Published in final edited form as:

Acta Biomater. 2010 June ; 6(6): 2329–2334. doi:10.1016/j.actbio.2009.11.021.

Direct Laser Processing of Tantalum Coating on Titanium for Bone Replacement Structures

Vamsi Krishna Balla, Shashwat Banerjee, Susmita Bose, and Amit Bandyopadhyay*

W. M. Keck Biomedical Materials Research Laboratory School of Mechanical and Materials Engineering, Washington State University Pullman, WA 99164-2920, USA.

Abstract

Recently, tantalum is gaining more attention as a new metallic biomaterial as it has been shown to be bioactive and biologically bond to the bone. However, relatively high cost of manufacture and inability to produce a modular all tantalum implant has limited its widespread acceptance. In this study, we have successfully deposited Ta coating on Ti using Laser Engineered Net Shaping (LENSTM) to enhance osseointegration properties. *In vitro* biocompatibility study, using human osteoblast cell line hFOB, showed excellent cellular adherence and growth with abundant extracellular matrix formation on Ta coating surface compare to Ti surface. Six times higher living cell density was observed on Ta coating than on Ti control surface during MMT assay. High surface energy and wettability of Ta surface were observed to contribute to its significantly better cell-materials interactions. Also, these dense Ta coatings do not suffer from low fatigue resistance due to the absence of porosity and sharp interface between the coating and the substrate, which is a major concern for porous coatings used for enhanced/early biological fixation.

Keywords

Laser processing; Tantalum; Coatings; Biocompatibility; Osteoblast; Bone implants; laser engineered net shaping (LENS)

1. Introduction

Great effort has been made over the last few decades in search for a suitable metallic biomaterial for orthopaedic applications. Titanium and its alloys are most widely used for load bearing metal implants due to their excellent biocompatibility derived from thin oxide surface layer. However, Ti and its alloys lack in sufficient osseointegration required for implant longevity. One consideration to improve bone-implant contact and healing process has been the use of bioactive calcium phosphate ceramics on metal implants. Unfortunately, cracking, delamination and low fracture toughness of ceramic coatings are big hurdles for its widespread application. In the last two decades, a variety of porous coatings and materials have also been used to obtain biological fixation of bone with metal implants in total hip and knee prostheses. The most common metallic materials include Ti and CoCrMo alloy sintered beads, diffusion bonded Ti, fiber metal, and Ti plasma spray [1-7]. Clinical and histological evidence from

© 2009 Acta Materialia Inc. Published by Elsevier Ltd. All rights reserved.

* Corresponding Author: Fax: 509-335-4662; amitband@wsu.edu.

Publisher's Disclaimer: This is a PDF file of an unedited manuscript that has been accepted for publication. As a service to our customers we are providing this early version of the manuscript. The manuscript will undergo copyediting, typesetting, and review of the resulting proof before it is published in its final citable form. Please note that during the production process errors may be discovered which could affect the content, and all legal disclaimers that apply to the journal pertain.

retrieved implants clearly demonstrates that these porous surfaces enhance bone tissue ingrowth and are effective in supplementing the stability of the implant by biological fixation [8-13]. These porous materials are best suited for use as coatings, since they do not readily have the required mechanical properties that would allow them to be used as bulk structural materials for implants, bone augmentation, or substitutes for bone graft. Moreover, these porous coated Ti alloy implants show 50 to 75% lower fatigue strength in comparison to their equivalent fully dense materials [14-16].

Recently, tantalum is gaining more attention as a new metallic biomaterial. Tantalum has been shown to be corrosion resistant [17] and bioactive *in vivo* [18]. Through the formation of a bonelike apatite layer in simulated body fluid, tantalum has been shown to be bioactive and biologically bond to the bone. In several *in vitro* and animal studies, the porous Ta metal has provided a scaffold for bone ingrowth and mechanical attachment [19,20-22]. These porous Ta components offer a low modulus of elasticity, high surface frictional characteristics, and excellent osseointegration properties (i.e., bioactivity, biocompatibility, and in-growth properties) [23,24]. Trabecular Metal™ is one such porous Ta marketed by Zimmer Inc. (Warsaw, IN, USA). It is being made from the pyrolysis of thermosetting polymer foam creating a low-density vitreous carbon skeleton. Then commercially pure Ta is deposited onto this interconnected scaffold using chemical vapour deposition/infiltration [19]. Relatively high cost of manufacture and inability to produce a modular all tantalum implant have limited its widespread acceptance. Moreover, one major limitation of these porous scaffolds is their significantly lower fatigue resistance in comparison to their equivalent fully dense materials. Besides its high affinity towards oxygen, extremely high melting temperature of Ta (3017°C) makes it difficult to process fully dense Ta implant structures via conventional processing routes. In this work, we have deposited Ta coating on Ti using high power lasers and Laser Engineered Net Shaping (LENS™) to enhance osseointegration properties. Such dense Ta coatings on Ti can potentially enhance cell-materials interactions of inherently bioinert Ti without sacrificing its excellent fatigue resistance. In this article, we report laser processing of dense Ta coatings and how surface energy of Ta enhanced the *in vitro* biocompatibility of Ti with Ta coatings. In addition, the specific advantages of using laser processing in conjunction with LENS™ are (i) ability to control melting of high melting point Ta thus creating fully dense or tailored porosity implants, (ii) net-shape fabrication of complex shaped implants, (iii) retention of high purity of the initial feed stock power in the final product, (iv) fine and uniform microstructural features due to localized melting and subsequent solidification improving mechanical, physical and biological properties. In recent years, LENS™ has been extensively used to create fully dense, porous, compositionally and structurally graded materials for load bearing metal implants [25-34].

2. Materials and Methods

2.1 Ta Coating Fabrication and Characterization

Ta metal powder (Grandview Materials Inc., Columbus, OH) with 99.5% purity and particles size between 45 and 75 μm was used. Ta coatings with 1.5 – 2.0 mm thickness having 10 mm diameter were deposited on a substrate of 3 mm thick rolled, commercially pure (CP) Ti plates using LENS™-750 (Optomec Inc. Albuquerque, NM) equipped with a 500W continuous wave Nd:YAG laser. Detailed description of LENS™ operation and capabilities can be found elsewhere [25,29]. All the coatings were fabricated in a glove box containing argon atmosphere with O_2 content less than 10 ppm to limit oxidation of Ta and Ti during processing. After process optimization, a laser power of 450 W, a scan speed of 7 mm s^{-1} and a powder feed rate of 106 g min^{-1} was used to deposit Ta metal on a Ti substrate. Under these processing conditions the deposited Ta coatings found to be visually sound with strong adherence to the substrate. Cross sectional microstructures of the samples were examined using light

microscope and Field Emission Scanning Electron Microscope (FE-SEM) (FEI – Quanta 200F). To reveal microstructural features, polished Ta coating samples were further chemical polished for 30 sec using a solution containing 25 ml lactic acid, 15 ml HNO₃ and 5 ml HF. The chemical polished samples were then etched with 20 ml HNO₃, 20 ml HF and 60 ml H₂SO₄ solution for 60 sec. Vickers microhardness measurements were made on the Ta coatings using a 100 g load for 15s, and an average value of 10 measurements is reported. In order to track the diffusion that may occur at the interface between the Ta coating and the Ti substrate, a series of microhardness indentations were placed from one end of the deposit to the other end, with neighboring indents being separated by 0.05 mm, using a Vickers microhardness tester at 100 g load applied for 15s.

2.2 Contact Angles and Surface Energy

Ta coating and Ti substrate surfaces for contact angle measurements were ground using series of SiC grinding papers with various sizes up to 1200 grit. Samples were polished on velvet cloth using series of Al₂O₃ powder up to 0.05 µm suspended in distilled water. Finally, just before testing, all the samples were ultrasonically cleaned in alcohol bath. This procedure ensured identical surface roughness and topography on all sample surfaces. Contact angles were measured using the sessile drop method with a face contact angle setup equipped with a microscope and a camera. A 1.0 µl droplet of distilled water was suspended from the tip of the microliter syringe. The sample surface was advanced toward the syringe tip until the droplets made contact with the sample surface. Images were collected with the camera and the contact angle between the drop and the substrate was measured from the magnified image. All the measurements were carried out at 21° C with a relative humidity of 21%. Contact angles were also measured using DMEM (Dubco Modified Eagle Medium) cell media. For each sample and liquid combination, an average of 8 measurements is reported. Surface energy of Ta coating and Ti control sample surfaces was calculated based on contact angle measurements, which were conducted with liquids with polar and non-polar characteristics: distilled water and diiodomethane. In this study, Fowkes' equation was used to calculate the energy:

$$\gamma_L (1 + \cos \theta) = 2 \left(\sqrt{\gamma_L^d \gamma_s^d} + \sqrt{\gamma_L^p \gamma_s^p} \right) \quad (1)$$

The reported average surface energy of Ti and Ta was calculated from 10-15 measurements on each sample.

2.3 Cell Culture and MTT Assay

The sample surfaces for cell culture and MTT assay were also polished smooth, following the procedure adopted for contact angle measurements, to ensure identical surface roughness on Ta coating and the Ti substrate. All samples were sterilized by autoclaving at 121°C for 20 min. In this study established human osteoblast cell line hFOB 1.19 (ATCC, Manassas, VA) was used. Cells were seeded onto the samples placed in 24-well plates. The base medium for this cell line was a 1:1 mixture of Ham's F12 Medium and Dulbecco's Modified Eagle's Medium (DMEM/F12, Sigma, St. Louis, MO), with 2.5 mM L-glutamine (without phenol red). The medium was supplemented with 10% fetal bovine serum (HyClone, Logan, UT) and 0.3 mg/ml G418 (Sigma, St. Louis, MO). Cultures were maintained at 34°C under an atmosphere of 5% CO₂. The culture medium was changed every 2-3 days for the duration of the experiment. Samples for testing were removed from culture at 3, 7 and 14 days of incubation. All samples for SEM observation were fixed with 2% paraformaldehyde/2% glutaraldehyde in 0.1M cacodylate buffer overnight at 4°C. Post-fixation was performed with 2% osmium tetroxide (OsO₄) for 2 h at room temperature. The fixed samples were then dehydrated in an ethanol series (30%, 50%, 70%, 95% and 100% three times), followed by a hexamethyldisilane

(HMDS) drying procedure. After gold coating, the samples were observed under FE-SEM for cell morphologies. MTT (3-(4,5-dimethylthiazol-2-yl)-2,5-diphenyl tetrazolium bromide) assay was used to evaluate cell proliferation. The MTT (Sigma, St. Louis, MO) solution of 5 mg/ml was prepared by dissolving MTT in PBS, and was filtered using a filter paper of 0.2 micron pore. 500 μ l of diluted MTT solution (50 μ l into 450 μ l in DMEM/F12 medium) was then added to each sample in 24-well plates. After 2 h of incubation, 500 μ l of solubilization solution made up of 10% Triton X-100, 0.1N HCl and isopropanol was added to dissolve the formazan crystals. 100 μ l of the resulting supernatant was transferred into a 96-well plate, and read by a plate reader at 570 nm.

Statistical analysis was performed using Student's t-test, and $p < 0.05$ was considered statistically significant. Triplicate samples were used in MTT assay experiments to insure reproducibility. Data are presented as mean $\pm$ standard deviation.

3. Results and Discussion

3.1 Microstructures and Hardness

Complete melting of Ta powder has been achieved at 450 W laser power with substrate moving at a scan speed of 7 mm s⁻¹ and at a powder feed rate of 106 g min⁻¹. As shown in Figure 1a, the Ta coatings exhibited good bonding between individual layers without any cracks or lack fusion defects indicating that the laser energy input was sufficient for complete melting of Ta powder and strong bonding with the Ti substrate. Figure 1b shows that the interface between the Ta coating and the Ti substrate was also sound and defect free. However, occasional unmelted/partially melted Ta powder and small amount of residual porosity was observed in these coatings. The lack of complete melting of Ta powder, shown with arrows in Figure 1a, at some regions of the coatings appears to be a direct consequence of effect of the laser absorption coefficient of Ta. Laser absorptivity of a material controls directly the energy transfer from the laser to the sample. The laser absorption coefficient of a material is directly proportional to its electrical resistance and inversely proportional to the wavelength of the laser [35]. Relatively low electrical resistivity of Ta ($13 \times 10^{-8} \Omega \text{ m}$) than Ti ($40 \times 10^{-8} \Omega \text{ m}$) decreases its laser absorption coefficient leading to incomplete melting of Ta powder, especially the coarse fraction of Ta powders used in this study. The unmelted powder can be easily eliminated either by increasing the laser energy input or by using a finer Ta powder. Presence of small, occasional residual porosity in Ta coatings is presumably due to boiling of Ti, as the boiling point of Ti (3287°C) is close to the melting point of Ta (3017°C). Porosity can also arise due to the presence of residual gases in the feed stock powder.

The grain size of laser processed Ta coating is shown in Figure 1c. In general, the solidification structures in LENSTM processed materials are fine due to high cooling rates associated with this process. However, in the present work the grain size of Ta coating, measured using linear intercept method, was found to be $70 \pm 20 \mu\text{m}$. This is due to high laser energy input used to deposit Ta coatings on Ti, which results in lower cooling rates due to shallow thermal gradients during deposition. The average hardness of laser deposited Ta coatings was $392 \pm 37 \text{ Hv}$, which is higher than the average hardness of Ti substrate $189 \pm 4 \text{ Hv}$. The large variation in hardness of Ta coating is due to variations observed in grain size across the coating. The hardness profile across the coating, shown in Figure 2, indicate that the hardness of Ta increased gradually from 250 HV at the interface to 400 Hv in the regions just below the top surface. The high hardness of Ti (263 HV) just below the coating interface is due to the solid solution strengthening as a result of Ta dissolution in Ti. The smooth nature of the hardness profile, without an abrupt increase in the hardness, on each side of the interface indicates the absence of undesirable phase formation due to the reaction between Ta and Ti. These results clearly demonstrate that high power lasers produce fully dense coatings and can be extended to fabricate net shape Ta structures when used in conjunction with LENSTM - a rapid manufacturing technology.

3.2 In vitro Biocompatibility

In vitro biocompatibility study using established human osteoblast cell line hFOB (CRL-11372) was aimed to evaluate any possible toxic effect of laser processing of Ta coating on Ti compared to that of widely used Ti, as a control sample. Initial attachment, growth and spreading of hFOB cells on Ti control and Ta coating surfaces were analyzed using field emission scanning electron microscope (FE-SEM). Figure 3 shows the morphology of hFOB cells on Ti control and Ta coatings after 3 and 14 days of culture. Ti surfaces showed less cell distribution even after 14 days of culture. Figure 3a and c shows that the cells on Ti surface exhibit few filopodia extensions indicating that they largely failed to attach and spread on the control surface. Cells on Ta coatings appeared cuboidal and three-dimensional morphology with more filopodia extensions, as shown in Figure 3b, suggesting significantly better cell attachment and spreading. After 14 days of culture, cells on Ta coatings become elongated forming a dense and confluent cellular layer. The cells on Ta coating surface generated an abundant amount of the extracellular matrix (ECM), which can be interpreted as an early stage of osteoblast differentiation. In contrast, Ti control surface showed no confluent layer formation even after 14 days.

The number of living cells determining the hFOB cell proliferation on laser processed Ta coating and Ti control surfaces was determined by MTT assay. The amount of cells on the samples as determined by MTT assay is shown in Figure 4. Significantly higher cell density was observed on Ta coating surface compared to that observed on Ti control for all culture durations. Cells proliferated rapidly on the Ta coating surface than on Ti surface. With the increase in culture time, differences in cell density increased and almost six times higher cell density was observed on Ta coating than on Ti control surface after 14 days of culture. This is an important finding as none of the currently used metallic biomaterial shows such a high number of living cell density during *in vitro* biocompatibility evaluation. The osteoblast cell attachment to the sample surface and its growth takes place using filopodia and lamelliopodia structures. After attachment, cells grow continually and start differentiating by ECM mineralization process and finally remodel as natural bone. More filopodia extensions of cell morphology on Ta coatings indicate its favorable surface characteristics for cell attachment, especially in the early stages of cell culture. Significantly better cell material interactions of Ta surface have been confirmed by the large number of living cells observed in the MTT assay. The cell attachment and growth are primarily associated with the material's surface characteristics such as wettability and surface energy. The Ta surface with high wettability and surface energy would have enhanced the cell-materials interactions. To confirm this hypothesis, contact angles were measured using cell media and distilled water on the Ta coating and Ti surfaces. The experimental results shown in Table 1 indicate significantly lower contact angles for Ta coating surface than for Ti surface improving the wettability of Ta surface. In general, a contact angle higher than 20° indicates hydrophobic nature of the surface. Since the cell media are water based, the cellular attachment will be poor in any hydrophobic surface. Also, a low contact angle of Ta surface suggests its high surface energy, which is another factor that can contribute to its better cell attachment. Table 1 also shows the surface energies of Ta coating and Ti substrate surfaces calculated using contact angles measured with two different well characterized liquids. Ta coating surface showed 29% higher surface energy of 55 mN m⁻¹ than the surface energy observed on Ti, which was 42 mN m⁻¹. The surface energy results corroborate the observations on contact angles of cell media and distilled water on these sample surfaces. High level of cell attachment on moderately hydrophilic surfaces, such as Ta in the present work, has also been observed [36]. A high contact angle and low surface energy of Ti surface with hydrophobic nature resulted in poor cell attachment on its surface. These results clearly indicate that surface energy plays an important role in early stage bone cell-materials interactions. Improved wettability and high surface energy lead to more cell attachment and

proliferation on Ta surface than with the control Ti surface. Similar results on better cell attachment on surfaces with high surface energy have also been reported [37-39].

Our present results indicate that dense Ta coating on Ti can provide a top surface with significantly better cell-materials interactions, due to its high wettability and surface energy, than that of Ti. Also, these dense Ta coatings do not suffer from low fatigue resistance due to the absence of porosity and sharp interface between the coating and the substrate, which is a major concern for porous coatings used for enhanced/early biological fixation. The inherent advantage of using LENSTM to fabricate Ta coatings is its ability to control melting of high melting point Ta creating fully dense or tailored porosity implants, which is impossible to process via conventional routes. The process can also be used to create modular full Ta implants with or without porosity.

4. Conclusions

The deposition of dense tantalum coatings on titanium has been successfully demonstrated using LENSTM. The process can also be used to create modular full Ta implants with or without porosity. *In vitro* biocompatibility study showed significantly better hFOB cell-materials interactions on laser processed Ta coating surface than on Ti surface indicating its potential for enhanced/early biological fixation. The enhanced cell attachment and proliferation on Ta surface are direct consequence of its high wettability and surface energy.

Acknowledgments

Authors would like to acknowledge the financial support from the W. M. Keck Foundation to establish a Biomedical Materials Research Lab at WSU. Authors also acknowledge the financial support from the Office of Naval Research (Grant number N00014-1-05-0583).

References

1. Bobyn JD, Pilliar RM, Cameron HU, Weatherly GC. The optimum pore size for the fixation of porous-surfaced metal implants by the ingrowth of bone. *Clin Orthop* 1980;150:263–70. [PubMed: 7428231]
2. Cameron HU, Pilliar RM, Macnab I. The rate of bone ingrowth into porous metal. *J Biomed Mater Res* 1976;10:295–302. [PubMed: 1254617]
3. Galante J, Rostoker W, Lueck R, Ray RD. Sintered fiber metal composites as a basis for attachment of implants to bone. *J Bone Joint Surg [Am]* 1971;53-A:101–14.
4. Pilliar RM, Cameron HU, Macnab I. Porous surface layered prosthetic devices. *J Biomed Eng* 1975;10:126–31.
5. Pilliar RM. Powder metal-made orthopaedic implants with porous surface for fixation by tissue ingrowth. *Clin Orthop* 1983;176:42–51. [PubMed: 6851341]
6. Spector, M. Bone ingrowth into porous metals. In: Williams, DF., editor. *Biocompatibility of orthopaedic implants*. CRC Press; Florida: 1982. p. 89-128.
7. Stein T, Armand C, Bobyn JD, Krygier JJ, Miller J, Brooks CE. Quantitative histological comparison of bone growth into titanium and cobalt-chromium porous coated canine implants. *Orthop Trans* 1991;15:178.
8. Collier JP, Mayor MB, Chae JC JC, Surprenant VA, Surprenant HP, Dauphinais LA. Macroscopic and microscopic evidence of prosthetic fixation with porous-coated materials. *Clin Orthop* 1988;235:173–180. [PubMed: 3416524]
9. Cook SD, Barrack RL, Thomas KA, Haddad RJ. Quantitative analysis of tissue growth into human porous total hip components. *J Arthroplasty* 1988;3:249–262. [PubMed: 3183679]
10. Cook SD, Barrack RL, Thomas KA, Haddad RJ. Tissue growth into porous primary and revision femoral stems. *J Arthroplasty* 1991;6:S37–46. [PubMed: 1774569]

11. Engh CA, Zettl-Schaffer KF, Kukita Y, Sweet D, Jasty M, Bragdon C. Histological and radiographic assessment of well functioning porous-coated acetabular components: a human postmortem retrieval study. *J Bone Joint Surg [Am]* 1993;75-A:814–824.
12. Friedman RJ, Black J, Galante JO, Jacobs JJ, Skinner HB. Current concepts in orthopaedic biomaterials and implant fixation. *J Bone Joint Surg [Am]* 1993;75-A:1086–1109.
13. Pidhorz LE, Urban RM, Jacobs JJ, Sumner DR, Galante JO. A quantitative study of bone and soft tissues in cementless porous-coated acetabular components retrieved at autopsy. *J Arthroplasty* 1993;8:213–225. [PubMed: 8478636]
14. Kohn DH, Ducheyne P. A parametric study of the factors affecting the fatigue strength of porous coated Ti-6Al-4V implant alloy. *J Biomed Mater Res* 1990;24:1483–1501. [PubMed: 2279982]
15. Cook SD, Georgette FS, Skinner HB, Haddad RJ Jr. Fatigue properties of carbon-and porous-coated Ti-6Al-4V alloy. *J Biomed Mater Res* 1984;18:497–512. [PubMed: 6736080]
16. Yue S, Pilliar RM, Weatherly GC. The fatigue strength of porous-coated Ti-6% Al-4% V implant alloy. *J Biomed Mater Res* 1984;18:1043–1058. [PubMed: 6544792]
17. Bermudez MD, Carrion FJ, Martinez-Nicolas G, Lopez R. Erosion-corrosion of stainless steels, titanium, tantalum and zirconium. *Wear* 2005;258:693–700.
18. Kato H, Nakamura T, Nishiguchi S, Matsusue Y, Kobayashi M, Miyazaki T, Kim H, Kokubo T. Bonding of alkali- and heat-treated tantalum implants to bone. *J Biomed Mater Res* 2000;53:28–35. [PubMed: 10634949]
19. Bobyn JD, Stackpool GJ, Hacking SA, Tanzer M, Krygier JJ. Characteristics of bone ingrowth and interface mechanics of a new porous tantalum biomaterial. *J Bone Joint Surg Br* 1999;81-B:907–914. [PubMed: 10530861]
20. Hacking SA, Bobyn JD, Toh K, Tanzer M, Krygier JJ. Fibrous tissue ingrowth and attachment to porous tantalum. *J Biomed Mater Res* 2000;52:631–638. [PubMed: 11033545]
21. Bobyn JD, Toh KK, Hacking SA, Tanzer M, Krygier JJ. Tissue response to porous tantalum acetabular cups: a canine model. *J Arthroplasty* 1999;14:347–354. [PubMed: 10220190]
22. Matsuno H, Yokoyama A, Watari F, Motohiro U, Kawasaki T. Biocompatibility and osteogenesis of refractory metal implants, titanium, hafnium, niobium, tantalum and rhenium. *Biomaterials* 2001;22:1253–1262. [PubMed: 11336297]
23. Miyazaki T, Kim HM, Kokubo T, Ohtsuki C, Kato H, Nakamura T. Mechanism of bonelike apatite formation on bioactive tantalum metal in a simulated body fluid. *Biomaterials* 2002;23:827–832. [PubMed: 11771702]
24. Kokubo T, Kim HM, Kawashita M. Novel bioactive materials with different mechanical properties. *Biomaterials* 2003;24:2161–2175. [PubMed: 12699652]
25. Vamsi Krishna B, Bose Susmita, Bandyopadhyay Amit. Fabrication of Porous NiTi Shape Memory Alloy Samples using Laser Engineered Net Shaping. *J Biomed Mater Res B* 2009;89-B:481–490.
26. Balla VK, DeVasConCellos Paul, Xue Weichang, Bose Susmita, Bandyopadhyay Amit. Fabrication of Compositionally and Structurally Graded Ti-TiO₂ Structures using Laser Engineered Net Shaping (LENS). *Acta Biomater* 2009;5:1831–1837. [PubMed: 19233752]
27. Vamsi Krishna B, Xue Weichang, Bose Susmita, Bandyopadhyay Amit. Engineered Porous Metals for Implants. *JOM* 2008;60(5):45–48.
28. Bandyopadhyay, Amit; Vamsi Krishna, B.; Xue, Weichang; Bose, Susmita. Application of Laser Engineered Net Shaping (LENS) to Manufacture Porous and Functionally Graded Structures for Load Bearing Implants. *J Mater Sci - Mater Med*. 2008 In press. DOI: 10.1007/s10856-008-3478-2.
29. Vamsi Krishna B, Xue Weichang, Bose Susmita, Bandyopadhyay Amit. Functionally Graded Co-Cr-Mo Coating on Ti-6Al-4V Alloy Structures. *Acta Biomater* 2008;4:697–706. [PubMed: 18054298]
30. Roy, Mangal; Vamsi Krishna, B.; Bandyopadhyay, Amit; Bose, Susmita. Laser Processing of Bioactive tricalcium phosphate coating on titanium for load bearing implant. *Acta Biomater* 2008;4:324–333. [PubMed: 18039597]
31. Vamsi Krishna B, Bose Susmita, Bandyopadhyay Amit. Low Stiffness Porous Ti Structures for Load Bearing Implants. *Acta Biomater* 2007;3:997–1006. [PubMed: 17532277]
32. Vamsi Krishna B, Bose Susmita, Bandyopadhyay Amit. Laser Processing of Net-Shape NiTi Shape Memory Alloy. *Metall Mater Trans A* 2007;38A:1096–1103.

33. Xue, Weichang; Vamsi Krishna, B.; Bandyopadhyay, Amit; Bose, Susmita. Processing and biocompatibility evaluation of laser processed porous titanium. *Acta Biomater* 2007;3:1007–1018. [PubMed: 17627910]
34. Balla, Vamsi Krishna; Xue, Weichang; Bose, Susmita; Bandyopadhyay, Amit. Laser Assisted Zr/ZrO₂ Coating on Ti for Load-bearing implants. *Acta Biomater* 2009;5:2800–2809. [PubMed: 19398221]
35. Duley, WW. CO₂ lasers: effects and applications. Academic Press; New York: 1976.
36. Webb K, Hlady V, Tresco PA. Relative importance of surface wettability and charged functional groups on NIH3T3 fibroblast attachment, spreading and cytoskeletal organization. *J Biomed Mater Res* 1998;41:422–430. [PubMed: 9659612]
37. Zhang YM, Bataillon-Linez P, Huang P, Zhao YM, Han Y, Traisnel M, Xu KM, Hilderbrand HF. Surface analyses of micro-arc oxidized and hydrothermally treated titanium and effect on osteoblast behavior. *J Biomed Mater Res* 2004;68A:383–391.
38. Redey SA, Nardin M, Bernache-Assolant D, Rey C, Delannoy P, Sedel L, Marie PJ. Behavior of human osteoblastic cells on stoichiometric hydroxyapatite and type A carbonate apatite: role of surface energy. *J Biomed Mater Res* 2000;50:353–364. [PubMed: 10737877]
39. Das, Kakoli; Bose, Susmita; Bandyopadhyay, Amit. Surface modifications and cell-materials interactions with anodized Ti. *Acta Biomaterialia* 2007;3:573–585. [PubMed: 17320494]

Figure 1.

Cross-sectional microstructures of Ta coating on Ti (a) full coating. Arrows indicate unmelted/partially melted Ta powder. (b) coating-substrate interface. (c) High magnification SEM image showing the grain size of laser processed Ta coating.

Figure 2.
Hardness profile across the laser processed Ta coating on Ti.

Figure 3. FE-SEM micrographs illustrating the hFOB cell morphologies, (a) Ti surface, after 3 days, (b) Ta surface, after 3 days, (c) Ti surface, after 14 days, and (d) Ta surface, after 14 days.

Figure 4. MTT assay of hFOB cells observed on Ta and Ti surfaces as a function of culture time. A higher optical density represents a higher concentration of living cells.

Table 1

Contact angles (°) of various liquids and surface energy (mN m^{-1}) of titanium and tantalum surfaces.

Surface	Distilled Water	Cell Media	Diiodomethane	Surface Energy
Titanium	69.0 ± 1.5	62.0 ± 2.5	55.0 ± 3.0	42
Tantalum Coating	51.0 ± 2.0	48.0 ± 0.8	48.8 ± 0.9	55