes and MSCs for cartilage regeneration, although this delivery has typically been in conjunction with the delivery of growth factors for inducing differentiation of encapsulated cells and surrounding host cells into chondrocytes.^{123,124} This and other examples of biomaterial facilitated delivery of bioactive molecules has been an expanding focus of our work and has led to a number of diverse and novel techniques for bioactive factor generation and delivery.

Growth Factor Carriers and Scaffolds

The continued focus and progressive discoveries in areas such as stem cell biology, cell signaling, and tissue specific microenvironments or niches has led to the identification of a number of key growth factors necessary for tissue repair or regeneration. In addition to studying the release of common pharmaceuticals from biomaterial matrices and particles to treat localized infection or disease,^{125–129} we along with many other researchers in tissue engineering have been attempting to mimic natural healing or development by delivering exogenous growth factors, often simultaneously with the cells upon which they are intended to act, to speed or enable tissue regeneration.

Transforming growth factor- β 1 (TGF- β 1) is a 25 kilodalton cytokine that is nearly ubiquitously expressed by immune cells and likely plays an important role in cell differentiation and wound healing.¹³⁰ Consequently, it is frequently investigated in tissue engineering applications. We have investigated TGF- β 1 coating of scaffolds and release from PLGA/PEG microparticles and found increased osteoblast proliferation and enhanced deposition of osteogenic markers.^{131–134} Gelatin microparticles, a popular drug delivery vehicle,¹³⁵ have also been utilized for TGF- β 1 release from OPF hydrogels^{136,137} and injectable calcium phosphate cements.¹⁰¹

We have extensively studied the release of TGF- β 1 from OPF hydrogels for cartilage tissue engineering. Bilayered OPF hydrogels with TGF- β 1 loaded into the superficial layer of the gel were used to repair osteochondral defects in a rabbit model,¹³⁸ and in vitro studies were undertaken to investigate the simultaneous delivery of chondrocytes and TGF- β 1.¹²³ Following promising early results in these studies, including a noted maintenance of chondrocytic phenotype for encapsulated cells, we, in collaboration with Dr. Arnold I. Caplan of Case Western Reserve University, demonstrated that rabbit mesenchymal stem cells encapsulated in OPF with TGF- β 1 loaded gelatin microparticles differentiate into chondrocyte-like cells with upregulated collagen II and aggrecan expression.¹²⁴ Expanding on successful TGF- β 1 release systems for cartilage engineering, we explored the tandem release of TGF- β 1 and insulin-like growth factor 1 (IGF-1). By varying the cross-linking extent and isoelectric point of gelatin microparticles, TGF- β 1 and IGF-1 release from OPF hydrogels was tailored to mimic the release profiles found naturally.¹³⁹ In vitro studies showed that marrow stromal cells exposed to both TGF- β 1 and IGF-1 had significantly upregulated expression of chondrogenesis-related genes compared to controls (Figure 6).¹⁴⁰ In vivo assessment of this dual release technique showed improved cartilage healing in defects treated with IGF-1 releasing hydrogels; however, this improvement was not observed with scaffolds releasing both

growth factors (Figure 7).¹⁴¹ This surprising result confirmed that a complex interplay between growth factors and the native healing site is at work, and careful evaluation of delivery strategies, even those that seemingly approximate a natural healing response, is necessary.

We have also studied dual growth factor delivery for bone tissue engineering. Using the well-characterized gelatin microparticle system for release of BMP-2 and vascular endothelial growth factor (VEGF),^{142,143} we, in collaboration with Dr. Yasuhiko Tabata of Kyoto University in Japan, found after 4-weeks implantation within a rat cranial defect, dual release of VEGF and BMP-2 regenerated significantly more bone tissue than release of either growth factor alone (Figure 8).²⁷ After 12 weeks, scaffolds releasing BMP-2 alone produced similar tissue regeneration to dual release scaffolds, indicating that neovascularization can promote improved bone regeneration early in the healing process. Subsequent studies found that vascularization as induced by VEGF is insufficient to rescue bone growth in scaffolds releasing lower doses of BMP-2, illustrating the need for a balanced interplay between angiogenesis and osteogenesis to yield a therapeutic effect.¹⁴⁴ Previous studies have demonstrated the efficacy of BMP-2 release from different scaffold-based systems for bone tissue engineering applications.^{145–147}

Work in our laboratory elucidating the release of angiogenic and osteogenic growth factors during bone healing did not begin with the study of VEGF and BMP-2 release. Prior work studied the release of TP508, an angiogenic and osteogenic 23 amino acid peptide sequence derived from thrombin.¹⁴⁸ TP508 released from PPF-based composite scaffolds enabled healing of segmental long bone defects in rabbits, and we found that the release kinetics of the peptide were of critical importance to bone healing, with a burst release of the peptide facilitating bony bridging in over 80% of the defects after 12 weeks.¹⁴⁹ More recent work in this area in collaboration with Dr. Achim Goeperich of the University of Regensburg in Germany has studied whether amino acid or bisphosphonate modification can target peptides selectively to bony sites, possibly decreasing necessary dosing by improving the efficiency of delivery.¹⁵⁰

Biomimetic Hydrogels

Peptide sequences have been utilized in our laboratory for purposes other than controlled release to stimulate tissue regeneration. Many of our aforementioned studies of material-cell interactions noted varying levels of cell attachment based on material properties, most notably decreases in cell attachment as the hydrophilicity of OPF and P(PF-co-EG) hydrogels increased. In addition to modulating material-cell interactions by varying material compositions, surface patterning of materials was previously used in our laboratory to regulate RPE attachment and morphology.^{151,152} Cyclic Arg-Gly-Asp (RGD) peptide sequences, long known to bind surface integrins necessary for cell attachment and migration in vivo, were first used in our laboratory to block integrin domains to study growth factor stimulated migration.¹⁵³ A later series of work built upon our previous work in surface patterning and also the relationship of RGD to cell attachment by modifying hydrogels with RGD sequences to

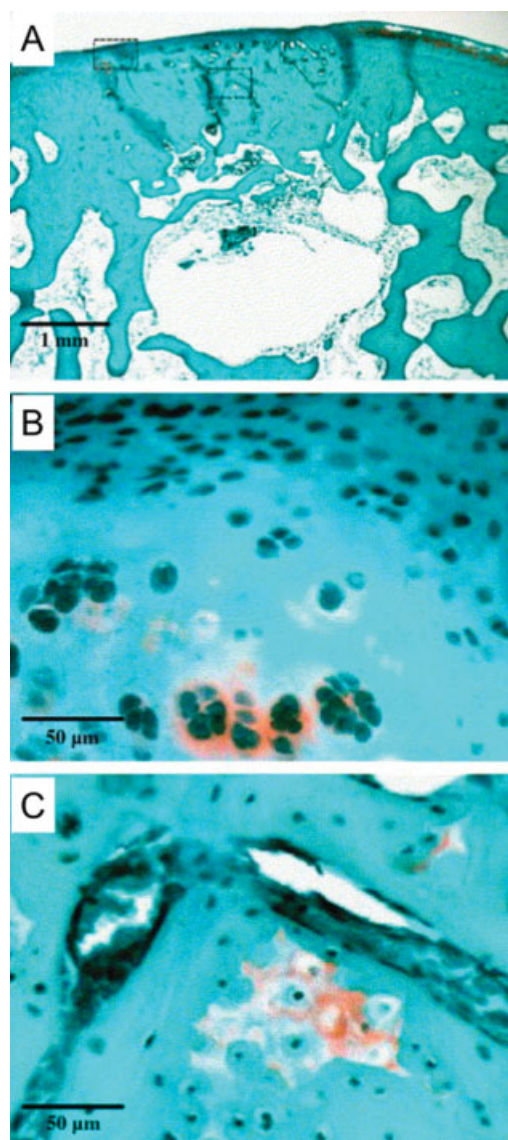


Figure 7. Histological section displaying fibrocartilage in-growth near the chondral defect margins and significant subchondral restoration.

The boxed regions in (A) ($\times 2$ magnification) are shown at $\times 20$ magnification to illustrate the spherical shape of cartilage cells in the neo-surface (B) and small regions of remodeling tissue in subchondral region (C). This defect was treated by IGF-1 delivery.¹⁴¹ Reproduced from Ref. 141, with permission from Elsevier. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

promote cell attachment and allow the hydrophilic materials to be more suitable for cell-based therapies and tissue regeneration.

Using 4-nitrophenyl chloroformate as an activator, OPF hydrogels were modified by covalently attaching Gly-Arg-Gly-Asp (GRGD) peptides.¹⁵⁴ Later incorporating a PEG spacer to avoid steric obscuration of the peptide, we found that marrow stromal cells adhered significantly better to GRGD-modified OPF hydrogels, and the sequence specific role of GRGD was confirmed by competitive inhibition of

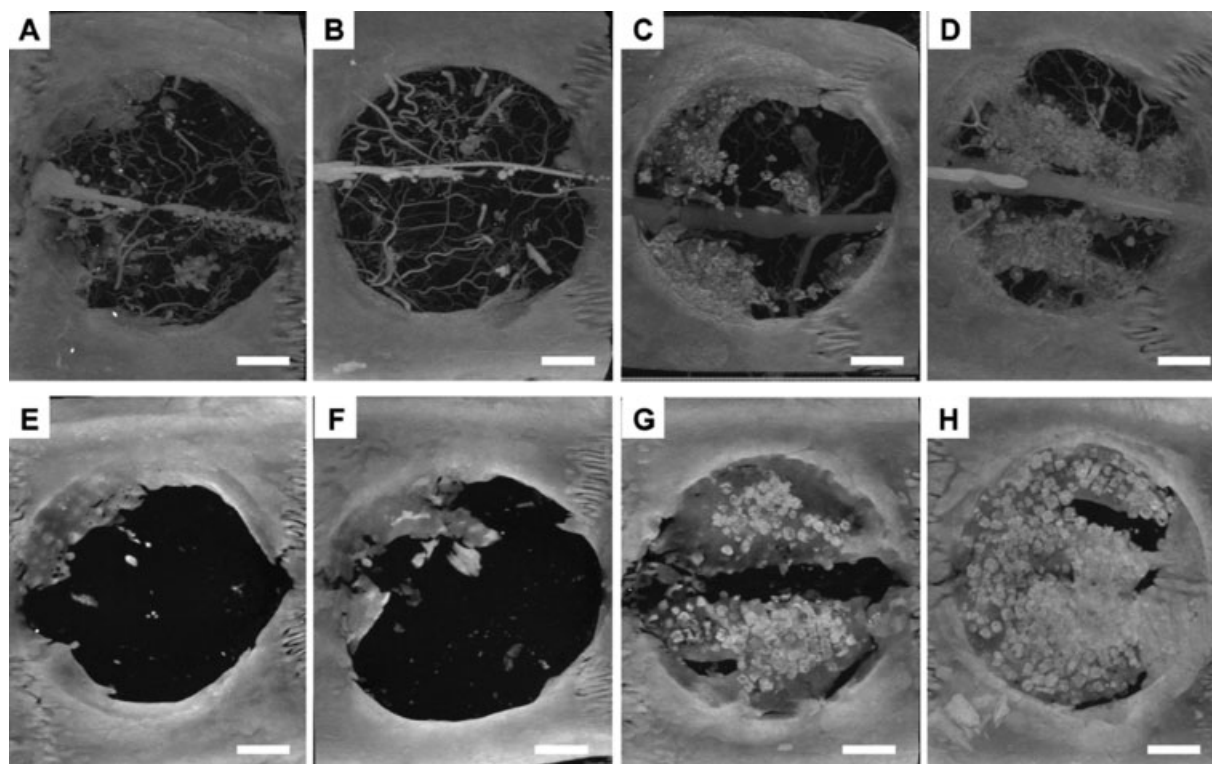


Figure 8. Microcomputed tomography images (maximum intensity projections) of cranial defects taken at 4 and 12 weeks after implantation.

Panels A–D represent scaffolds releasing no growth factor (control), VEGF, BMP-2, and both VEGF and BMP-2 respectively at 4 weeks prior to decalcification. Blood vessels were filled with a silicon-based radiopaque material so that both blood vessels and mineralized tissue are visible. Panels E–H represent scaffolds releasing no growth factor (control), VEGF, BMP-2, and both VEGF and BMP-2 respectively at 12 weeks; no blood vessels were visible because perfusion with the radiopaque material was not done at this time point. Bar represents 200 μm for all panels. Scaffolds releasing both growth factors exhibited significantly greater amounts of bone-like tissue regeneration after 4 weeks but were not significantly different from scaffolds releasing BMP-2 only after 12 weeks.²⁷ Reproduced from Ref. 27, with permission from Elsevier.

cell attachment following incubation with free GRGD.¹⁵⁵ RGDS modification of P(PF-co-EG) hydrogels found concentration dependent attachment of marrow derived osteoblasts.¹⁵⁶ Although other work investigating bulk modification of hydrogels to improve cell attachment using agmatine^{117,157,158} and osteopontin derived peptides,¹⁵⁹ an important discovery that followed our ongoing work in the area of bioreactors was the role surface peptide sequences played towards influencing cell differentiation. RGD-modified hydrogels were capable of inducing progenitor cells to differentiate down an osteoblastic lineage, even in the absence of culture supplements normally required for differentiation.^{160,161}

Generated Matrix with Bioreactors

Flow perfusion bioreactor systems

The seemingly anomalous decrease in bone mineral density found in astronauts returning from prolonged missions in space¹⁶² coupled with the prevention of this bone loss by repeated mechanical loading¹⁶³ are indicative of the importance of mechanical stimulation on the maintenance of healthy bone tissue. Comparing a spinner flask, rotary vessel, and flow perfusion bioreactor, we began investigating the

effects of convection on bone growth within biomaterial scaffolds, surmising that convection of media could both overcome diffusional limitations encountered during static culture of large scaffolds and provide a mechanical stimulus capable of augmenting bone formation.^{164,165} Flow perfusion systems were determined to be the most appropriate method for achieving these effects¹⁶⁴ and designed specifically for bone tissue engineering applications within our laboratory.¹⁶⁶

When seeded on titanium fiber mesh scaffolds within a flow perfusion bioreactor, marrow stromal cells differentiate into osteoblast-like cells and deposit far greater mineralized matrix than cells in constructs cultured under static conditions.^{20,167} Similar results were observed using PLLA fiber mesh scaffolds,¹⁶⁸ starch-based scaffolds,¹⁶⁹ and CaP scaffolds.¹⁷⁰ Furthermore, by varying the fluid viscosity such that fluid shear stress could be varied but transport of nutrients and removal of waste remained constant, we found that mineral deposition increased and extracellular matrix distribution was more even, revealing a dose dependent relationship between shear stresses and mineralized tissue generation.¹⁷¹ In vivo implantation of cell-scaffold constructs following culture in a flow perfusion bioreactor showed the constructs to be osteoinductive and that the in

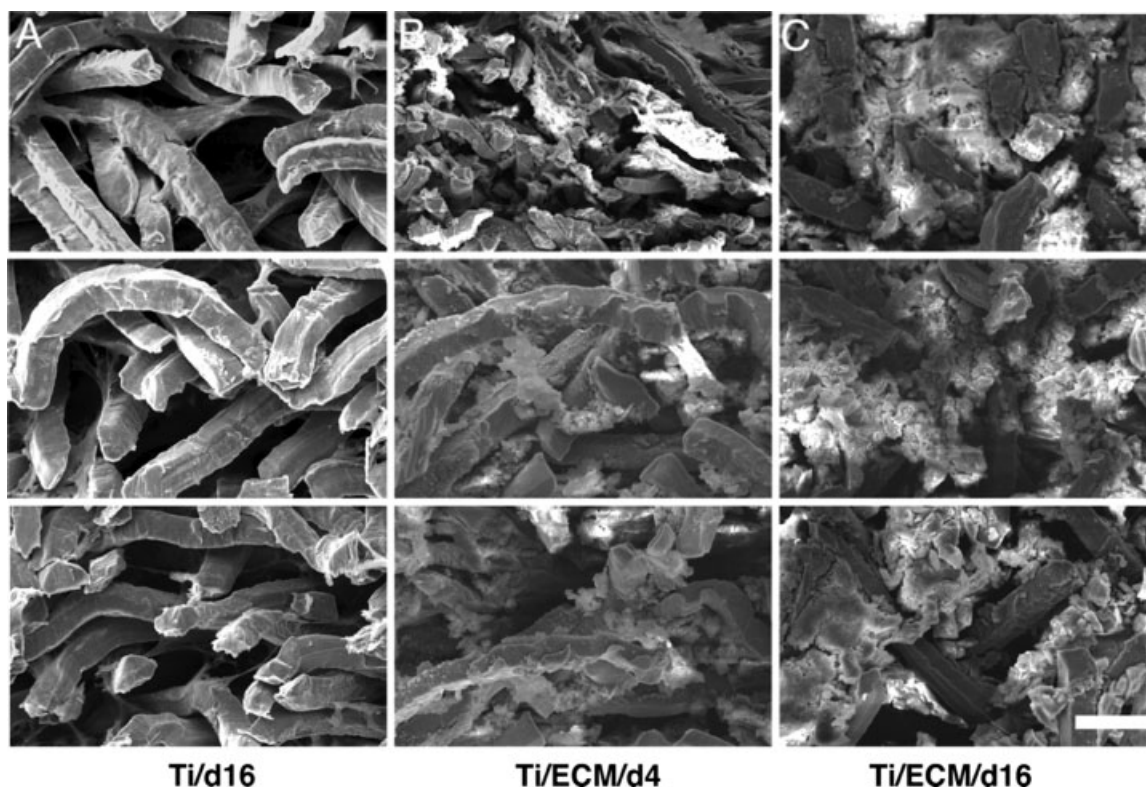


Figure 9. SEM images obtained from the cross-sections of 0.8-mm thick constructs cultured in vitro in a flow-perfusion bioreactor.

Each image consists of three panels. (Top) One-hundred micrometers from the top of the construct. (Middle) The middle of the construct. (Bottom) One-hundred micrometers from the bottom surface. These images show the plain Ti construct after 16 days of culture (A), the Ti/ECM construct after 4 days of culture (B), and the Ti/ECM construct after 16 days of culture (C). (Scale bar, 50 μm for all SEM images.)¹⁷⁷ Reproduced from Ref. 177, with permission from The National Academy of Sciences, USA.

vitro culture period can affect in vivo outcomes following implantation.¹⁷²

Continuing work using the flow perfusion bioreactor system determined a time dependent affect of titanium fiber network mesh size on cell differentiation and matrix deposition,¹⁷³ and increasing the overall porosity of degradable starch-based scaffolds yielded significantly greater matrix deposition and cell proliferation when cultured under flow conditions.¹⁷⁴ In the absence of culture in the flow perfusion system, implantation of undifferentiated stromal cells on titanium fiber meshes led to poor bone regeneration when compared to cells precultured in dexamethasone, a factor known to induce osteogenic differentiation of MSCs.¹⁷⁵ These results showed that without flow perfusion, culture on titanium fiber meshes, even in the presence of RGD peptide to facilitate cell attachment, was not sufficient to differentiate cells to an extent at which bone regeneration in vivo was similar to that of constructs seeded with differentiated cells.

In vitro generated ECM

The presence of fluid shear stresses within the flow perfusion bioreactor, however, was found to be sufficient to induce osteoblast-like differentiation of marrow stromal cells in the absence of dexamethasone.¹⁷⁶ Although dexa-

methasone and shear stresses in tandem have a synergistic effect on cell differentiation, a series of pivotal studies determined the utility of culture within the flow perfusion bioreactor towards differentiating cells in the absence of media supplements.

Titanium fiber meshes seeded with marrow stromal cells were cultured for 12 days in the flow perfusion system to generate bone-like extracellular matrix (ECM), after which the scaffolds were decellularized using repeated freeze-thaw cycles. Subsequent culture of MSCs in the absence of dexamethasone on scaffolds with previously deposited ECM showed a 40-fold increase in mineralization compared with control scaffolds. Similar increases were found compared to scaffolds with denatured ECM. The presence of dexamethasone further enhanced matrix deposition; however, this work demonstrated the powerful capacity of flow and in vitro generated ECM to guide progenitor cells down an osteogenic lineage (Figures 9 and 10).¹⁷⁷ Previous work had shown that ECM/titanium composites enhanced differentiation of MSCs cultured statically,¹⁷⁸ but the drastic increases in matrix deposition in the absence of dexamethasone were most apparent when ECM was combined with mechanical forces imparted by the flow perfusion system. Localization of growth factors following flow perfusion on starch-based scaffolds found TGF- β 1, fibroblast growth factor, VEGF, and BMP-2 deposited on scaffolds cultured under flow, with

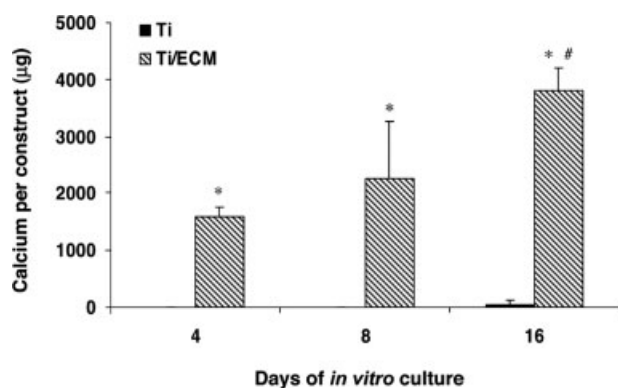


Figure 10. The calcium content of Ti and Ti/ECM constructs cultured in the flow-perfusion bioreactor after 4, 8, and 16 days of culture.

The data represent means of four samples, with the error bars representing the standard deviations. Statistical differences ($P < 0.05$) between Ti and Ti/ECM constructs are indicated with an asterisk; # designates a statistical difference ($P < 0.05$) between all other data points.¹⁷⁷ Reproduced from Ref. 177, with permission from The National Academy of Sciences, USA.

increases in deposition area found following culture at higher flow rates.¹⁷⁹ In vivo implantation of titanium/ECM composite scaffolds revealed an increase in scaffold vascularity corresponding to increases in the amount of deposited matrix present.¹⁸⁰ Marrow stromal cells cultured on matrices with previously deposited mineralized ECM showed significant increases in osteoblast-specific genes, likely due to the prior deposition of growth factors and matrix molecules by cells during the original mineralization.¹⁸¹ Using electrospinning, we have also fabricated and characterized poly(ϵ -caprolactone) fiber mesh scaffolds with mixed micro- and nanofiber layers to simulate the scale of the ECM, and this physical mimicry of the ECM facilitated increased cell spreading compared to scaffolds without the nanofiber domains (Figure 11).⁵¹

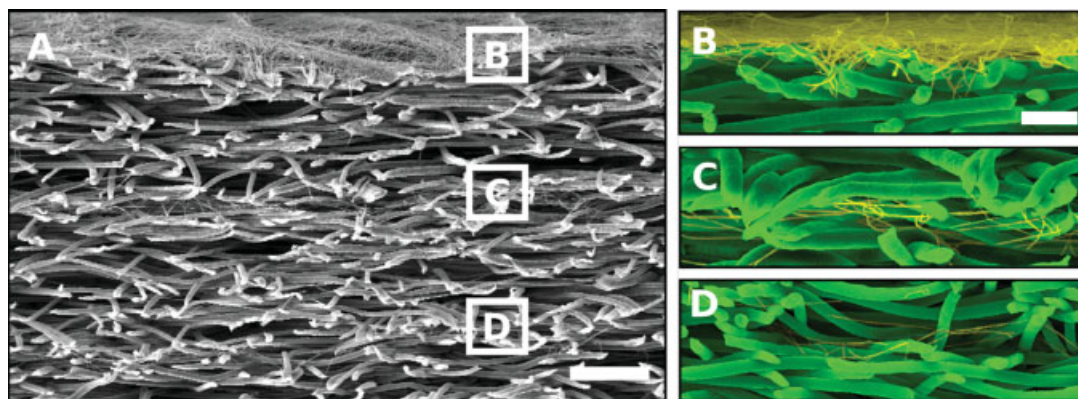


Figure 11. Scanning electron micrographs of cross-sections of layered scaffolds generated by sequential electrospinning.

(A) Cross-section illustrating (from top to bottom) a nano-micro-nano-micro-nano-microfiber layered scaffold. The white boxes correspond to the nanofiber layers and their respective magnified images shown to the right. (B) Magnification of the nanofiber layer electrospun for 5 min. (C) Magnification of the nanofiber layer electrospun for 30 s. (D) Magnification of the nanofiber layer electrospun for 30 s. The scale bar shown for (A) is 100 μm , and for (B–D) it is 25 μm . Microfibers are false colored green while nanofibers are false colored yellow to enhance contrast.⁵¹ Reprinted with permission from.⁵¹ Reproduced from Ref. 51, with permission from ACS publications. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

Injectable Plasmid DNA Carriers

Tuning of material properties, growth factor delivery, culture in supplemented media, and induction via exposure to generated extracellular matrix have all been demonstrated as viable methods for influencing cell behavior and specifically for promoting the differentiation of progenitor cells down tissue specific lineages. All of these approaches have indirect effects that likely influence or regulate gene transcription and/or expression. More directly towards this end, we have also investigated material mediated methods for the delivery of plasmid DNA. Our earliest work related to this area used PLGA microparticles to deliver antisense oligodeoxy-nucleotides to inhibit smooth muscle cell proliferation and migration, phenomena which are among the leading causes of restenosis after vascular intervention.¹⁸²

Poly(ethylenimine)/DNA complexes

Subsequent studies focused on the delivery of plasmid DNA using the polycation poly(ethylenimine) (PEI) as a delivery vehicle. Modification of standard PEI/DNA complex delivery ameliorated PEI toxicity and improved packing of PEI around plasmid DNA. The end result was improvement in PEI transfection efficiency from 37 to 53%.¹⁸³ PEI size matters as well; transfection efficiency using green fluorescent protein transfection of human endothelial cells increased with increasing PEI molecular weight. PEI with molecular weights of 70,000 Da successfully transfected 25.6% of cells, while PEI with molecular weights under 1,800 Da did not transfect any cells.¹⁸⁴ A mechanistic study of PEI transfection revealed the pathway of transfection from cellular uptake via endocytosis through nuclear localization and transfection (Figure 12).¹⁸⁵ This study also found that PEI, independently of complexation with DNA, localizes to the nucleus through the same pathway, and a later study determined that contrary to the commonly held hypothesis that endocytosed PEI merges with lysosomes, endocytosed PEI enters the nucleus without binding to lysosomes.¹⁸⁶ When

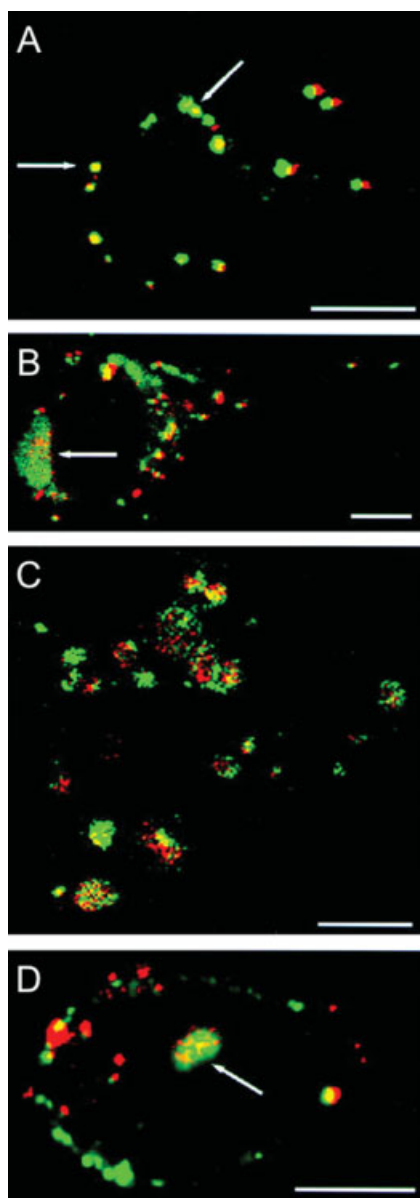


Figure 12. Tracking of double-labeled PEI/DNA complexes.

The fluorescence patterns for single-labeled complexes are also seen for double-labeled complexes. (A) At 2-h post-transfection, visible complexes appear as clumps on the exterior of cells, as indicated by arrows. (B) At 3-h post-transfection, both surface aggregation of complexes and endosomal formation (indicated by an arrow) are visible. The arrow indicates endosomal formation. (C) At 4-h post-transfection, endosomes containing both PEI and DNA are visible throughout the cell cytoplasm. (D) At 4.5-h post-transfection, fluorescent structures containing both PEI and DNA inside the cell nucleus are present, as indicated by the arrow. (Bar = 10 μ m).⁸⁵ Reproduced from Ref. 185, with permission from Wiley Interscience. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

used to transfect endothelial cells with a reporter plasmid, we found a bimodally distributed pattern of cell death, attributable to death from free PEI and PEI/DNA complexes, although transfection was successful and translation increased.¹⁸⁷ More recently we have used branched PEI conjugated with

hyaluronic acid to improve cell viability and transfection efficiency over branched PEI alone.¹⁸⁸

Gene transfection for tissue engineering applications

In addition to polycationic gene delivery using PEI as a model delivery vehicle, we have also studied more conventional methods of gene transfection for tissue engineering. In the first study comparing transfection vectors for the delivery of osteogenic genes, adenoviral, retroviral, and cationic lipid vectors were used to transfect rat marrow stromal cells with the gene encoding human BMP-2. Both in vitro and in vivo, cells transfected with adenoviral vectors displayed enhanced osteogenic and bone healing potential vs. those transfected with other vectors and controls.¹⁸⁹ Culture in dexamethasone led to a threefold increase in transgene expression over culture in nonsupplemented media when using adenoviral vectors but did not increase transgene expression from other vector types.¹⁹⁰

Sustained plasmid DNA release from gelatin microspheres embedded within OPF hydrogels was studied for use in bone tissue engineering applications. Entrapment within cationized gelatin microspheres and subsequent embedding within OPF extended DNA bioavailability, and release was primarily mediated by OPF degradation.^{191–193} Attempts to regenerate critical-sized bone defects in vivo using plasmid DNA encoding BMP-2 released from gelatin microparticles were unsuccessful compared to controls, perhaps due to poor release or most likely the lack of a vector.¹⁹⁴ Future improvements to the scaffold-based delivery system, such as incorporation of adenoviral or PEI-based vectors, may improve tissue regeneration; however, the vector technologies developed hold great potential in future applications.

Enabling Technologies and Translational Approaches

As the fields of tissue engineering and regenerative medicine evolve, need arises for reliable tools to evaluate emerging technologies. Although reporting of specific combinations of cells, bioactive factors, and biomaterials as a whole help to advance the field, the emergence of new techniques in material and scaffold fabrication, drug and cell delivery, and the interaction of these factors continue to be the driving forces pushing strategies closer to the clinic. Although difficult to specifically categorize, our work in enabling technologies and new approaches to addressing problems that potentially can be solved by tissue engineering have continued relevance today. Work in the areas of cell culture techniques,^{195,196} nanomaterials for high resolution MRI,¹⁹⁷ and evaluation of scaffold and cellular technologies^{198,199} have proven useful for the advancement of the field. Strategies similar to those based on cell lines created through our collaborations²⁰⁰ have been used to evaluate new strategies in tissue engineering.^{201,202} The development of animal models^{203–205} and modalities to evaluate tissue regeneration within those models²⁰⁶ will continue to be necessary as expansion into engineering a variety of tissues and multiple tissues occurs. Through collaboration with Dr. Mark Wong of the University of Texas Health Science Center Dental Branch in Houston, we have recently developed a model to investigate alveolar

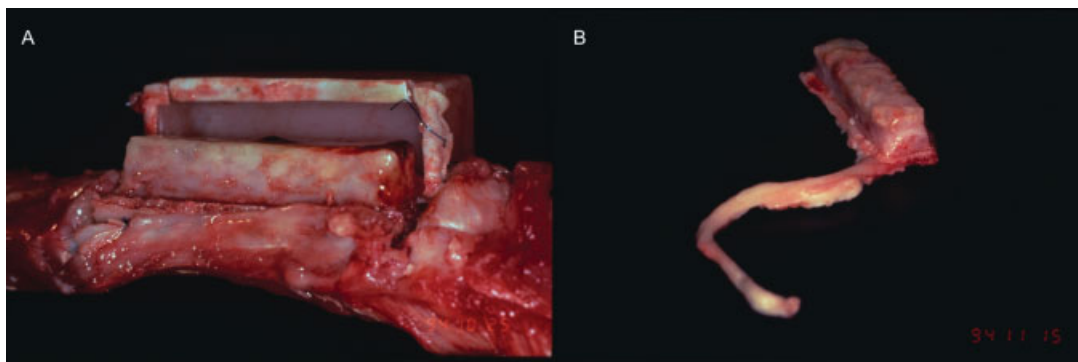


Figure 13. Vascularized bone developed using an in vivo bioreactor approach.

(A) Molded block of bone harvested after 6 weeks conforming to a $10 \times 40 \times 10\text{-mm}^3$ PMMA chamber and attached to the vascularized periosteal bed. Note the polytetrafluoroethylene (PTFE) cuff bonded to the perimeter of the chamber, used to sew the implant to the periosteum. (B) Molded bone flap attached to vascularized pedicle (artery and vein) removed from a $10 \times 40 \times 10\text{-mm}^3$ chamber implanted for 6 weeks.²⁰⁸ Reproduced with permission from Ref. 208, with permission from John Wiley & Sons. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

bone regeneration within a nonhealing defect in the rabbit mandible,²⁰³ and through this collaboration we continue to explore further clinically relevant animal models and tissue engineering strategies. An animal model developed early in the course of studies in our laboratory represents a different in vivo approach; rather than the evaluation of tissue engineering strategies, a sheep model for bone flap creation was developed to provide transplantable vascularized bone flaps.²⁰⁷

The sheep model developed in collaboration with Drs. Michael J. Miller and Alan W. Yasko, now at The Ohio State University and Northwestern University's Feinberg School of Medicine respectively, utilized a seldom used but clinically relevant approach to tissue engineering.²⁰⁷ Rather than the more common approach of regenerating tissues at the site where they are needed and a defect exists, PMMA chambers filled with morcellized bone graft were implanted adjacent to the ribs of a sheep, with the chamber open to the rib periosteum. New, well formed bone penetrated the implants after 6 weeks, and removal of the newly formed bone with an attached vascular pedicle yielded a vascularized implant that could be used for autologous tissue augmentation at a distant defect site (Figure 13).²⁰⁸ PLGA foam filled chambers were also investigated for use in this technique.²⁰⁹ Recently, Cheng et al. applied a similar technique to reconstruct the mandible of a patient with a locally invasive squamous cell carcinoma, and after 1 year the implanted bone flap remained viable such that dental implants could be placed within it, allowing a near complete reconstruction of the patient's mandible.²¹⁰ Strategies such as this may prove useful in treating traumatic injuries where inflammation or localized infection may require treatment before attempts at tissue reconstruction or regeneration can be made.

Future Directions in Biomaterials Science and Tissue Engineering

Within the fields of biomaterials and tissue engineering, a number of emerging and recent trends will likely shape the future and lead to even greater successes in both the labora-

tory and also the clinic. Tissue engineering has always been tied to discoveries in a number of diverse fields, and this trend will continue.

From biology, greater knowledge about the role stem cells play in tissue regeneration and healing will play a pivotal role in translating tissue engineering technologies into the clinic. As we better understand the capabilities and limitations of these progenitor cells, as well as gain better understanding of the extracellular and intracellular pathways that determine their differentiation in vivo, researchers in biomaterials science and tissue engineering will be able to fabricate better materials upon which stem cells can proliferate and differentiate, devise more tailored cell delivery systems, and understand the applications for which these cells are best suited. Similarly, an increased knowledge of the role more diffuse biological processes, such as inflammation and development, play in tissue regeneration will allow future work to focus on crafting materials and identifying bioactive factor delivery regimes to modulate, recreate, or exploit these natural phenomena.

Materials science and the specific field of biomaterials science will continue to develop under a reciprocal exchange of ideas and new technologies. As nanotechnology continues to develop and we better appreciate and understand the importance of nanoscale stimuli and interactions in biology, biomaterials with tailored nanoscale properties will be fabricated and used to mimic, stimulate, and augment biological processes. Given the clinical success of many growth factors and growth factor delivery systems, spatiotemporal control of bioactive molecule release and the release of multiple factors simultaneously and in series to mimic useful biological cascades must be explored.

Finally, in the near future we predict advances in two broad areas. The first, as described earlier, will be advances in our understanding of natural biological processes critical to tissue engineering, the integration of this understanding into specific tissue engineering technologies, and a continued pursuit of new biomaterial technologies integrating advances in chemistry, biology, and materials science. The second area for major advancement will be in harnessing existing tech-

nologies for clinical use. Motivated by the clinical success of products related to tissue engineering, clinicians and researchers are now gaining a better understanding of the toolbox available to them in the form of FDA regulated technologies. Even as better, more application-specific approaches are being developed and in their early stages, significant translational advances will come from novel integration of existing technologies. These advances will be the nidus for bringing new and better technologies in tissue engineering through the regulatory process and will guarantee a continued commitment to and recognition of biomaterials and tissue engineering as viable and promising fields with a truly impactful scope.

Summary

From beginnings involved in the somewhat disparate studies of synthesizing and characterizing new biomaterials and developing new technologies and techniques for tissue engineering-based on existing materials, work in our laboratory has evolved over the last 16 years to encompass scaffold fabrication and tissue regeneration using novel biomaterials, drug, DNA, and growth factor delivery, bioreactor technology and bioactive ECM generation, and the development of animal models for the evaluation and development of tissue engineering constructs. During this time, biomaterials science has similarly expanded to include many sophisticated technologies and methods from the pure biological sciences as well as the emerging nanosciences, and advances in tissue engineering have followed progress in these fields.

Our laboratory specifically has developed tissue engineering approaches using fumarate-based materials. Working with PPF, we have developed a variety of methods towards ex vivo fabrication of porous hydrophobic scaffolds with properties suitable for bone tissue engineering. PPF, as well as OPF and P(PF-co-EG), has been used as an injectable, in situ cross-linkable material, and we continue to investigate injectable materials for tissue engineering and cell delivery. The model hydrogel OPF has been examined to elucidate the interplay between hydrophobic and hydrophilic domains and their interaction with cells. Modification of OPF to mimic naturally occurring substrates encouraged cell attachment, and cartilage tissue engineering through the incorporation of cells and growth factors in OPF has been extensively investigated. We continue studying the delivery of growth factors for bone tissue engineering, building upon the use of flow perfusion generated ECM and gene delivery strategies for the overexpression of osteogenic factors, and the impact of our early work in scaffold fabrication and bone flap generation is widely implemented in research throughout the field and in preliminary clinical applications. Finally, areas for future growth and promising advances within the field have been described, and we expect continued success for and realization of the promise of biomaterials and tissue engineering.

Acknowledgments

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Notation

ALP = alkaline phosphatase
 APS = ammonium persulfate
 BAPO = bis(2,4,6-trimethylbenzoyl) phenylphosphine oxide
 BMP-2 = bone morphogenetic protein-2
 BP = benzoyl peroxide
 β TCP = β -tricalcium phosphate
 CaP = calcium phosphate
 ECM = extracellular matrix
 FDA = Food and Drug Administration
 FGF-2 = fibroblast growth factor-2
 GRGD = Gly-Arg-Gly-Asp
 IGF-1 = insulin-like growth factor-1
 MSC = mesenchymal stem cell
 OPF = oligo(poly(ethylene glycol) fumarate)
 PEG = poly(ethylene glycol)
 PEG-DMA = poly(ethylene glycol)-dimethacrylate
 PEI = poly(ethylenimine)
 PF-DA = propylene fumarate-diacylate
 PLA = poly(D,L-lactic acid)
 PLGA = poly(D,L-lactic-co-glycolic acid)
 PLLA = poly(L-lactic acid)
 PMMA = poly(methyl methacrylate)
 P(PF-co-EG) = poly(propylene fumarate-co-ethylene glycol)
 PPF = poly(propylene fumarate)
 PTFE = polytetrafluoroethylene
 RGD = Arg-Gly-Asp
 RPE = retinal pigment epithelium
 SWNT = single-walled carbon nanotube
 TGF- β 1 = transforming growth factor- β 1
 TP508 = thrombin peptide 508
 VEGF = vascular endothelial growth factor

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