Silk based biomaterials to heal critical sized femur defects

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Abstract

Bone auto- and allografts have inherent drawbacks, therefore the treatment of non-unions and critical size defects in load bearing long bones would benefit from the use of osteopromotive biodegradable, biocompatible and mechanically durable matrices to enhance migration or delivery of cell populations and/or morphogens/cytokines. Silk fibroin biomaterial scaffolds were evaluated as osteopromotive matrices in critical sized mid-femoral segmental defects in nude rats. Four treatment groups were assessed over 8 weeks in vivo: silk scaffolds (SS) with human mesenchymal stem cells (hMSCs) that had previously been differentiated along an osteoblastic lineage in vitro (group I; pdHMSC/SS); SS with undifferentiated hMSCs (group II; udHMSC/SS); SS alone (group III; SS); and empty defects (group IV). When hMSCs were cultured in vitro in osteogenic medium for 5 weeks, bone formation was characterized with bimodal peak activities for alkaline phosphatase at 2 and 4 weeks. Calcium deposition started after 1 week and progressively increased to peak at 4 weeks, reaching cumulative levels of deposited calcium at 16 μg per mg scaffold wet weight. In vivo osteogenesis was characterized by almost bridged defects with newly formed bone after 8 weeks in group I. Significantly (P<0.01) greater bone volumes were observed with the pdHMSC/SS (group I) implants than with groups II, III or IV. These three groups failed to induce substantial new bone formation and resulted in the ingrowth of cells with fibroblast-like morphology into the defect zone. The implantation of pdHMSC/SS resulted in significantly (P<0.05) greater maximal load and torque when compared to the other treatment regimens. The pdHMSC/SS implants demonstrated osteogenic ability in vitro and capacity to thrive towards the healing of critical size femoral segmental defects in vivo. Thus, these new constructs provide an alternative protein-based biomaterial for load bearing applications.

Keywords: Human mesenchymal stem cells; Silk implant; Tissue engineering; Long bone defect; Mechanical characterization; Micro-computed tomography

Introduction

The clinical need for bone replacement is expected to increase, due in part to the substantial rise in the elderly population of the western world [1,2]. Currently, close to 1 million bone grafts are performed each year for the purpose of skeletal augmentation [3–7]. Autografts remain the clinical gold standard, driving bone repair by providing host cells, growth factors, and a template for bone regeneration. Unfortunately, donor site morbidity, cosmetic concerns, and prolonged hospitalization are just a few of the drawbacks of this procedure [8]. Similarly, the use of allografts presents risks due to immune rejection and disease transmission [9,10]. Orthopedic surgeons would therefore benefit from the
introduction of osteopromotive scaffolds that could be used to enhance migration and/or delivery of osteogenic cells and cytokines. These materials should combine essential characteristics including biocompatibility, porosity, and appropriate mechanical properties [11–14].

The use of mechanically robust 3D scaffolds offers additional options to more traditional fixation or cast methods to regenerate damaged bone tissue. A major advantage of a mechanically robust matrix is that it is less likely to undergo deformation caused by adjacent tissues (i.e., muscle) at the implant site. This deformation can be a problem, leading to the implant becoming compressed within the defect or even displaced beyond the borders of the defect site, giving rise to the potential for heterotopic bone. Several studies have shown that hMSCs can differentiate along an osteogenic lineage and form three-dimensional bone-like tissue but also detail limitations. Some scaffolds (e.g., calcium phosphate) are characterized with a slow degradation [15], while others degrade too fast [16]. Polymeric scaffolds used for bone tissue engineering, such as polyactic-co-glycolic acid) or poly-l-lactic acid, can induce inflammation due to the acidity of their hydrolysis products [6,17]. Moreover, matching mechanical properties of native bone remains an issue with most polyesters [7,18]. Therefore, there is a need to identify alternate biomaterials to overcome these limitations and meet the challenging combination of biological, mechanical, and degradation features for bone tissue engineering.

Silk-fibroin-based porous biomaterial scaffolds may contribute to this important niche in biomaterials applications. These scaffolds have low immunogenicity when properly separated engineering. of biological, mechanical, and degradation features for bone tissue and overcomes these limitations and meet the challenging combination of bone structure, mechanical, and degradation features for bone tissue engineering. 

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trypsinized and replated every 6–8 days at approximately 80% confluence. The 2nd passage (P2) of the cells was used if not otherwise stated.

hMSCs were characterized with respect to (a) the expression of surface antigens and (b) the ability to selectively differentiate into chondrogenic or osteogenic lineages in response to environmental stimuli, as follows. The expression of the following six surface antigens: CD14 (lipopolysaccharide receptor), CD31 (PECAM-1/endothelial cells), CD34 (sialomucin/hematopoietic progenitors), CD44 (hyaluronate receptor), CD71 (transferrin receptor/proliferating cells), and CD105 (endoglin), was characterized by Fluorescence Activated Cell Sorting (FACS) analysis, as in our previous studies [30,31]. Cells were detached with 0.05% (w/v) trypsin, pelleted, and resuspended at a concentration of 1 × 10^6 cells/ml. Fifty microliters of aliquots of the cell suspension was incubated (30 min on ice) with 2 μl of each of the following antibodies: anti-CD14 and anti-CD44 conjugated with fluoresceine isothiocyanate (CD14-FITC, CD44-FITC), anti-CD31 conjugated with phycoerythrin (CD31-PE), anti-CD34 and anti-CD71 conjugated with allophycocyanine (CD34-APC, CD71-APC), and anti-CD105 with a secondary rat-anti-mouse IgG-FITC antibody. Cells were washed, suspended in 100 μl of 2% formalin, and subjected to FACS analysis.
Histology and immunohistochemistry

Fixed and decalcified (12% EDTA, pH 7.0) femur explants were dehydrated in graded ethanol solutions and xylene, embedded in paraffin and cut into 3-μm-thick sections. For a general overview, sections were stained with H&E. Immunohistochemical studies were performed for bone sialoprotein (BSP) and collagen type I using a rabbit anti-bone sialoprotein II polyclonal antibody (Chemicon, Temecula, CA) and mouse anti-collagen type I monoclonal antibody (Sigma, St. Lois, MA), respectively. For collagen I antibody, sections were preincubated with 0.05% Pronase (Dako Cytomation, Baar, Switzerland) for 10 min and blocked two times 10 min with 3% H₂O₂ and with serum-free protein block (Dako Cytomation, Baar, Switzerland). The primary antibody was added to the slides at a dilution of 1:100 overnight, and after washing, an anti-mouse-peroxidase labeled polymer (EnVision+® System, K4001 Dako Cytomation, Baar, Switzerland) was applied for 30 min at room temperature. Samples were developed with an AEC Substrate Kit (Zymed Laboratories Inc., San Francisco, CA) under a light microscope. For BSP, the antibody was diluted 1:400, and the same procedure was followed except that there was no pronase pretreatment and the secondary antibody was an anti-rabbit peroxidase labeled polymer (EnVision+® System K4003 Dako Cytomation, Baar, Switzerland).

Micro-CT imaging and mechanical testing

Femoral defects were scanned using 55 kVp, 250 ms integration time, and 0.030 mm voxel side lengths (isotropic) on a micro-CT-40 system (Scanco Medical, Bassersdorf, Switzerland). Specimens were immersed in saline to homogenize the background the image and provide better contrast. The distance between the threaded pins was consistent, and this regularity aided in the scanning procedure. Micro-CT scans spanned the entire distance between the pins proximally and distally adjacent to the critical defect. After image reconstruction, volumes of interest (VOIs) were created for every specimen. VOIs were drawn contoured to the external boundary of the bone, beginning ten image slices distal to the fixator pin proximal to the defect and ending ten image slices proximal to the pin distal to the defect. VOIs were created only in regions where bone was present; in unhealed specimens, the gaps were excluded from the VOIs. Thus, morphometric analyses represented the geometry and bone morphology of the mineralized tissue in the region. Prior to morphometric analysis, a constrained Gaussian filter was used to suppress noise in the image (0.7 filter width) and images were thresholded. A previously described [33] adaptive, iterative threshold protocol was used that determined the threshold based on the characteristics of the image's histogram [38,39]. Performance tests (precision, accuracy, and stability of iterative algorithm) of this threshold approach on a variety of image types have proven it to be robust [40,41]. After thresholding, the total bone volume (BV), bone volume fraction (BV/TV), and average bone thickness (Th) were quantified. Bone volume was calculated by counting the total number of bone voxels and multiplying by their known volume. The bone volume fraction was calculated as the ratio of the bone voxels to the total number of voxels (bone and non-bone voxels after thresholding). Bone thickness was measured in three dimensions in accordance with the method established by Hildebrand and Ruegsegger [42].

Both ends of specimens were embedded in blocks of polymethylmethacrylate (PMMA) to provide gripping points for torsional testing. The specimens were kept wet during the PMMA exothermic curing process to avoid drying. Mechanical testing was performed using a torsion testing system that applies a force couple via cables attached to a pulley fixed to the top of a cylindrical bone specimen such that the axis of rotation of the pulley is coincident with the geometric neutral axis of the cylindrical bone specimen. A module containing a roller clutch, an XY sliding surface, and coupling cables was mounted to an MTS200 screw axis load frame (MTS, Saint Paul, MN, USA) for the application of torque. The module maintains the prescribed torque on the specimen when it is removed from the load frame. Torsional testing was performed using a strain rate of 5 rad/min, and the data were collected with a LabVIEW (National Instruments, Austin, TX) acquisition system. Maximum torsional displacement (rad) and load (N) and maximum torque (Nm) were measured for each specimen. Additionally, torsional stiffness (N.m.rad) was calculated from the slope fit to the linear portion of the torsional load displacement curve.

Statistical analysis

For statistical significance, samples were evaluated using a Student’s t test as well as ANOVA where appropriate. ANOVA was followed by a post hoc assessment.

Fig. 1. Characterization of human mesenchymal stem cells. (A) Cells had a fibroblast-like morphology and presented the (B) CD105 and (C) CD71, but not the CD34 antigen on their surfaces, respectively. Pellet culture in chondrogenic medium (D) resulted in the deposition of an intensively red staining matrix with safranin O. Pellets cultured in osteogenic medium and stained with von Kossa displayed black staining. Magnifications (A) 40× and 100× (inset), (D, E) scale bar length = 100 μm. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)
using the Tukey HSD method. Differences were considered significant when equal or less than $P=0.05$.

**Results**

**Cell characteristics and bioreactor culture**

The cells were adherent to tissue culture plastic after about 6 h and showed a fibroblast-like morphology (Fig. 1A). Their surface antigen pattern was CD105$^+$ (Fig. 1B), CD71$^+$, and CD34$^-$ (Fig. 1C) and CD44$, CD14^-$, and CD31$^-$ (data not shown). After 5 weeks of pellet culture in chondrogenic medium, the cells deposited an intensely red staining matrix with safranin O (Fig. 1D), indicating deposition of extracellular glycosaminoglycans. The culture in osteogenic medium resulted in the formation of a black staining matrix in a von Kossa staining (Fig. 1E) noting for mineralized extracellular matrix. In control medium, no positive safranin O or von Kossa staining was observed (data not shown).

When cells were seeded on silk fibroin scaffolds (SS) and cultured in spinner flask bioreactors with osteogenic medium for 5 weeks, alkaline phosphatase activity followed an undulating pattern, with significantly higher activity after 2 and 4 weeks, when compared to weeks 1 and 3 (Fig. 2A; $P<0.05$ or $<0.01$). Mineralization determined by calcium assay started between week 1 and 2 ($P<0.01$) and continued to increase between week 3 and 4 ($P<0.01$). No increase in mineralization was observed between week 4 and 5 (Fig. 2B).

**In vivo studies**

The implantation of pdHMSC/SS (group I, Table 1) into critical size femoral defects resulted in good integration of the implant with the fracture ends at 8 weeks post-surgery. Substantial new bone formation (25–40%) was observed, with more cortical bone in the outer implant zones and more interconnected, woven bone in the centers, as shown with H&E staining (Fig. 3A). Positive staining for bone sialoprotein (BSP) and collagen type I (data not shown) was found in the newly formed matrix of actively bone-forming sites. Aside from long cells with fibroblastic morphology and stranded connective tissue, the formation of new woven bone was observed (Fig. 3A, insert). No apparent signs of silk scaffold degradation were observed, as indicated by the smooth lattice surfaces, although foreign body cells were present (Fig. 3A).

The implantation of udHMSC/SS (group II, Table 1) resulted in the formation of a fibroblast tissue throughout the implantation zone (Fig. 3B), with some few capillaries randomly dispersed through the implant’s lattice (Fig. 3B, insert). Bone formation was mainly observed at the defect margins with the parent bone but not at the center of the defect. Moderate positive staining for BSP and type I collagen was mainly found at the interface between the defect border and the newly formed bone as well as few and randomly clusters distributed throughout the matrix (data not shown). A higher number of foreign body cells than in group I were present in the defect zone, but the SS seemed well embedded in the cell- and extracellular-matrix-rich environment.

Defects treated with unseeded scaffolds SS (group III, Table 1) developed a dense cellular network in the vicinity of the defect borders and appositional to the scaffold lattice (Fig. 3C). Substantially fewer cells were found in the defect center, and scaffold interaction with the host tissue was weaker as observed in groups I and II and as indicated by lower cell numbers and void spaces at the host–implant interface (Fig. 3C). Immunohistochemistry showed positive staining for both BSP and type I collagen within the matrix and cells with a fibroblastic morphology within the scaffold center, whereas loosely connected cells with a connective-tissue-like morphology stained positive only for type I collagen but not BSP. The small amounts of new woven bone stained positive only for BSP.

Tissue formation in the empty defects (group IV, Table 1) was confined to the defect margins with parent bone, with some clusters of non-bone-forming cells found in more distal regions (Fig. 3D and insert). These clusters did stain only faintly for BSP and positive for type I collagen, respectively.

The morphology of the formed bone was evaluated using μCT (Fig. 4). In general, defects treated with pdHMSC/SS (group I) were spanned by cortical-like bone in the outer zones and a cancellous-like but irregularly and loosely connected bone tissue within the defect center. The defects in three of the six rats were completely bridged with no apparent cracks, two by small
in the center of the implant and along the line of the stacked implant discs and one by large cracks spanning throughout the defect zone. The implantation of udHMSC/SS (group II) resulted in bone formation at the defect margins with parent bone and some minor formation within the defect center, however, no bridging of the defect was observed. Finally, no bone formation was apparent for defects treated with unseeded SS (group III) or empty control defects (group IV).

Morphometric analysis based on the tomographic evaluations quantified the qualitative observations previously detailed. Significantly more bone volume was observed in group I as compared to all other treatment groups \((P<0.01; \text{Fig. 5A})\). The ratio of bone volume to total volume, as an indicator of the relative amount of formed bone, was significantly higher for group-I-treated defects when compared to all other groups \((P<0.01; \text{Fig. 5B})\). The thickness of the bone formed was consistent among groups, with apparently high standard deviations (data not shown).

The defects were mechanically characterized (Fig. 6). The maximal load before failure was significantly higher for group I when compared to groups II \((P<0.05)\) and III \((P<0.01)\). The implantation of udHMSC/SS (group II) resulted in significantly higher maximal load levels when compared to group III \((P<0.01)\). Similarly, the maximal torque was higher for pdHMSC/SS (group I) when compared to groups II \((P<0.05)\) and III \((P<0.01)\), respectively. Maximal torque was significantly higher for group II when compared to the SS (group III) \((P<0.01)\). The total stiffness of the defects was higher for the pdHMSC/SS (group I) when compared to group III \((P<0.01)\), but no significant differences were observed between group I and II or between group II and III.

Discussion

The goal of the present study was to evaluate a new protein-based biomaterial scaffold system with mechanically robust and durable properties upon implantation into long bone and critically sized femoral defects. A tissue-engineered trabecular bone-like tissue was generated from pdHMSC/SS (group I, Table 1) which were pre-cultured in osteogenic medium for 5 weeks. These implants were inserted into the defect site, and osteogenesis was compared to defects treated with udHMSC seeded silk scaffolds, those not pre-cultured in osteogenic medium (group II, Table 1), unseeded silk scaffolds (group III, Table 1), and untreated control defects (empty, group IV, Table 1).

Silk fibroin, the structural, hydrophobic, and approximately 370 kDa protein isolated from the cocoon of *B. mori* [43–46] that has emerged as a novel substrate for engineering biomaterials, has proven to be biocompatible [22,33], offers distinguishing mechanical properties when compared to other naturally derived or degradable polymeric biomaterials [47], and exhibits low inflammatory response [21,22].

Stem cells are believed to be one of the most interesting cell sources for tissue engineering, particularly for the (re)generation
of bone tissue. This is due to characteristics inherent in stem cells such as (i) a high proliferation capacity, (ii) ability for self-renewal, (iii) multilineage differentiation and, therefore, potential for the repair of various tissues [48–52]. The cells used in this study were characterized as stem cells based on their surface markers as well as the multipotent and selective differentiation outcomes shown to bone or cartilage. Silk fibroin scaffolds seeded with hMSCs and cultured in osteogenic medium in spinner flask bioreactors generated bone-like biochemical responses as typically observed for mesenchymal cells on other biomaterials [31,32,53]. The high degree of mineralization observed, beginning between weeks 1 and 2 post-seeding, is evidence of the strong osteogenic commitment of the hMSCs on silk scaffolds and in the presence of BMP-2 (one of the osteogenic media components).

Substantial differences were observed in vivo in bone healing between pdHMSC/SS (group I), udHMSC/SS (group II) and the SS (group III). The implantation of pdHMSC/SS into the defect resulted in substantial and homogenous bridging of the majority of defects (Figs. 3, 4), significantly \(P<0.01\) higher bone volumes and bone volumes per trabecular volume (which indicates a shift from trabecular to cortical bone), and a substantial increase in mechanical properties when compared to the other groups. The premineralization of the silk scaffolds appears to be important in
overall bone restoration in vivo. Thus, future studies should also address the relative contribution of mineralization on silk fibroin scaffolds related to osteoconductivity with and without the presence of hMSCs during implantation. Although bone formation within the defect zone was restricted to the defect margins, the udHMSC seeded scaffolds (group II) resulted in significantly ($P < 0.01$) higher maximal load and torque when compared to unseeded SS (group III). This may be due to the formation of a dense cellular- and matrix-rich network supported by the silk scaffolds between the host bone and the scaffold center, fixing the stacked scaffold disks in place and thereby resulting in a more stabilized mechanical environment. However, the mechanical data were presented to provide a relative strength profile across study groups. This must be noted as the femurs were fixed before measurements, instead of most biomechanical studies. We have not tested if the fixation altered the mechanical properties of bone. Therefore, one should be careful to compare the presented data to normal unfixed rat bones since only the relative strength of specimens across different groups is of consequence in this study. As part of the morphometric analysis of the engineered bone-like tissue, we evaluated the bone thickness using an algorithm validated for trabecular bone (data not shown). However, the repair side consisted of complex, inhomogenous structures other than trabecular bone (Fig. 3). The validation for the algorithm used was not validated for the type of tissue formed in the defect site. Consistently, our data showed a large variation. The validation of algorithms provided should be carefully evaluated for the intended application.

The gold standard of bone replacement is autograft bone or bone that has been harvested from another anatomic location in the same individual and then inserted into the fracture zone [8,54]. The advantage of this approach is the combined transplantation of autologous cells, scaffold material, and growth factors stored within the matrix [55,56]. However, harvesting autologous bone is associated with second site morbidity and hampered by insufficient amounts of collected bone substance to fill the entire defect [57]. Apart from other options such as the use of allograft bone, which is associated with the risk of disease transmission or immune rejection, metals and ceramics are used to treat larger defects [58]. Metals provide instant mechanical stability but are associated with poor overall integration and can fail because of...
infection or due to fatigue loading [7]. Ceramics generally have very low tensile strength, are brittle, and cannot be used in locations of significant torsion such as long bones [3,4,59–61]. Alternatively, biomaterials such as type I collagen base implants are frequently used but have limitations related to rapid biodegradation or – when stabilized through cross-linking – foreign body responses and uncontrolled calcification [62,63]. The combined use of an organic template based on silk fibroin and hMSCs predifferentiated in bioreactors with osteogenic medium (group I) provides an alternative route for future treatment scenarios of larger femoral defects. In this study, the filling of the defect was achieved by stacking five silk scaffolds, each of which was individually engineered, to result in good bone integration. In vivo bone formation resulted in a good formation of bone between the stacked disks and the fracture ends, respectively. However, the engineering of a pdHMSC/SS using a single block of silk instead of staggered disks should further improve the mechanical properties, integrity, and safety. However, this is currently challenging due to limitations in nutrient and gas transport during predifferentiation in vitro. Bone is a highly vascularized and metabolically active tissue and especially prone to low oxygen tensions, quickly resulting in necrotic responses. These limitations curtail current tissue engineering approaches, resulting in bone thicknesses limited to approximately 500 μm to 1 mm [32,64]. Future studies aiming at increased mass and gas transport in the scaffolds when cultured in vitro are important and could be addressed through novel designs for the bioreactors, scaffolds, and medium used in the process (e.g., by addition of oxygen binding molecules mimicking the function of hemoglobin [64]).

Tissue engineering can provide adequate bone replacement resulting in full recovery of the patient [65]. The results of the present study indicate that the implantation of tissue-engineered bone on silk scaffolds was suitable for this application because it leads the way to bridging of critical size defects with newly formed functional bone. This was also reflected in mechanical properties, with group I exceeding all other groups in terms of maximal load before failure, maximal torque, and torsional stiffness. Although bone formation was limited in groups II and III, some mechanical integrity was achieved due to the presence of the silk and some fixation due to the invasion of fibrous tissues into the implant materials. The osteoinduction achieved in group I rats resulted in complete bridging of femoral, segmental critically sized defects with a callus on the outside and around 30% newly formed woven bone tissue inside the defect. These findings encourage the continued investigation of potential use of silks as a mechanically stable and robust scaffold for tissue-engineered constructs and load bearing bones. Future studies should be conducted to validate the effectiveness of this procedure in larger animals.

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