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(71) Applicant(s)
Ossio Ltd.

(72) Inventor(s)
Preiss-Bloom, Orahn;Lindner, Taly Pnina

(74) Agent / Attorney
WRAYS PTY LTD, L7 863 Hay St, Perth, WA, 6000, AU

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(71) Applicant: **OSSIO LTD.** [IL/IL]; 58 Hatachana St., 3052643 Binyamina (IL).

(72) Inventors: **PREISS-BLOOM, Orahn**; 4 HaShlosha St., 30900 Zichron Yakov (IL). **LINDNER, Taly Pnina**; 39 Hazait St., 56905 Savyon (IL).

(74) Agent: **TURIEL, Avi**; DR D GRAESER, c/o Wework, 1 Aire Shenkar St., 4672501 Herzlia (IL).

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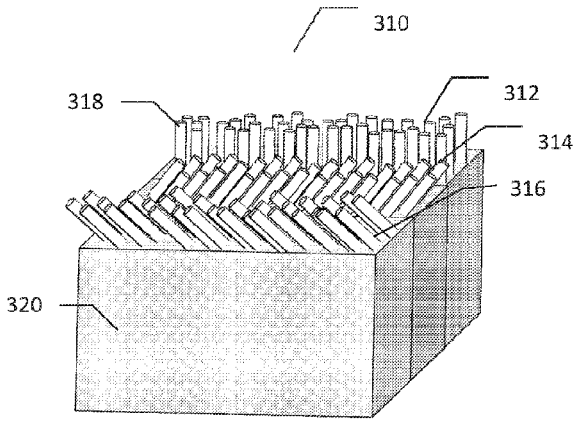
(54) **Title:** FIBER REINFORCED BIOCOMPOSITE MEDICAL IMPLANTS WITH HIGH MINERAL CONTENT

Figure 32b. Cut-away view of continuous-fiber reinforced sheet structure wherein sheet is comprised of multiple layers, each aligned at an angle to each other.



(57) **Abstract:** A medical implant comprising a plurality of layers, each layer comprising a polymer and a plurality of uni-directionally aligned continuous reinforcement fibers.

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PCT APPLICATION

Title: FIBER REINFORCED BIOCOMPOSITE MEDICAL IMPLANTS WITH HIGH MINERAL CONTENT

BACKGROUND

Permanent Orthopedic Implant Materials

Medical implants can be manufactured from metals, alloys, ceramics or both degradable and stable composites. In load-bearing, orthopedic applications that require high strength, usually stainless steel or titanium alloys are used. Metal implants have a long history of successful use in orthopedic surgery but also carry many risks for complications. Although these materials are inert, they are also used in situations in which the need for the implant is only temporary, like in fracture fixation. In the case of metal rods and plates for fracture fixation, a second surgery for device removal may be recommended about one year after confirmation of osseous union. Implant removal causes additional risk and added morbidity for the patient, occupies the availability of clinics, and increases the overall procedure costs. If the device is not removed, it may cause remodeling of the bone. Such remodeling may in turn weaken the bone due to stress shielding or inflammation of the host tissue. The stress shielding can occur due to the high stiffness (modulus) and strength of the metals compared to the stiffness and strength of the cortical bone, so that the metal stresses the bone, which can result in periprosthetic fractures or loss of bone strength.

Examples of load-bearing medical implants that have traditionally been constructed of metal alloys include bone plates, rods, screws, tacks, nails, clamps, and pins for the fixation of bone fractures and/or osteotomies to immobilize the bone fragments for healing. Other examples include cervical wedges, lumbar cages and plates and screws for vertebral fusion and other operations in spinal surgery.

Biostable polymers and their composites e.g. based on polymethacrylate (PMMA), ultra high molecular weight polyethylene (UHMWPE), polytetrafluoroethylene (PTFE), polyetheretherketone (PEEK), polysiloxane and acrylic polymers have also been used to manufacture medical implants. These

materials are not biodegradable or bioresorbable and therefore face many of the same limitations as the metals when used for medical implant applications, for example they may require a second surgery for replacing or removing the implant at some point of the lifetime of the implant. Furthermore, these materials are weaker (less strong and stiff) than metal such that they are more susceptible to mechanical failure, particularly after repeated dynamic loading (i.e. through material fatigue or creep).

Existing degradable polymer medical implants

Resorbable polymers have been used to develop resorbable implants, which can also be referred to as absorbable, bioabsorbable, or biodegradable implants. The advantage of using biocompatible, resorbable polymers is that the polymers, and thus the implant, resorb in the body and release non-toxic degradation products that are metabolized by the metabolic system. Polymers, including polylactic and polyglycolic acids and polydioxanone, are resorbable biocompatible materials that are currently used as orthopedic plates, rods, anchors, pins or screws for non-load bearing medical implant applications, such as craniofacial applications. These medical implant materials offer the advantage of eventual resorption, eliminating the need for later removal, while allowing stress transfer to the remodeling fracture. However, current bioabsorbable materials and implants do not have mechanical properties to match metallic implants. The mechanical strength and modulus (approximately 3-5 GPa) of non-reinforced resorbable polymers, is insufficient to support fractured cortical bone, which has an elastic modulus in the range of approximately 15-20 GPa (Snyder SM, et al. measured the bending modulus of human tibial bone to be about 17.5 GPa *Snyder SM Schneider E, Journal of Orthopedic Research, Vol. 9, 1991, pp. 422-431*). Therefore, the indications of existing medical implants constructed from resorbable polymers are limited and their fixation usually requires protection from motion or significant loading. These devices are only a consideration when fixation of low stress areas is needed (i.e. non-load bearing applications) such as in pediatric patients or in medial malleolar fractures, syndesmotic fixation, maxillofacial, or osteochondral fractures in adults.

Reinforced degradable polymer materials

Recently, reinforced polymer materials with improved strength and stiffness (modulus) have been introduced. These biodegradable composites comprise polymers reinforced by fillers, usually in fiber form. In composite materials, usually a relatively flexible matrix (i.e. a polymer) is combined with a stiff and strong reinforcement material to enhance the mechanical properties of the composite matrix. For example, biodegradable glass or mineral material can be used to improve the stiffness and strength of a biodegradable polymer matrix. In the prior art, several attempts to produce such a composite were reported where bioactive glass particles, hydroxyapatite powder, or short glass fibers were used to enhance the properties of a biodegradable polymer. In most cases, the strength and stiffness of these composites is lower than cortical bone or becomes lower than cortical bone following rapid degradation in a physiological environment. Therefore, the majority of these composite materials are not appropriate for use in load-bearing medical implant applications. However, biodegradable composites with strength and stiffness equivalent to or greater than cortical bone have recently been reported, for example a biodegradable composite comprising a biodegradable polymer and 20-70 vol% glass fibers (WO2010128039 A1). Other composite material implants, for example formed of polymer reinforced with fibers, are disclosed in US Patents 4,750,905, 5,181,930, 5,397,358, 5,009,664, 5,064,439, 4,978,360, 7,419,714, the disclosures of which are incorporated herein by reference

Degradation Mechanism of Reinforced Degradable Polymer Materials

When biodegradable composites are used for load-bearing medical implant applications, such as to fixate bone fractures, the mechanical properties of the medical implant must be retained for an extended period. Degradation of the composite will result in premature loss of implant strength or stiffness and can lead to implant function failure, such as insufficient fixation of bone segments resulting in improper bone healing.

Biodegradable composites will begin to hydrolytically degrade once they come into contact with body fluid. This degradation can be a result of degradation of

the biodegradable polymer, reinforcing filler, or both. Such degradation in an aqueous environment, such as the physiological environment, can particularly result in a sharp drop-off of mechanical strength and stiffness in certain reinforced polymer materials that are reinforced by inorganic compounds. Where the absorbable polymer matrix is organic material, and the fillers are inorganic compounds, the adhesion between the absorbable polymer matrix and the filler may be reduced by degradation of either the polymer or filler in the aqueous environment and become rapidly reduced such that the initial mechanical properties of the reinforced polymer drop-off rapidly and become less than desirable for adequate load-bearing performance. Aside from the degradation of the polymer and filler separately, poor polymer to reinforcement interface interaction and adhesion can result in early failure at the interface in a aqueous environment, thereby resulting in sharp mechanical property drop off as the reinforcement detaches from the polymer and the reinforcing effect of the filler is lost.

Törmälä et al. (WO 2006/114483) described a composite material containing two reinforcing fibers, one polymeric and one ceramic, in a polymer matrix and reported good initial mechanical results (bending strength of 420 +/-39 MPa and bending modulus of 21.5 GPa) equivalent to the properties of cortical bone. However, the prior art teaches that bioabsorbable composites reinforced with absorbable glass fibers, have a high initial bending modulus but that they rapidly lose their strength and modulus *in vitro*.

While improved interfacial bonding (such as covalent bonding) between the polymer and reinforcement can significantly prolong reinforced bioabsorbable polymer mechanical property retention in an aqueous environment (WO2010128039 A1), continued hydrolysis of the polymer, reinforcement, or interface between the two will result in loss of mechanical properties over time. Since osseous union may take several months or longer, even the prolonged mechanical property degradation profile in covalently bonded reinforced bioabsorbable polymers may be insufficient for optimal function of medical implants used for load-bearing orthopedic applications.

An example of strength loss in a reinforced degradable polymer implant is described with regard to self-reinforced poly-L-lactic acid *(Majola A et al., Journal of Materials Science Materials in Medicine, Vol. 3, 1992, pp.43-47).* There, the strength and strength retention of self-reinforced poly-L-lactic acid (SR-PLLA) composite rods

were evaluated after intramedullary and subcutaneous implantation in rabbits. The initial bending strength of the SR-PLLA rods was 250-271 MPa. After intramedullary and subcutaneous implantation of 12 weeks the bending strength of the SR-PLLA implants was 100 MPa.

Co- and terpolyesters of PLA, PGA and PCL are of interest in the tailoring of the optimal polymer for resorbable composite material for medical devices. The choice of monomer ratio and molecular weight significantly affects the strength elasticity, modulus, thermal properties, degradation rate and melt viscosity of resorbable composite materials and all of these polymers are known to be degradable in aqueous conditions, both *in vitro* and *in vivo*. Two stages have been identified in the degradation process: First, degradation proceeds by random hydrolytic chain scission of the ester linkages which decreases the molecular weight of the polymers. In the second stage measurable weight loss in addition to chain scission is observed. The mechanical properties are mainly lost or at least a remarkable drop will be seen in them at the point where weight loss starts. Degradation rate of these polymers is different depending on the polymer structure: crystallinity, molecular weight, glass transition temperature, block length, racemization and chain architecture. *(Middleton JC, Tipton AJ, Biomaterials 21, 2000, 2335-2346)*

<u>The unsolved problem of mineral content in orthopedic implants</u>

As previously described, attempts have been made to produce orthopedic fixation implants from bioabsorbable polymers such as poly lactic acid (PLA). However, these implants derived their mechanical properties solely from the PLA acidic polymer chains. Thus, their strength was limited (a fraction of the strength and modulus of bone) and the acidic burst degradation process of these bioabsorbable polymer implants resulted in problematic local tissue response (cysts, abcesses, etc). The bone attachment to these implants was poor.

Manufacturers have responded to the inflammatory local tissue response and poor bone attachment of bioabsorbable fixation devices by mixing various mineral compositions into the bioabsorbable polymer compositions. For mineral compositions, companies have used minerals or mineral compositions with

osteoconductive properties. Some use Tricalcium phosphate, some use hydroxyapatite, some use calcium sulfate, some use mixtures of these. These mixed composition implants are called "biocomposite" implants and incorporate 25-35% mineral and the mineral powder is evenly distributed into the polymer composition.

Unfortunately, the mineral additive in these biocomposite implants reduces the mechanical properties of the implants since the mechanical strength of these implants derives from the bioabsorbable polymer and there is less polymer in the implant once the mineral composition has been added. Thus, biocomposite implants tend to be more brittle than equivalent implants comprised entirely of bioabsorbable polymers. Higher amounts of mineral than the existing 25-35% cannot be used since the implant will be lacking in mechanical properties .

On the other hand, without the mineral composition, the long term implantation results of existing biocomposite implants are problematic. These implants still suffer from the inflammatory tissue response that has plagued bioabsorbable polymer implants. For example, in ACL interference screws comprised of biocomposite compositions, it has been demonstrated (Cox CL et al. J Bone Joint Surg Am. 2014; 96:244-50) that biocomposite screws result in a very high percentage of inflammatory reactions (cysts, edema). Furthermore, they don't really encourage biointegration. As the article concludes "Even though these newer-generation bioabsorbable screws were designed to promote osseous integration, no tunnel narrowing was noted".

Besides for these inflammatory problems, the current biocomposite screws also are lacking in sufficient mechanical properties (Mascarenhas et al. Arthroscopy: J Arthroscopic & Related Surg 2015: 31(3): pp 561-568). As the article concludes, "The major findings of this study were prolonged knee effusion, increased femoral tunnel widening, and increased screw breakage associated with Bioabsorbable Interference Screw use".

On a mechanical level, higher percentage level of mineral composition in a biocomposite implant can lead to poor mechanical results and specifically mechanical results that are inferior to the mechanical results of implants comprised solely of

bioabsorbable polymer. For example, the effect of different percentages of beta-tricalcium phosphate (βTCP) on the mechanical properties of a PLA based biocomposite have been studied (Ferri JM et al. J Composite Materials. 2016; 0(0): 1-10).

In that study, it was shown that higher percentages of βTCP result in a significant loss of tensile strength for the PLA- βTCP biocomposite, shown in Figure 1 of that reference.

Furthermore, an increase in the percentage of βTCP results in a significant loss in the amount of energy the biocomposite can absorb, as measured as Charpy's impact energy. This is a very important parameter in orthopedic implants since a key property of an orthopedic implant is the ability to withstand impact without fracturing. Table 2 (taken from the above reference) demonstrates this problem.

Table 2. Shore D hardness values and Charpy's absorbed energy of PLA/β-TCP composites in terms of the β-TCP weight percent.

Wt% β-TCP	Shore D hardness	Charpy's impact energy (J/m^2)
0	71 ± 1	1.85 ± 0.2
10	74 ± 1	1.68 ± 0.3
20	75 ± 1	1.40 ± 0.2
30	77 ± 1	1.25 ± 0.1
40	79 ± 1	1.10 ± 0.2

SUMMARY OF THE INVENTION

There is a great need for a reinforced bioabsorbable polymer material exhibiting improved mechanical properties for use in load-bearing medical implant applications, such as structural fixation for load-bearing purposes, where the high strength and stiffness of the implant are retained at a level equivalent to or exceeding cortical bone for a period at least as long as the maximum bone healing time.

The construction of biocomposite fiber-reinforced materials with the requisite high strength and stiffness is known in the art to be a difficult problem, which so far has not been provided with an adequate solution.

Specifically within such fiber-reinforced composites, achieving the high strengths and stiffness required for many medical implant applications can require the use of fiber reinforcement with a high mineral content percentage comprised of either continuous fibers or short or long fiber reinforcement. This creates a significant difference from the implant structures, architectures, designs, and production techniques that have been previously used with medical implants produced from polymers or composites comprising lower mineral content particle or short fiber reinforced polymers. Those implants are most commonly produced using injection molding, or occasionally 3-D printing, production techniques.

Unlike with bulk materials, the properties of parts made from composite materials are highly dependent on the internal structure of the part. This is a well-established principle in the design of parts from composite materials where the mechanical properties of fiber-reinforced composite materials are known to be dependent on the angles and orientations of the fibers within the composite parts.

The vast majority of prior composite material part design focused exclusively on the mechanical properties of the parts. However, these parts were permanent parts and not degradable or absorbable. Therefore, no attention had to be given to the mechanisms of degradation or absorption of the composite materials within the part. Even previous orthopedic implants comprised of composite materials have largely adhered to these same classical composite material design principles.

However, the herein invention relates to medical implants comprised of a new class of composite materials that are biocompatible and in many cases are bioabsorbable. The design challenges in creating medical implants with these materials involve consideration of many more aspects and parameters than just the mechanical properties that have previously been considered with composite material parts.

Furthermore, with regard to bioabsorbable fiber-reinforced composite implants, the degradation profile of the composite material within the implant must also be taken into consideration in ensuring that the fibers will provide strength and stiffness reinforcement both initially at the initial time of device implantation and also over the course of its functional period within the body.

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Mechanical properties that are critical to the performance of medical implants in the herein invention include: flexural, tensional, shear, compressional, and torsional strength and stiffness (modulus). In these bioabsorbable medical implants, these properties are critical both at time zero (i.e. in the implant following production) and following a period of implantation in the body. As with previously described parts made from fiber-reinforced composite material, the mechanical properties at time zero are dependent on the alignment and orientation of fibers within the part. However, retaining a large percentage of the mechanical properties following implantation in the body (or simulated implantation) requires additional and different considerations.

As will be described in more detail below, such considerations for the medical implant design can include the following parameters: compositions, component ratios (including specifically mineral content percentage), fiber diameters, fiber distribution, fiber length, fiber alignments and orientations, etc.

These parameters can impact several additional aspects and properties of the herein described medical implant performance:

1. Material degradation rate (degradation products, local pH and ion levels during degradation)

2. Surface properties that affect interface of implant with surrounding local tissue

3. Biological effects such as anti-microbial or osteoconductive properties

4. Response to sterilization processes (such as ethylene oxide gas, gamma or E-beam radiation)

The present invention aims to provide a solution to these problems or at least ameliorate, one or more of the deficiencies of the prior art, or to provide the consumer with a useful or commercial choice by providing, in at least some embodiments, implant compositions from fiber reinforced biocompatible composite materials that are a significant step forward from previous implants in that they can achieve sustainably high, load bearing strengths and stiffness. Additionally, many embodiments of the present invention additionally facilitate these high strength levels with efficient

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implants of low volume. Furthermore, the biocomposite materials described herein are also optionally and preferably bioabsorbable.

The present invention therefore aims to overcome the limitations of previous approaches and provides medical implants comprising (optionally biodegradable) biocomposite compositions featuring fiber-reinforcement that retain their mechanical strength and stiffness for an extended period.

In one embodiment of the present invention, there is provided a medical implant comprising a biocomposite, said biocomposite comprising a polymer and a plurality of reinforcement mineral fibers, wherein a weight percentage of a mineral composition within the biocomposite medical implant is in the range of 40-65%, wherein an average diameter of said fibers is in a range of 3-30 microns; and wherein the reinforcing fibers are fiber segments with an average fiber segment length in the range of 0.5-20 mm, wherein a residual monomer content in the medical implant following production is less than 3%; wherein the mineral composition is provided by a reinforcing mineral fiber made from the mineral composition.

The present invention, in at least some embodiments, further aims to overcome the limitations of previous biocomposite medical implants by providing medical implants comprised of a biocomposite material composition with a high percentage of mineral content and yet with superior mechanical properties. Preferably the mineral composition is provided by a reinforcing fiber made from the mineral composition.

Preferably, the weight percentage of the mineral composition within the biocomposite medical implant is in the range of 40-90%, more preferably the weight percentage is in the range of 40%-70%, more preferably in the range of 40%-65%, and even more preferably the weight percentage is in the range of 45%-60%.

Surprisingly, the inventors have found that such a high percentage or amount of mineral content can yield implants with superior mechanical properties.

Additionally, previous attempts to construct implants with higher mineral contents failed because biocomposite implants are typically injection molded. The flow properties of a composite with an amount or percentage of mineral content in the above high range are more challenging to injection mold.

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These preferential ranges derive from a critical balance between biocompatibility (quiescent inflammatory response) and strong mechanical properties. As discussed previously, higher mineral content percentage in the medical implant has potential beneficial in increasing biocompatibility and safety profile of the implant with the surrounding tissues, especially bony tissues. However, mineral content that is too high can result in an undesirable reduction in mechanical properties. In some cases a reduction in implant mechanical properties will be seen immediately. In other cases, high mineral content can result in an accelerated mechanical degradation process wherein the implant will lose its mechanical properties at an accelerated rate and thereby lose its ability to provide mechanical fixation for an in vivo time period sufficient to support tissue (and especially orthopedic tissue) healing .

[Text continued on page 11]

Preferably the density of the biocomposite composition for use in herein invention is between 1 to 2 g/mL. More preferentially, density is between 1.2 to 1.9 g/mL. Most preferentially between 1.4 to 1.8 g/mL.

Preferably, the mineral content is provided by a reinforcing mineral fiber made from the mineral composition.

Optionally, the diameter of reinforcing fiber for use with herein reinforced biocomposite medical implant can be in the range of 1-100 μm. Preferably, fiber diameter is in the range of 1-20 μm. More preferably, fiber diameter is in the range of 4-16 μm, and most preferably in the range of 9-14 μm.

The standard deviation of fiber diameter between fibers within the medical implant is preferably less than 5 μm, more preferably less than 3 μm, and most preferably less than 1.5 μm. Uniformity of fiber diameter is beneficial for consistent properties throughout the implant.

In one embodiment, reinforcing fibers are fiber segments inside the polymer matrix. Preferably such fiber segments are, on average, of length 0.5-20mm, more preferably the fiber segment length is in the range of 1-15mm, more preferably in the range of 3-10 and most preferably in the range of 4-8mm.

Optionally and preferably the above mineral composition is provided in the form of a reinforcing fiber, present in a sufficiently high amount and with a sufficiently high mineral quantity to provide the above weight percentage of the mineral composition within the implant.

The overall structure of the implant may optionally be heterogeneous and/or amorphous. If heterogeneous, the structure may optionally be continuous in its properties. Alternatively, the implant may optionally be divided into layers.

According to at least some embodiments, there is provided a medical implant comprising a plurality of biocomposite layers, said layers comprising a polymer, which is optionally biodegradable, and a plurality of uni-directionally aligned continuous reinforcement fibers. The layers may optionally be amorphous or aligned. Optionally and preferably, the biodegradable polymer is embodied in a biodegradable composite. Also optionally and preferably, the fibers are embedded in a polymer matrix comprising one or more bioabsorbable polymers.

According to at least some embodiments, the composite layers are each comprised of one or more composite tapes, said tape comprising a polymer, which is optionally biodegradable, and a plurality of uni-directionally aligned continuous reinforcement fibers. Optionally and preferably, the biodegradable polymer is embodied in a biodegradable composite. Also optionally and preferably, the fibers are embedded in a polymer matrix comprising one or more bioabsorbable polymers.

Optionally and preferably, the fiber-reinforced biodegradable composite within the implant has a flexural modulus exceeding 5 GPa and flexural strength exceeding 80 MPa.

Preferably, the fiber-reinforced biodegradable composite within the implant has flexural strength in range of 150 – 800 MPa, more preferably 150-400 MPa. Elastic modulus is preferably in range of 5 – 27 GPa, more preferably 16 – 27 GPa.

Preferably, the fiber-reinforced composite within the implant has strength retention of Elastic Modulus above 5 GPa after 8 weeks implantation and flexural strength above 60 MPa after 8 weeks.

Preferably, the fiber-reinforced composite within the implant has mechanical property retention of Flexural Modulus above 12 GPa and flexural strength above 180 MPa after 5 days of simulated physiological degradation.

More preferably, the fiber-reinforced composite within the implant has mechanical property retention of Flexural Modulus above 10 GPa and flexural strength above 120 MPa after 5 days of simulated physiological degradation.

The term “biodegradable” as used herein also refers to materials that are resorbable, bioabsorbable or absorbable in the body.

BRIEF DESCRIPTION OF THE DRAWINGS

Figure 1: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 6 mm pin with 50% fiber content by weight, such as those described in Example 1. Magnification of this image is 2,500 x. This image shows a magnification of the cross section of reinforcing mineral fibers

102 embedded within bioabsorbable polymer matrix **104**. The fiber diameter is indicated within the image **106.**

Figure 2: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 6 mm pin with 50% fiber content by weight, such as those described in Example 1. Magnification of this image is 2,500 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. The distance between adjacent fibers is indicated by **202**.

Figure 3: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 6 mm pin with 50% fiber content by weight, such as those described in Example 1. Magnification of this image is 500 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. Each layer **306 308 310** is comprised of reinforcement fibers **304** and is of a certain thickness **302.**

Figure 4: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 6 mm pin with 50% fiber content by weight, such as those described in Example 1. Magnification of this image is 150 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix.

Figure 5: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 6 mm pin with 50% fiber content by weight, such as those described in Example 1. Magnification of this image is 500 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. Each layer is separated by an area of bioabsorbable polymer matrix **502.**

Figure 6: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 6 mm pin with 70% fiber content by weight, such as those described in Example 1. Magnification of this image is 500 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. The distance between adjacent fibers is indicated.

Figure 7: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 6 mm pin with 70% fiber content by weight, such as those described in Example 1. Magnification of this image is 500 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix.

Figure 8: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 2,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. The fiber diameter is indicated within the image.

Figure 9: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 2,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. The distance between adjacent fibers is indicated.

Figure 10: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 1,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix.

Figure 11: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 5,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers **1102** embedded within bioabsorbable polymer matrix **1104**.

Figure 12: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au Sputtered cross section of a 2 mm pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 1,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers

embedded within bioabsorbable polymer matrix. Each layer is separated by an area of bioabsorbable polymer matrix.

Figure 13: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au Sputtered cross section of a 2 mm cannulated pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 1,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. The fiber diameter is indicated within the image.

Figure 14: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au Sputtered cross section of a 2 mm cannulated pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 1,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. The distance between adjacent fibers is indicated.

Figure 15: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm cannulated pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 1,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix.

Figure 16: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au Sputtered cross section of a 2 mm cannulated pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 1,000 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. Each layer is separated by an area of bioabsorbable polymer matrix.

Figure 17: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 2 mm plate with 50% fiber content by weight, such as those described in Example 3. Magnification of this image is 1250 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. The fiber diameter is indicated within the image.

Figure 18: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 2 mm plate with 50% fiber content by weight, such as those described in Example 3. Magnification of this image is 1250 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. The distance between adjacent fibers is indicated.

Figure 19: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 2 mm plate with 70% fiber content by weight, such as those described in Example 3. Magnification of this image is 250 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. Each layer **1902, 1904** is comprised of fibers. The distance between adjacent fibers is indicated.

Figure 20: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 2 mm plate with 70% fiber content by weight, such as those described in Example 3. Magnification of this image is 250 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix.

Figure 21: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 2 mm plate with 70% fiber content by weight, such as those described in Example 3. Magnification of this image is 500 x. This image shows a magnification of the cross section of reinforcing mineral fibers embedded within bioabsorbable polymer matrix. Each layer is separated by an area of bioabsorbable polymer matrix.

Figure 22: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 300x. This image shows a magnification of the longitudinal axis of reinforcing mineral fibers **2202**.

Figure 23: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm cannulated pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this

image is 250 x. This image shows a magnification of the cannulated portion and the continuous, reinforcing mineral fibers. The tangential angle **2302** is defined as the deviation from the direction of the curve at a fixed starting point, where the fixed starting point is the point where the fiber touches or is closest to coming into contact with the center of the cross-sectional circular area.

Figure 24: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 6 mm pin with 50% fiber content by weight, such as those described in Example 1. Magnification of this image is 500x. This image shows a magnification of the cross section of reinforcing mineral fibers, bundled tightly together in groups **2402** embedded within bioabsorbable polymer matrix.

Figure 25: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm cannulated pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 500x. This image shows a magnification of the cross section of reinforcing mineral fibers surrounding the inner cannulation of the pin **2502**.

Figure 26: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 2 mm cannulated pin with 50% fiber content by weight, such as those described in Example 2. Magnification of this image is 1000 x. This image shows a magnification of the cross section of reinforcing mineral fibers, embedded within bioabsorbable polymer matrix layers in alternating 0° and 45° orientation.

Figure 27: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 6 mm pin with 85% fiber content by weight, such as those described in Example 1. Magnification 160x. This image shows a magnification of the cross section of reinforcing mineral fibers, embedded within layers **2702** in alternating 0° and 45° orientation, with little or no bioabsorbable polymer matrix separating the layers.

Figure 28: Scanning Electron Microscope (SEM) image using a secondary electron detector of Au sputtered cross section of a 6 mm pin with 85% fiber content by weight, such as those described in Example 1. Magnification 1000x. This image

shows a magnification of the cross section of reinforcing mineral fibers, with little or no bioabsorbable polymer matrix surrounding the said fibers.

Figure 29: Scanning Electron Microscope (SEM) image using a Back-Scattered Electrons (BSE) detector of a cross section of a 2 mm pin with 50% fiber content by weight, such as those described in Example 2. Magnification 60x. This image shows a magnification of the edge of the pin, indicating that the bioabsorbable polymer is present at the outer surface of the implant **2902.**

Figure 30 shows an example of a continuous fiber-reinforced tape of the type that can be used to form a layer in a medical implant comprised of continuous fiber-reinforced layers.

Figure 31 shows an example of a cut-away, three-dimensional view of a continuous fiber-reinforced tape (200).

Figure 32a shows an example of a top-view of a reinforced bioabsorbable composite sheet (300) comprised of three layers of uni-directional fibers at different angles.

Figure 32b shows an example of a cut-away view of a reinforced bioabsorbable composite structure (310) comprised of three layers of uni-directional fibers at different angles.

Figure 33 shows an example of the wall of a continuous-fiber reinforced composite medical implant.

Figure 34 shows an example of a bone filler cage that consists of continuous-fiber reinforced composite medical implant walls (500) that additionally contains perforations (502) to allow tissue and cellular ingrowth into the bone filler material (504) contained within the bone filler cage.

Figure 35 shows an example of a bioabsorbable cannulated screw (600) that is a medical implant.

DETAILED DESCRIPTION

A medical implant according to at least some embodiments of the present invention is suitable for load-bearing orthopedic implant applications and comprises

one or more biocomposite, optionally bioabsorbable, materials where sustained mechanical strength and stiffness are critical for proper implant function and wherein the implant is additionally comprised of a moisture barrier coating that restricts or eliminates fluid exchange into the implant.

The present invention, according to at least some embodiments, thus provides medical implants that are useful as structural fixation for load-bearing purposes, exhibiting sustained mechanical properties as a result of impeded degradation of the bioabsorbable materials that comprise the implant.

Relevant implants may include bone fixation plates, intramedullary nails, joint (hip, knee, elbow) implants, spine implants, suture anchors, screws, pins, wires, bone cages, and other devices for such applications such as for fracture fixation, tendon reattachment, spinal fixation, soft tissue repair, and spinal cages.

According to at least some embodiments, the herein invention relates to medical implants comprised of a biocomposite material composition. Preferably the biocomposite material composition is comprised of (an optionally bioabsorbable) polymer reinforced by a mineral composition. Preferably the mineral composition reinforcement is provided by a reinforcing fiber made from the mineral composition. As described above, the mineral content of the implant is preferably quite high.

Optionally, the medical implant or part thereof is comprised of a number of biocomposite layers, each layer comprising bioabsorbable polymer reinforced by uni-directional reinforcing fibers. The properties of the implant are optionally and preferably determined according to the layer composition and structure, and the placement of the layers in regard to the device, for example with regard to layer direction. The fibers may optionally remain discrete but optionally some melting of the polymer may occur to bind the layers together.

A biocomposite layer can be defined as a continuous or semi-continuous stratum running through part or all of a medical implant, wherein the layer is comprised of reinforcing fibers that aligned uni-directionally. Layers can be seen in several figures showing the internal structure of reinforced biocomposite medical implants, including in figure 7, 10, and 20.

Preferably, there are between 1-100 reinforcing fibers forming the thickness of each biocomposite layer. Preferably, there are between 2-40 reinforcing fibers in each layer thickness and most preferably there are between 4-20 reinforcing fibers.

Optionally, the directional fiber orientation between adjacent layers within the implant alternates between layers such that each adjacent layer is out of phase (of a different angle) from the layer that is adjacent to it. Preferably, the average or median angle difference between layers is between 15 to 75 degrees, more preferably between 30 to 60 degrees, and most preferably between 40 to 50 degrees. Microscopic images of such out of phase adjacent biocomposite layers can be seen in figure 26 and 27.

Preferably, the biocomposite layers within the medical implant are well approximated to each other. More preferably, the distance between layers, as measured by the distance between the last fiber in one layers and the first fiber in the subsequent layer is between 0-200 μm, more preferably between 0-60 μm, 1-40 μm, and most preferably between 2-30 μm. Good approximation of the fibers within a layer to the fibers within the adjacent layer allow each layer to mechanically support the adjacent layer. However, some distance between the layers may be desirable to allow for some polymer to remain between the fibers of adjacent layers and thus adhere the layers together, prevent layer dehiscence under high mechanical load .

Optionally, the diameter of a majority of reinforcing fiber for use with herein reinforced biocomposite medical implant is in the range of 1-100 μm. Preferably, fiber diameter is in the range of 1-20 μm. More preferably, fiber diameter is in the range of 4-16 μm, and most preferably in the range of 9-14 μm.

Optionally, the average diameter of reinforcing fiber for use with herein reinforced biocomposite medical implant is in the range of 1-100 μm. Preferably, fiber diameter is in the range of 1-20 μm. More preferably, fiber diameter is in the range of 4-16 μm, and most preferably in the range of 9-14 μm.

The standard deviation of fiber diameter between fibers within the medical implant is preferably less than 5 μm, more preferably less than 3 μm, and most preferably less than 1.5 μm. Uniformity of fiber diameter is beneficial for consistent properties throughout the implant.

In one embodiment, reinforcing fibers are fiber segments inside the polymer matrix. Preferably such fiber segments are, on average, of length 0.5-20mm, more preferably the fiber segment length is in the range of 1-15mm, more preferably in the range of 3-10 and most preferably in the range of 4-8mm.

Preferably, a majority of reinforcing fiber segments are of length 0.5-20mm, more preferably the fiber segment length is in the range of 1-15mm, more preferably in the range of 3-10 and most preferably in the range of 4-8mm.

Optionally, the reinforcing fibers are continuous fibers. Said continuous fibers are preferably longer than 5 mm, more preferably longer than 8 mm, 12 mm, 16 mm, and most preferably longer than 20 mm. A microscopic image of such continuous fibers can be seen in figure 22.

Alternatively, or in addition, the reinforcing fiber length can be defined as a function of implant length wherein at least a portion of the reinforcing fibers, and preferably a majority of the reinforcing fibers, are of a continuous length at least 50% the longitudinal length of the medical implant or medical implant component that is comprised of these fibers. Preferably, the portion or majority of the reinforcing fibers are of continuous length at least 60% of the length of the medical implant, and more preferably at least 75% of the length of the medical implant. Such continuous reinforcing fibers can provide structural reinforcement to a large part of the implant.

Optionally, the distance between adjacent reinforcing fibers within a biocomposite layer is in the range of 0.5-50 μm, preferably the distance between adjacent fibers is in the range of 1-30 μm, more preferably in the range of 1-20 μm, and most preferably in the range of 1-10 μm.

Preferably, the weight percentage of the reinforcing fibers (mineral composition) within the biocomposite medical implant is in the range of 40-90%, more preferably the weight percentage is in the range of 40%-70%, more preferably in the range of 40%-60%, and even more preferably the weight percentage is in the range of 45%-60%.

Preferably, the volume percentage of reinforcing fibers within the biocomposite medical implant is in the range of 30-90%, more preferably the volume percentage is in the range of 40%-70%.

Optionally, a plurality of fibers within the implant are uni-directionally aligned. Optionally, the aligned fiber segments are, on average, of length 5-12mm.

Preferably, the uni-directionally aligned fibers are aligned in the longitudinal access of the implant (0º alignments in relation to the longitudinal axis). Preferably, a majority of fibers are uni-directionally aligned in the longitudinal axis. Optionally, more than 70%, 80%, 90%, 95% of fibers are uni-directionally aligned in the longitudinal axis.

Optionally, a plurality or a majority of fibers within the implant are aligned in the longitudinal axis. Optionally, a plurality of fibers are additionally aligned in up to 3 additional directions. Optionally, a plurality of fibers are aligned in a selection of each of the following alignments in relation to the longitudinal axis: 0º, 30º, -30º, 45º, -45º, 90º. Preferably, a plurality of fibers are aligned in a selection of each of the following alignments in relation to the longitudinal axis: 0º, 45º, -45º, 90º. More preferably, a plurality of fibers are aligned in a selection of each of the following alignments in relation to the longitudinal axis: 0º, 45º, -45º.

Optionally, a majority of fibers are aligned in the longitudinal access of the implant and a plurality of fibers are aligned in each of the following alignments in relation to the longitudinal axis: 45º, -45º.

Optionally and alternatively, fiber segments are arranged amorphously.

While the biocomposite composition within the implant is important in determining the mechanical and bulk properties of the implant, the specific composition and structure that comes into contact with the surface edge of the implant has unique significance in that this composition and structure can greatly affect how surrounding cells and tissue interact with the implant following implantation into the body. For example, the absorbable polymer part of the biocomposite may be hydrophobic in nature such that it will repel surrounding tissues to a certain degree while the mineral reinforcing fiber part of the biocomposite may be hydrophilic in nature and therefore encourage surrounding tissues to attach to the implant or create tissue ingrowth .

In an optional embodiment of the herein invention, the surface presence of one of the compositional components by percentage of surface area is greater than the presence of that component in the bulk composition of the implant by volume percentage. For example, the amount of mineral on the surface might be greater than the amount of polymer, or vice versa. Without wishing to be limited by a single hypothesis, for greater integration with bone, a greater amount of mineral would optionally and preferably be present on the surface. For reduced integration with bone, a greater amount of polymer would optionally and preferably be present on the surface. Preferably, the percentage of surface area composition of one component is more than 10% greater than the percentage of volume percentage of that component in the overall biocomposite implant. More preferably, the percentage is more than 30% greater, and most preferably more than 50% greater. Fig 25 shows a microscopic image of a biocomposite medical implant with a predominance of mineral reinforcing fiber along the inner surface area edge of the implant. Fig 29 shows a microscopic image of a biocomposite medical implant with a predominance of bioabsorbable polymer along the outer surface area of the implant .

Optionally, one surface of the medical implant may have a local predominance of one of the biocomposite components while a different surface, or different part of the same surface, may have a local predominance of a different biocomposite component

Optionally, mineral content is not present in a majority of the surface area (i.e. a majority of the surface of the implant is covered with a polymer film). Optionally, the surface polymer film is, on average, 0.5-50 μm in thickness, more preferably 5-50 μm and most preferably 10-40 μm..

Optionally, there are fibers exposed at the surface of the implant. Optionally, exposed fibers comprise 1-60% of implant surface. Optionally, exposed fibers comprise 10-50% of implant surface. Optinally, exposed fibers comprise 15-30% of implant surface.

Optionally, the medical implant is a threaded screw or other threaded implant. Preferably, the outer layer of the implant will be directionally aligned such that the direction of the fibers approximates the helix angle of the threading. Preferably, the

alignment angle of the fiber direction is within 45 degrees of the helix angle. More preferably, the alignment angle is within 30 degrees, and most preferably the alignment angle is within 15 degrees of the helix angle. Approximating the fiber alignment angle to the helix angle in this manner can improve the robustness of the threading and prevent dehiscence of the reinforcing fibers within the threading .

With regard to circular implants, the reinforcing fibers may optionally take the full circular shape of the implant and curve around the circle shape of the implant without deviation from its circumference. Preferably, a portion or a majority of the reinforcing fibers deviate from the circle shape of the implant such that a tangential angle is formed. The tangential angle is defined as the deviation from the direction of the curve at a fixed starting point, where the fixed starting point is the point where the fiber touches or is closest to coming into contact with the center of the cross-sectional circular area. Figure 23 depicts the tangential angle of reinforcing fibers to a cannulated circular pin .

Preferably the tangential angle between reinforcing fibers within the circular medical implant and the curvature of the implant is less than 90 degrees, more preferably less than 45 degrees.

Preferably the density of the biocomposite composition for use in herein invention is between 1 to 2 g/mL. More preferentially, density is between 1.2 to 1.9 g/mL. Most preferentially between 1.4 to 1.8 g/mL.

Bioabsorbable Polymers

In a preferred embodiment of the present invention, the biodegradable composite comprises a bioabsorbable polymer.

The medical implant described herein may be made from any biodegradable polymer. The biodegradable polymer may be a homopolymer or a copolymer, including random copolymer, block copolymer, or graft copolymer. The biodegradable polymer may be a linear polymer, a branched polymer, or a dendrimer. The biodegradable polymers may be of natural or synthetic origin. Examples of suitable biodegradable polymers include, but are not limited to polymers such as those made from lactide, glycolide, caprolactone, valerolactone, carbonates (e.g., trimethylene carbonate, tetramethylene carbonate, and the like), dioxanones (e.g., 1,4-

dioxanone), δ-valerolactone, 1,dioxepanones)e.g., 1,4-dioxepan-2-one and 1,5-dioxepan-2-one), ethylene glycol, ethylene oxide, esteramides, γ-ydroxyvalerate, β-hydroxypropionate, alpha-hydroxy acid, hydroxybuterates, poly (ortho esters), hydroxy alkanoates, tyrosine carbonates ,polyimide carbonates, polyimino carbonates such as poly (bisphenol A-iminocarbonate) and poly (hydroquinone-iminocarbonate,(polyurethanes, polyanhydrides, polymer drugs (e.g., polydiflunisol, polyaspirin, and protein therapeutics(and copolymers and combinations thereof. Suitable natural biodegradable polymers include those made from collagen, chitin, chitosan, cellulose, poly (amino acids), polysaccharides, hyaluronic acid, gut, copolymers and derivatives and combinations thereof.

According to the present invention, the biodegradable polymer may be a copolymer or terpolymer, for example: polylactides (PLA), poly-L-lactide (PLLA), poly-DL-lactide (PDLLA); polyglycolide (PGA); copolymers of glycolide, glycolide/trimethylene carbonate copolymers (PGA/TMC); other copolymers of PLA, such as lactide/tetramethylglycolide copolymers, lactide/trimethylene carbonate copolymers, lactide/d-valerolactone copolymers, lactide/ε-caprolactone copolymers, L-lactide/DL-lactide copolymers, glycolide/L-lactide copolymers (PGA/PLLA), polylactide-co-glycolide; terpolymers of PLA, such as lactide/glycolide/trimethylene carbonate terpolymers, lactide/glycolide/ ε -caprolactone terpolymers, PLA/polyethylene oxide copolymers; polydepsipeptides; unsymmetrically – 3,6-substituted poly-1 ,4-dioxane-2,5-diones; polyhydroxyalkanoates; such as polyhydroxybutyrates (PHB); PHB/b-hydroxyvalerate copolymers (PHB/PHV); poly-b-hydroxypropionate (PHPA); poly-p-dioxanone (PDS); poly-d-valerolactone - poly-ε-capralactone, poly(ε-caprolactone-DL-lactide) copolymers; methylmethacrylate-N-vinyl pyrrolidone copolymers; polyesteramides; polyesters of oxalic acid; polydihydropyrans; polyalkyl-2-cyanoacrylates; polyurethanes (PU); polyvinylalcohol (PVA); polypeptides; poly-b-malic acid (PMLA): poly-b-alkanbic acids; polycarbonates; polyorthoesters; polyphosphates; poly(ester anhydrides); and mixtures thereof; and natural polymers, such as sugars; starch, cellulose and cellulose derivatives, polysaccharides, collagen, chitosan, fibrin, hyalyronic acid, polypeptides and proteins. Mixtures of any of the above-mentioned polymers and their various forms may also be used.

Reinforced Bioabsorbable Polymers

According to at least some embodiments of the present invention, the medical implant comprises a reinforced bioabsorbable polymer (i.e. a bioabsorbable composite that includes the previously described polymer and also incorporates a reinforcing filler, generally in fiber form, to increase the mechanical strength of the polymer).

In a more preferred embodiment of the present invention, the reinforced bioabsorbable polymer is a reinforced polymer composition comprised of any of the above-mentioned bioabsorbable polymers and a reinforcing filler, preferably in fiber form. The reinforcing filler may be comprised of organic or inorganic (that is, natural or synthetic) material. Reinforcing filler may be a biodegradable glass, a cellulosic material, a nano-diamond, or any other filler known in the art to increase the mechanical properties of a bioabsorbable polymer. The filler is preferably made from a material or class of material other than the bioabsorbable polymer itself. However, it may also optionally be a fiber of a bioabsorbable polymer itself.

Numerous examples of such reinforced polymer compositions have previously been documented. For example: A biocompatible and resorbable melt derived glass composition where glass fibers can be embedded in a continuous polymer matrix (EP 2 243 749 A1), Biodegradable composite comprising a biodegradable polymer and 20-70 vol% glass fibers (WO2010128039 A1), Resorbable and biocompatible fiber glass that can be embedded in polymer matrix (US 2012/0040002 A1), Biocompatible composite and its use (US 2012/0040015 A1), Absorbable polymer containing poly[succinimide] as a filler (EP0 671 177 B1).

In a more preferred embodiment of the present invention, the reinforcing filler is bound to the bioabsorbable polymer such that the reinforcing effect is maintained for an extended period. Such an approach has been described in US 2012/0040002 A1 and EP 2243500B1, which discusses a composite material comprising biocompatible glass, a biocompatible matrix polymer and a coupling agent capable of forming covalent bonds.

As noted above, the biodegradable composite and fibers are preferably arranged in the form of biodegradable composite layers, where each layer comprises

uni-directionally aligned continuous reinforcement fibers embedded in a polymer matrix comprised of one or more bioabsorbable polymers.

The biodegradable composite layers are preferably comprised of one or more biodegradable composite tapes, where each tape comprises uni-directionally aligned continuous reinforcement fibers embedded in a polymer matrix comprised of one or more bioabsorbable polymers.

The biodegradable composite is preferably embodied in a polymer matrix, which may optionally comprise any of the above polymers. Optionally and preferably, it may comprise a polymer selected from the group consisting of PLLA (poly-L-lactide), PDLLA (poly-DL-lactide), PLDLA, PGA (poly-glycolic acid), PLGA (poly-lactide-glycolic acid), PCL (Polycaprolactone), PLLA-PCL and a combination thereof. If PLLA is used, the matrix preferably comprises at least 30% PLLA, more preferably 50%, and most preferably at least 70% PLLA. If PDLA is used, the matrix preferably comprises at least 5% PDLA, more preferably at least 10%, most preferably at least 20% PDLA.

Preferably, the inherent viscosity (IV) of the polymer matrix (independent of the reinforcement fiber) is in the range of 1.2 to 2.4 dl/g, more preferably in the range of 1.5 to 2.1 dl/g, and most preferably in the range of 1.7 to 1.9 dl/g.

Inherent Viscosity (IV) is a viscometric method for measuring molecular size. IV is based on the flow time of a polymer solution through a narrow capillary relative to the flow time of the pure solvent through the capillary.

Reinforcement Fiber

Preferably, reinforcement fiber is comprised of silica-based mineral compound such that reinforcement fiber comprises a bioresorbable glass fiber, which can also be termed a bioglass fiber composite.

Mineral composition may include beta-tricalcium phosphate, calcium phosphate, calcium sulfate, hydroxyapatite, or a bioresorbable glass (also known as bioglass).

Additional optional glass fiber compositions have been described previously by Lehtonen TJ et al. (Acta Biomaterialia 9 (2013) 4868–4877), which is included here by reference in its entirety; such glass fiber compositions may optionally be used in place of or in addition to the above compositions.

Additional optional bioresorbable glass compositions are described in the following patent applications, which are hereby incorporated by reference as if fully set forth herein: Biocompatible composite and its use (WO2010122098); and Resorbable and biocompatible fibre glass compositions and their uses (WO2010122019).

In a more preferred embodiment of the present invention, the reinforcing filler is bound to the bioabsorbable polymer such that the reinforcing effect is maintained for an extended period. Such an approach has been described in US 2012/0040002 A1 and EP 2243500B1, which discusses a composite material comprising biocompatible glass, a biocompatible matrix polymer and a coupling agent capable of forming covalent bonds.

Bioresorbable glass fiber may optionally have oxide compositions in the following mol.% ranges:

Na_2O: 11.0 - 19.0 mol.%

CaO: 9.0 – 14.0 mol.%

MgO: 1.5 – 8.0 mol.%

B_2O_3: 0.5 – 3.0 mol.%

Al_2O_3: 0 – 0.8 mol.%

P_2O_3: 0.1 – 0.8 mol.%

SiO_2: 67 – 73 mol.%

And more preferably in the following mol.% ranges:

Na_2O: 12.0 - 13.0 mol.%

CaO: 9.0 – 10.0 mol.%

MgO: 7.0 – 8.0 mol.%

B_2O_3: 1.4 – 2.0 mol.%

P_2O_3: 0.5 – 0.8 mol.%

SiO_2: 68 – 70 mol.%

Additional optional glass fiber compositions have been described previously by Lehtonen TJ et al. (*Acta Biomaterialia* 9 (2013) 4868–4877), which is included here by reference in its entirety; such glass fiber compositions may optionally be used in place of or in addition to the above compositions.

Additional optional bioresorbable glass compositions are described in the following patent applications, which are hereby incorporated by reference as if fully set forth herein: Biocompatible composite and its use (WO2010122098); and Resorbable and biocompatible fibre glass compositions and their uses (WO2010122019).

Optional Additional Features

The below features and embodiments may optionally be combined with any of the above features and embodiments.

Tensile strength of the reinforcement fiber is preferably in the range of 1200-2800 MPa, more preferably in the range of 1600-2400 MPa, and most preferably in the range of 1800-2200 MPa.

Elastic modulus of the reinforcement fiber is preferably in the range of 30-100 GPa, more preferably in the range of 50-80 GPa, and most preferably in the range of 60-70 GPa.

Fiber diameter is preferably in the range of 6-20 μm, more preferably in the range of 10-18 μm, and most preferably in the range of 14-16 μm.

Optionally, a majority of reinforcement fibers aligned to the longitudinal axis of the medical implant are of a length of at least 50% of the total length of the implant, preferably at least 60%, more preferably at least 75%, and most preferably at least 85%.

Optionally, fibers may be aligned at an angle to the longitudinal axis (i.e. on a diagonal) such that the length of the fiber may be greater than 100% of the length of the implant. Optionally and preferably, a majority of reinforcement fibers are aligned at an angle that is less than 90°, alternatively less than 60°, or optionally less than 45° from the longitudinal axis.

Preferably, the implant preferably comprises between 2-20 composite tape layers, more preferably between 2-10 layers, and most preferably between 2-6 layers; wherein each layer may be aligned in a different direction or some of the layers may be aligned in the same direction as the other layers.

Preferably, the maximum angle between fibers in at least some of the layers is greater than the angle between the fibers in each layer and the longitudinal axis. For example, one layer of reinforcing fibers may be aligned and a right diagonal to the longitudinal axis while another layer may be aligned at a left diagonal to the longitudinal axis.

Compatibilizer

Optionally and preferably, the composite composition additionally includes a compatibilizer, which for example be such an agent as described in WO2010122098, hereby incorporated by reference as if fully set forth herein.

Biodegradable Composite Alternative Forms

Alternatively, biodegradable composite may comprise composite strands comprising continuous reinforcement fibers or fiber bundles impregnated with bioabsorbable polymer. Preferably, strands are less than 1 cm in diameter. More preferably, strands are less than 8 mm, less than 5 mm, less than 3 mm, or less than 2 mm in diameter.

Alternatively, biodegradable composite may comprise a woven mesh of continuous reinforcement fibers wherein woven mesh is pre-impregnated with bioabsorbable polymer or woven mesh is comprised of reinforcement fibers and subsequently impregnated with bioabsorbable polymer.

Preferably, biodegradable composite mesh layer is less than 1 cm in thickness. More preferably, impregnated mesh is less than 8 mm, less than 5 mm, less than 3 mm, or less than 2 mm in thickness.

Mineral Content

The present invention, in at least some embodiments, further overcomes the limitations of previous biocomposite medical implants by providing medical implants comprised of a biocomposite material composition with a high percentage of mineral content and yet with superior mechanical properties. Preferably the mineral composition is provided by a reinforcing fiber made from the mineral composition.

Preferably, the weight percentage of the mineral composition within the biocomposite medical implant is in the range of 40-90%, more preferably the weight percentage is in the range of 40%-70%, and even more preferably the weight percentage is in the range of 45%-60%.

Preferably the density of the biocomposite composition for use in present invention, in at least some embodiments, is between 1 to 2 g/mL. More preferentially, density is between 1.2 to 1.9 g/mL. Most preferentially density is between 1.4 to 1.8 g/mL.

The diameter of reinforcing fiber for use with the reinforced biocomposite medical implant can be in the range of 1-100 μm. Preferably, fiber diameter is in the range of 1-20 μm. More preferably, fiber diameter is in the range of 4-16 μm.

The standard deviation of fiber diameter between fibers within the medical implant is preferably less than 5 μm, more preferably less than 3 μm, and most preferably less than 1.5 μm. Uniformity of fiber diameter is beneficial for consistent properties throughout the implant.

Optionally and preferably, the fiber-reinforced biodegradable composite within the implant has a flexural modulus exceeding 5 GPa and flexural strength exceeding 80 MPa .

Preferably, the fiber-reinforced biodegradable composite within the implant has flexural strength in range of 150 – 800 MPa, more preferably 150 – 400 MPa. Elastic modulus is preferably in range of 5 – 27 GPa, more preferably 10 – 27 GPa.

Preferably, the fiber-reinforced composite within the implant has strength retention of Elastic Modulus above 10 GPa after 8 weeks implantation and flexural strength above 150 MPa after 8 weeks.

According to the present invention, in at least some embodiments, the biodegradable polymer may be a copolymer or terpolymer, for example: polylactides (PLA), poly-L-lactide (PLLA), poly-DL-lactide (PDLLA); polyglycolide (PGA); copolymers of glycolide, glycolide/trimethylene carbonate copolymers (PGA/TMC); other copolymers of PLA, such as lactide/tetramethylglycolide copolymers, lactide/trimethylene carbonate copolymers, lactide/d-valerolactone copolymers, lactide/ε-caprolactone copolymers, L-lactide/DL-lactide copolymers, glycolide/L-lactide copolymers (PGA/PLLA), polylactide-co-glycolide; terpolymers of PLA, such as lactide/glycolide/trimethylene carbonate terpolymers, lactide/glycolide/ ε -caprolactone terpolymers, PLA/polyethylene oxide copolymers; polydepsipeptides; unsymmetrically – 3,6-substituted poly-1 ,4-dioxane-2,5-diones; polyhydroxyalkanoates; such as polyhydroxybutyrates)PHB); PHB/b-hydroxyvalerate copolymers (PHB/PHV); poly-b-hydroxypropionate (PHPA); poly-p-dioxanone (PDS); poly-d-valerolactone - poly-ε-capralactone, poly(ε-caprolactone-DL-lactide) copolymers; methylmethacrylate-N-vinyl pyrrolidone copolymers;

polyesteramides; polyesters of oxalic acid; polydihydropyrans; polyalkyl-2-cyanoacrylates; polyurethanes (PU); polyvinylalcohol (PVA); polypeptides; poly-b-malic acid (PMLA): poly-b-alkanbic acids; polycarbonates; polyorthoesters; polyphosphates; poly(ester anhydrides); and mixtures thereof; and natural polymers, such as sugars; starch, cellulose and cellulose derivatives, polysaccharides, collagen, chitosan, fibrin, hyalyronic acid, polypeptides and proteins. Mixtures of any of the above-mentioned polymers and their various forms may also be used.

The biodegradable composite is preferably embodied in a polymer matrix, which may optionally comprise any of the above polymers. Optionally and preferably, it may comprise a polymer selected from the group consisting of PLLA (poly-L-lactide), PDLLA (poly-DL-lactide), PLDLA, PGA (poly-glycolic acid), PLGA (poly-lactide-glycolic acid), PCL (Polycaprolactone), PLLA-PCL and a combination thereof. If PLLA is used, the matrix preferably comprises at least 30% PLLA, more preferably 50%, and most preferably at least 70% PLLA. If PDLA is used, the matrix preferably comprises at least 5% PDLA, more preferably at least 10%, most preferably at least 20% PDLA .

Preferably, the inherent viscosity (IV) of the polymer matrix (independent of the reinforcement fiber) is in the range of 1.2 to 2.4 dl/g, more preferably in the range of 1.5 to 2.1 dl/g, and most preferably in the range of 1.7 to 1.9 dl/g .

Inherent Viscosity (IV) is a viscometric method for measuring molecular size. IV is based on the flow time of a polymer solution through a narrow capillary relative to the flow time of the pure solvent through the capillary.

Mineral composition may optionally include beta-tricalcium phosphate, calcium phosphate, calcium sulfate, hydroxyapatite, or a bioresorbable glass (also known as bioglass).

Bioresorbable glass fiber may optionally have oxide compositions in the following mol.% ranges:

Na2O: 11.0 - 19.0 mol%.

CaO: 9.0 – 14.0 mol%.

MgO: 1.5 – 8.0 mol%.

B2O3: 0.5 – 3.0 mol%.

Al2O3: 0 – 0.8 mol%.

P2O3: 0.1 – 0.8 mol%.

SiO2: 67 – 73 mol%.

And more preferably in the following mol.% ranges:

Na2O: 12.0 - 13.0 mol%.

CaO: 9.0 – 10.0 mol%.

MgO: 7.0 – 8.0 mol%.

B2O3: 1.4 – 2.0 mol%.

P2O3: 0.5 – 0.8 mol%.

SiO2: 68 – 70 mol%.

Additional optional glass fiber compositions have been described previously by Lehtonen TJ et al. (Acta Biomaterialia 9 (2013) 4868–4877), which is included here by reference in its entirety; such glass fiber compositions may optionally be used in place of or in addition to the above compositions .

Additional optional bioresorbable glass compositions are described in the following patent applications, which are hereby incorporated by reference as if fully set forth herein, which are owned in common with the instant application and which have inventor(s) in common: Biocompatible composite and its use (WO2010122098); and Resorbable and biocompatible fibre glass compositions and their uses (WO2010122019).

In a more preferred embodiment of the present invention, the reinforcing filler is bound to the bioabsorbable polymer such that the reinforcing effect is maintained for an extended period. Such an approach has been described in US 2012/0040002 A1 and EP 2243500B1, which discusses a composite material comprising biocompatible glass, a biocompatible matrix polymer and a coupling agent capable of forming covalent bonds.

Optionally, fibers may be aligned at an angle to the longitudinal axis (i.e. on a diagonal) such that and preferably, a majority of reinforcement fibers are aligned at an angle that is less than 90°, alternatively less than 60°, or optionally less than 45° from the longitudinal axis.

Medical Implant Composite Structure

Implant may be selected from a group that includes orthopedic pins, screws, plates, intramedullary rods, hip replacement, knee replacement, meshes, etc.

The average wall thickness in the implant is preferably in the range of 0.2 to 10 mm, more preferably in the range of 0.4 to 5 mm, more preferably in the range of 0.5 to 2 mm, and most preferably in the range of 0.5 to 1.5 mm.

The implant preferably comprises between 2-20 composite tape layers, more preferably between 2-10 layers, and most preferably between 2-6 layers.

Optionally, implant may comprise reinforcing ribs, gussets, or struts.

Rib base thickness is preferably less than 100% of the adjoining wall thickness. More preferably, thickness is less than 85%, and most preferably less than 75%. Rib base thickness is preferably more than 20% of adjoining wall thickness, more preferably more than 30%, and most preferably more than 50% of adjoining wall thickness.

Preferably, rib height is at least 2.0 times the adjoining wall thickness, more preferably at least 3.0 times the wall thickness.

Draft angle of reinforcing ribs is preferably between 0.2-0.8°, more preferably between 0.4-0.6°.

Preferably, distance between ribs is at least 2 times adjoining wall thickness. More preferably, at least 3 times adjoining wall thickness.

Preferably, reinforcing rib or other element increases bending stiffness of implant by at least 20% without increasing compressive or tensile stiffness by more than 10%.

Optionally, ribs along one axis, for example the longitudinal axis of the implant, are taller than the ribs along the perpendicular axis, for example the latitudinal axis of the implant, in order to facilitate easier insertion of the implant.

Optionally, the implant may comprise one or more bosses to accommodate screw insertion. Preferably, the boss is between 2-3 times the screw diameter for self-tapping screw applications. Boss may additionally include supportive gusses or ribs.

Optionally, one or more sides of implant may be textured.

Optionally, implant may contain continuous fibers aligned in a circular arrangement around holes, such as screw or pin holes, within the implant.

Perforated implant part walls

In some medical implants, it is desirable for there to be cellular or tissue ingrowth through the implant so as to strengthen the incorporation of the implant into the tissue and to increase compliance of the implant in physiological function. In order to further promote such ingrowth, it is beneficial to have gaps or holes in the walls of the herein described medical implant.

Preferably, if present, such perforations in implant walls comprise at least 10% of the surface area of the implant, more preferably at least 20%, at least 30%, at least 40%, or at least 50% of the surface area of the implant.

In one optional embodiment of the present invention, the implant is a screw and the fenestrations of the threading contain perforation.

In one embodiment of the present invention, the implant contains perforations between composite tapes or between the reinforcement fibers within composite tapes making up the implant.

In a preferred embodiment, a majority of perforations are between reinforcement fibers and do not penetrate reinforcement fibers.

Cages full of bone filler

In another embodiment of herein invention, the implant comprises an orthopedic implant and the implant forms a partial or full container and an osteoconductive or osteoinductive material is contained within the implant container.

In a preferred embodiment, the implant container is additionally perforated so as to allow improved bone ingrowth into the osteoconductive or osteoinductive material contained within the implant cage.

In an optional embodiment, the implant comprises an opening or door through which bone filler can be introduced and/or bone ingrowth can take place.

In an optional embodiment, the implant comprises two or more discrete parts or separate parts joined by a joint such that implant cage may be filled with bone filler material and subsequently assembled or closed to trap bone filler inside.

Framework of continuous fiber reinforced structure with non-reinforced surrounding material

Whereas continuous fiber reinforced bioabsorbable composite structures provide the optimal mechanical strength and stiffness to a medical implant, it may also be beneficial in certain cases to have additional features or layers in the medical implant that cannot be made from continuous fiber reinforced composite tapes. In such cases, the mechanical strength of the continuous fiber reinforced bioabsorbable composite structures can be incorporated into the implant but additional sections or layers of non-reinforced polymer may be added to improve or customize the implant. These

sections or layers are preferably added to the implant either by overmolding onto the structure or by 3-D printing onto the structure.

In one embodiment of the present invention, medical implant comprises a structural support comprised of a continuous fiber-reinforced bioabsorbable composite material and additionally comprises a section or layer comprised of non-reinforced polymer material.

Optionally the second layer functions as a bone interface layer comprised of a non-reinforced absorbable polymer material. Also optionally the structural support and non-reinforced polymer section are each fabricated using a different production technique. Also optionally the structural support is fabricated by machining, compression molding, or composite flow molding and the interface layer is fabricated by injection molding or 3D printing; optionally the interface layer is fabricated on top of the prefabricated structural support.

Optionally the non-reinforced polymer section is a bone interface layer and dimensions of the interface layer are partially or entirely determined by the bone geometry of a specific patient or patient population.

Optionally the bone geometry of patient or patient population is determined by measuring through imaging technique such as X-Ray, CT, MRI.

Optionally the elastic modulus and/or flexural strength of structural support is at least 20% greater than that of the non-reinforced polymer section.

Optionally, continuous-fiber reinforced composite material in implant is coated with a polymer resin wherein the polymer resin on fiber in the composite material has a higher or lower melting temp than the flowable matrix resin; or polymer resin on fiber

has slower or faster degradation rate than flowable matrix resin; or polymer resin on fiber is more hydrophobic or more hydrophilic than flowable matrix resin

In an optional embodiment, an additional section or layer is comprised of a reinforced polymer but where polymer is reinforced by non-continuous fibers, preferably fibers less than 10mm in length, and more preferably less than 5mm in length.

In an optional embodiment, an additional section or layer of non-reinforced or non-continuous fiber reinforced polymer additional comprises an additive.

Optionally, additive comprises an osteoconductive material or combination of osteoconductive materials such as beta tricalcium phosphate, calcium phosphate, hydroxyapatite, decellularized bone.

Optionally, the additive comprises an anti-microbial agent or bone inducing agent.

Production Method

Continuous-fiber reinforced bioabsorbable implants may optionally be produced using any method known in the art. Methods can include compression molding, injection molding, extrusion, machining, or any combination of these methods.

Preferably, moisture content of implant following production is less than 50%, more preferably less than 1%, even more preferably less than 0.4%, 0.2%.

Low moisture content is important so as to avoid degradation of the implant during storage.

Preferably, residual monomer content in implant following production is less than 3%, preferably less than 2%, and more preferably less than 1%.

Without wishing to be limited by a single hypothesis, where mineral content is high relative to biocomposite implants, it is particularly important that the polymer component be predominantly comprised of polymer, with very low monomer component, since the monomer component does not contribute to the mechanical function of the implant.

Implant contact with surrounding tissue

In an optional embodiment of the present invention, less than 100% of implant surface area is in contact with surrounding tissue. This may be clinically desirable for several reasons:

1. Reduced friction with surrounding tissue upon insertion, easing insertion
2. Reduced bone contact can reduce interference to bone surface blood flow

In a preferred embodiment, implant contains surface protrusion elements of at least 0.1 mm in height and less than 2 mm in height that come into contact with tissue surrounding implant.

Preferably, total percentage of surface area of implant that comes into contact with surrounding tissue is less than 80%, more preferably less than 60%, 50% , 40%, 30%.

Balloons

In an optional embodiment of herein invention, implant additionally comprises a balloon. Balloon walls are preferably comprised of between 1-3 layers of reinforced composite.

Fabrication of the Implant

Any of the above-described bioabsorbable polymers or reinforced bioabsorbable polymers may be fabricated into any desired physical form for use with the present invention. The polymeric substrate may be fabricated for example, by compression molding, casting, injection molding, pultrusion, extrusion, filament winding, composite flow molding (CFM), machining, or any other fabrication technique known to those skilled in the art. The polymer may be made into any shape, such as, for example, a plate, screw, nail, fiber, sheet, rod, staple ,clip, needle, tube, foam, or any other configuration suitable for a medical device.

Load-bearing mechanical strength

The herein invention particularly relates to bioabsorbable composite materials that can be used in medical applications that require high strength and a stiffness compared to the stiffness of bone. These medical applications require the medical implant to bear all or part of the load applied by or to the body and can therefore be referred to generally as "load-bearing" applications. These include fracture fixation, tendon reattachment, joint replacement, spinal fixation, and spinal cages.

The flexural strength preferred from the herein described load-bearing medical implant is at least 100 MPa, preferably above 400 MPa, more preferably above 600 MPa, and even more preferably above 800 MPa. The Elastic Modulus (or Young's Modulus) of the bioabsorbable composite for use with herein invention is preferably at least 6 GPa, more preferably above 15 GPa, and even more preferably above 20 GPa but not exceeding 100 GPa and preferably not exceeding 60 GPa.

Sustained mechanical strength

There is a need for the bioabsorbable load-bearing medical implants of the herein invention to maintain their mechanical properties (high strength and stiffness) for an extended period to allow for sufficient bone healing. The strength and stiffness preferably remains above the strength and stiffness of cortical bone, approximately 150-250 MPa and 15-25 GPa respectively, for a period of at least 3 months, preferably at least 6 months, and even more preferably for at least 9 months *in vivo* (i.e. in a physiological environment).

More preferably, the flexural strength remains above 400 MPa and even more preferably remains above 600 MPa.

In another embodiment of the present invention, the mechanical strength degradation rate of the medical implant approximates the material degradation rate of the implant, as measured by weight loss of the biodegradable composite.

In a preferred embodiment, the implant retains greater than 50% of its mechanical strength after 3 months of implantation while greater than 50% of material degradation and hence weight loss occurs within 12 months of implantation.

In a preferred embodiment, the implant retains greater than 70% of its mechanical strength after 3 months of implantation while greater than 70% of material degradation and hence weight loss occurs within 12 months of implantation.

In a preferred embodiment, the implant retains greater than 50% of its mechanical strength after 6 months of implantation while greater than 50% of material degradation and hence weight loss occurs within 9 months of implantation.

In a preferred embodiment, the implant retains greater than 70% of its mechanical strength after 6 months of implantation while greater than 70% of material degradation and hence weight loss occurs within 9 months of implantation.

The mechanical strength degradation and material degradation (weight loss) rates of the medical implant can be measured after *in vivo* implantation or after *in vitro* simulated implantation. In the case of *in vitro* simulated implantation, the simulation may be performed in real time or according to accelerated degradation standards.

"Biodegradable" as used herein is a generalized term that includes materials, for example polymers, which break down due to degradation with dispersion *in vivo.* The decrease in mass of the biodegradable material within the body may be the result of a passive process, which is catalyzed by the physicochemical conditions (e.g. humidity, pH value) within the host tissue. In a preferred embodiment of biodegradable, the decrease in mass of the biodegradable material within the body may also be eliminated through natural pathways either because of simple filtration of degradation by-products or after the material's metabolism ("Bioresorption" or "Bioabsorption"). In either case, the decrease in mass may result in a partial or total elimination of the initial foreign material. In a preferred embodiment, said biodegradable composite comprises a biodegradable polymer that undergoes a chain cleavage due to macromolecular degradation in an aqueous environment.

A polymer is "absorbable" within the meaning of this invention if it is capable of breaking down into small, non-toxic segments which can be metabolized or

eliminated from the body without harm. Generally, absorbable polymers swell, hydrolyze, and degrade upon exposure to bodily tissue, resulting in a significant weight loss. The hydrolysis reaction may be enzymatically catalyzed in some cases. Complete bioabsorption, i.e. complete weight loss, may take some time, although preferably complete bioabsorption occurs within 24 months, most preferably within 12 months.

The term "polymer degradation" means a decrease in the molecular weight of the respective polymer. With respect to the polymers, which are preferably used within the scope of the present invention said degradation is induced by free water due to the cleavage of ester bonds. The degradation of the polymers as for example used in the biomaterial as described in the examples follows the principle of bulk erosion. Thereby a continuous decrease in molecular weight precedes a highly pronounced mass loss. Said mass loss is attributed to the solubility of the degradation products. Methods for determination of water induced polymer degradation are well known in the art such as titration of the degradation products, viscometry, differential scanning calorimetry (DSC).

The term “Biocomposite” as used herein is a composite material formed by a matrix and a reinforcement of fibers wherein both the matrix and fibers are biocompatible and optionally bioabsorbable. In most cases, the matrix is a polymer resin, and more specifically a synthetic bioabsorbable polymer. The fibers are optionally and preferably of a different class of material (i.e. not a synthetic bioabsorbable polymer), and may optionally comprise mineral, ceramic, cellulosic, or other type of material.

Clinical Applications

The medical implants discussed herein are generally used for bone fracture reduction and fixation to restore anatomical relationships. Such fixation optionally and preferably includes one or more, and more preferably all, of stable fixation, preservation of blood supply to the bone and surrounding soft tissue, and early, active mobilization of the part and patient.

There are several exemplary, illustrative, non-limiting types of bone fixation implants for which the materials and concepts described according to at least some embodiments of the present invention may be relevant, as follows:

Bone Plate

A bone plate is typically used to maintain different parts of a fractured or otherwise severed bone substantially stationary relative to each other during and/or after the healing process in which the bone mends together. Bones of the limbs include a shaft with a head at either end thereof. The shaft of the bone is generally elongated and of relatively cylindrical shape.

It is known to provide a bone plate which attaches to the shaft or head and shaft of a fractured bone to maintain two or more pieces of the bone in a substantially stationary position relative to the one another. Such a bone plate generally comprises a shape having opposing substantially parallel sides and a plurality of bores extending between the opposing sides, wherein the bores are suitable for the receipt of pins or screws to attach the plate to the bone fragments.

For proper function of the bone plate in maintaining different parts of a fractured bone stationary relative to each other, the plate must be of sufficient mechanical strength and stiffness to maintain the position of the bone fragments or pieces. However, it must achieve these mechanical properties within a low profile thickness profile to ensure that there will be sufficient space for the bone plate to fit between bone and the surrounding soft tissue. The thickness of the bone plate is generally in the range of 2.0 mm to 8.0 mm and more commonly in the range of 2.0 mm to 4.0 mm. The widths of the plates are variable but

Screws

Screws are used for internal bone fixation and there are different designs based on the type of fracture and how the screw will be used. Screws come in different sizes for use with bones of different sizes. Screws can be used alone to hold a fracture, as

well as with plates, rods, or nails. After the bone heals, screws may be either left in place or removed.

Screws are threaded, though threading can be either complete or partial. Screws can include compression screws, locking screws, and/or cannulated screws. External screw diameter can be as small as 0.5 or 1.0 mm but is generally less than 3.0mm for smaller bone fixation. Larger bone cortical screws can be up to 5.0mm and cancellous screws can even reach 7-8 mm. Some screws are self-tapping and others require drilling prior to insertion of the screw. For cannulated screws, a hollow section in the middle is generally larger than 1mm diameter in order to accommodate guide wires.

Wires/Pins

Wires are often used to pin bones back together. They are often used to hold together pieces of bone that are too small to be fixed with screws. They can be used in conjunction with other forms of internal fixation, but they can be used alone to treat fractures of small bones, such as those found in the hand or foot. Wires or pins may have sharp points on either one side or both sides for insertion or drilling into the bone.

"K-wire" is a particular type of wire generally made from stainless steel, titanium, or nitinol and of dimensions in the range of 0.5 – 2.0 mm diameter and 2-25 cm length. "Steinman pins" are general in the range of 2.0 – 5.0 mm diameter and 2-25 cm length. Nonetheless, the terms pin and wire for bone fixation are used herein interchangeably.

Anchors

Anchors and particularly suture anchors are fixation devices for fixing tendons and ligaments to bone. They are comprised of an anchor mechanism, which is inserted into the bone, and one or more eyelets, holes or loops in the anchor through which the suture passes. This links the anchor to the suture. The anchor which is inserted into the bone may be a screw mechanism or an interference mechanism. Anchors are generally in the range of 1.0 – 6.5 mm diameter

Cable, ties, wire ties

Cables, ties, or wire ties can be used to perform fixation by cerclage, or binding, bones together. Such implants may optionally hold together bone that cannot be fixated using penetration screws or wires/pin, either due to bone damage or presence of implant shaft within bone. Generally, diameter of such cable or tie implants is optionally in the range of 1.0 mm – 2.0 mm and preferably in the range of 1.25 – 1.75 mm. Wire tie width may optionally be in the range of 1 – 10 mm.

Nails or Rods

In some fractures of the long bones, medical best practice to hold the bone pieces together is through insertion of a rod or nail through the hollow center of the bone that normally contains some marrow. Screws at each end of the rod are used to keep the fracture from shortening or rotating, and also hold the rod in place until the fracture has healed. Rods and screws may be left in the bone after healing is complete. Nails or rods for bone fixation are generally 20-50 cm in length and 5-20 mm in diameter (preferably 9-16mm). A hollow section in the middle of nail or rod is generally larger than 1mm diameter in order to accommodate guide wires.

Any of the above-described bone fixation implants may optionally be used to fixate various fracture types including but not limited to comminuted fractures, segmental fractures, non-union fractures, fractures with bone loss, proximal and distal fractures, diaphyseal fractures, osteotomy sites, etc.

Example #1 – Large Diameter Pins

Below example describes production of large diameter orthopedic pins with reinforced biocomposite materials. This example demonstrates how different medical implant pins comprised of reinforced biocomposite materials can have different performance properties with regard to flexural modulus and strength, both at time zero

(following production) and following simulated degradation, relating to the compositional structure, geometry, and composition of each type of pin.

Materials & Methods

Three types of pin implants, each of outer diameter 6 mm and 5 cm length were produced using reinforced composite material. Material composite was comprised of PLDLA 70/30 polymer reinforced with 50% w/w, 70%, or 85% w/w continuous mineral fibers. Mineral fibers composition was approximately Na_2O 14%, MgO 5.4%, CaO 9%, B_2O_3 2.3%, P_2O_5 1.5%, and SiO_2 67.8% w/w. Testing samples were manufactured by compression molding of multiple layers of composite material into a tubular mold, either with or without a 3mm pin insert in the center. Each layer was comprised of the PLDLA polymer with embedded uni-directionally aligned continuous fibers. Orientation of layers relative to longitudinal axis of implant were 0° (parallel to implant longitudinal axis), 45°, 0°, -45°, 0°, in a repetitive manner according to number of layers in the implant. Each layer was approximately 0.18 mm thick. Three (3) pin samples were produced for each pin group.

Implant samples were tested in a tensile testing system (220Q1125-95, TestResources, MN, USA) for flexural strength, flexural modulus and maximum flexural load according to modified standard test method, ASTM D790 (Standard Test Methods for Flexural Properties of Unreinforced and Reinforced Plastics and Electrical Insulating Materials, http://www.astm.org/Standards/D790.htm, ASTM International, PA, USA). Testing was conducted initially and following simulated *in vitro* degradation according to modified ASTM F1635 (Standard Test Method for *in vitro* Degradation Testing of Hydrolytically Degradable Polymer Resins and Fabricated Forms for Surgical Implants, http://www.astm.org/Standards/F1635.htm ASTM International, PA, USA), wherein samples were incubated in simulated body fluid (SBF), 142 Na^+, 5 K^+, 1.5 Mg^{2+},2.5 Ca^{2+}, 147.8 Cl^-, 4.2 HCO_3^-, 1 HPO_4^{3-}, 0.5 SO_4^{2-} mol/m^3, for 5 days at a temperature of 50°C, while shaking at 30 rpm. Mechanical testing was performed using a 5KN load cell and an appropriate fixture for three point bending testing. Sample span was 40 mm at the beginning of the test and cross head speed was set at 2 mm/min. Dimensions, weight and density of samples were recorded.

Scanning electron microscope (SEM) (FEI Quanta FEG 250, Holland) images were captured for cross-sections of implant samples at several magnifications, with and without Au sputtering, and using either SE or BSE detectors. ImageJ™ (NIH Image Processing Software, http://www.*imagej.nih.gov/ij/*, National Institute of Health, Maryland, USA) was used to count or measure the following parameters:

1. Distance between fibers
2. Distance between layers
3. Number of fibers per layer
4. Fiber diameter
5. Tangential angle to curvature

MATLAB (http://www.mathworks.com/products/matlab/, Mathworks, MA, USA) was used to count or measure the following parameters:

1. Volume distribution of fibers within cross section of implant

Results

Table 1a shows the mechanical performance results of implant pins made from three different types of reinforced composites as described above. The structural properties of these implants are described by the production methods discussed above and their internal compositions are seen in the associated images. Quantification of several parameters related to the internal composition structure of the implants can be seen in table 1b.

Pin Type	**E [MPa]**	**Flexural Strength [MPa]**	**Max Load [N]**	**Density [gr/ml]**	**Volume [mm^3]**
Full pin. OD 6mm. 50% w/w fiber. T=0	8697.0 ± 237.8	243.7 ± 14.5	549.6 ± 57.3	1.60	1472.7
Full pin. OD	6423.5 ±	118.6 ± 16.6	267.9 ±	1.64	1480.5

6mm. 50% w/w fiber. T=5d	243.6		41.3		
Full pin. OD 6mm. 70% w/w fiber. T=0	14207.5 ± 811.7	224.6 ± 51.6	455.1 ± 130.5	1.83	1365.9
Full pin. OD 6mm. 70% w/w fiber. T=5d	6745.0 ± 677.6	85.1 ± 15.2	209.7 ± 48.6	1.78	1567.7
Hollow pin. OD 6mm. ID 3mm. 50% w/w fiber. T=0	7244.6 ± 1736.9	148.5 ± 5.4	294.0 ± 5.1	1.58	1067.4
Hollow pin. OD 6mm. ID 3mm. 50% w/w fiber. T=5d	4281.6 ± 1608.2	81.2 ± 12.5	169.6 ± 27.4	1.63	1113.1

Table 1a: Mean values and standard deviations of the mechanical properties and bulk properties of the implants (n=3).

Full pin samples produced with OD 6mm, 85% w/w fiber severely lacked in cohesive strength, likely due to insufficient amount of polymer binding between fiber layers. These samples failed during loading onto the tensile testing system and therefore mechanical property results were not recorded. Images of these pins can be seen in Figures 27 and 28, which show high amount of fibers and absence of polymer.

As can be seen in Table 1A, incubation for 5 days in SBF at 50 ^{0}C, which accelerates degradation rate, resulted in a decrease in modulus of 26%, 53% and 41% in the full 50% w/w, full 70% w/w and hollow 6mm pins respectively. Incubation for 5 days in SBF at 50 ^{0}C, which accelerates degradation rate, resulted in a decrease in flexural strength of 51%, 62% and 45% in the full 50% w/w, full 70% w/w and hollow 6mm pins respectively. Incubation for 5 days in SBF at 50 ^{0}C, which accelerates

degradation rate, resulted in a decrease in maximum flexural load of 51%, 53% and 42% in the full 50% w/w, full 70% w/w and hollow 6mm pins respectively.

	Fiber diameter range (µm)	**Distance between fibers (µm)**	**Fibers in layer thickness**	**Layer thickness (µm)**	**Distance between layers (µm)**
Full pin. OD 6mm. 50% w/w fiber	9.38 – 12.83 (Fig 1)	1.39 - 8.7 (Fig 2)	7 – 9 (Fig 3)	92.6 -185.0 (Fig 3, 4)	28.77 – 50.05 (Fig 5)
Full pin. OD 6mm. 70% w/w fiber		4.63 - 31.45 (Fig 6)	9 – 13 (Fig 7)	161.52 (Fig 7)	

Table 1b: Measured structural parameters relating the reinforcing fibers and biocomposite layers within two types of biocomposite pins.

Without wishing to be limited by a single hypothesis, it is believed that reinforcing fiber content, diameter, distribution, and arrangement into layers seen in this example (Example 1) were the cause or at least a significantly contributing factor.

Specifically with regard to reinforcing fiber, increasing reinforcing fiber content may contribute positively to mechanical properties of a medical implant, as seen by the stronger and stiffer samples produced with 70% fiber as compared with those produced with 50% fiber. However, the 70% fiber implants seemed to lose mechanical properties at a faster rate. Thus, there are potential benefits to each of these amount of fibers. Above a certain point, overly high fiber content can result in failure of the implant, as observed with the 85% fiber pins.

Example #2 – Small Diameter Pins

Below example describes production of small diameter orthopedic pins with reinforced biocomposite materials. This example demonstrates how different medical implant pins comprised of reinforced biocomposite materials can have different performance properties with regard to flexural modulus and strength, both at time zero (following production) and following simulated degradation (for example upon insertion to the body), relating to the compositional structure, geometry, and composition of each type of pin.

Materials & Methods

Three types of pin implants, each of outer diameter 2 mm and 5 cm length were produced using reinforced composite material. Material composite was comprised of PLDLA 70/30 polymer reinforced with 50% w/w or 70% w/w continuous mineral fibers. Mineral fiber composition was approximately Na_2O 14%, MgO 5.4%, CaO 9%, B_2O_3 2.3%, P_2O_5 1.5%, and SiO_2 67.8% w/w. Testing samples were manufactured by compression molding of multiple layers of composite material into a tubular mold, either with or without a 1mm pin insert in the center. Each layer was comprised of the PLDLA polymer with embedded uni-directionally aligned continuous fibers. Orientation of layers relative to longitudinal axis of implant were 0° (parallel to implant longitudinal axis), 45°, 0°, -45°, 0°, in a repetitive manner according to number of layers in the implant. Each layer was approximately 0.18 mm thick. Three (3) pin samples were produced for each pin group.

Implant samples were tested in a tensile testing system (220Q1125-95, TestResources, MN, USA) for flexural strength, flexural modulus and maximum flexural load according to modified standard test method , ASTM D790 (Standard Test Methods for Flexural Properties of Unreinforced and Reinforced Plastics and Electrical Insulating Materials, http://www.astm.org/Standards/D790.htm, ASTM International, PA, USA). Testing was conducted initially and following simulated *in vitro* degradation according to modified ASTM F1635,(Standard Test Method for *in vitro* Degradation Testing of Hydrolytically Degradable Polymer Resins and Fabricated Forms for Surgical Implants, http://www.astm.org/Standards/F1635.htm ASTM International, PA, USA) wherein samples were incubated in simulated body fluid (SBF), 142 Na^+, 5 K^+, 1.5 Mg^{2+},2.5 Ca^{2+}, 147.8 Cl^-, 4.2 HCO_3^- , 1 HPO_4^{3-}, 0.5 SO_4^{2-} mol/m^3, for 5 days at a temperature of 50°C, while shaking at 30

rpm. Mechanical testing was performed using a 500 N load cell and an appropriate fixture for three point bending testing. Sample span was 40 mm at the beginning of the test and cross head speed was set at 2 mm/min. Dimensions, weight and density of samples were recorded.

Scanning electron microscope (SEM) (FEI Quanta FEG 250, Holland) images were captured for cross-sections of implant samples at several magnifications, with and without Au sputtering, and using either SE or BSE detectors. ImageJ™ (NIH Image Processing Software, http://www.*imagej.nih.gov/ij/*, National Institute of Health, Maryland, USA) was used to count or measure the following parameters:

1. Distance between fibers
2. Distance between layers
3. Number of fibers per layer
4. Fiber diameter
5. Tangential angle to curvature

MATLAB (http://www.mathworks.com/products/matlab/, Mathworks, MA, USA) was used to count or measure the following parameters:

1. Volume distribution of fibers within cross section of implant: The percentage of fiber to polymer was calculated by summing the entire fiber area in the image divided by the area of the entire implant cross section in the image.

Percentage of Fiber to Polymer=Sum of Fiber AreaArea of Entire Cross Section*100

Results

Table 2a shows the mechanical performance results of three different types of reinforced composites implant pins produced as described above. The structural properties of these implants are described by the production methods discussed above and their internal compositions are seen in the associated images. Quantification of

several parameters related to the internal composition structure of the implants can be seen in tables 2b, c and d.

Pin Type	**E [MPa]**	**Flexural Strength [MPa]**	**Max Load [N]**	**Density [gr/ml]**	**Volume [mm^3]**
Full pin. OD 2mm. 50% w/w fiber. T=0	273.6 ± 48.3	11761.0 ± 1028.8	25.7±3.79	1.43	180.7
Full pin. OD 2mm. 50% w/w fiber. T=5d	127.2 ± 23.4	11954.9 ± 2885.5	12.45±2.4	1.37	185.88
Full pin. OD 2mm. 70% w/w fiber. T=0	290.6 ± 2.7	14062.2 ± 2158.3	30.16 ± 1.6	1.55	192.43
Full pin. OD 2mm. 70% w/w fiber. T=5d	78.9 ± 14.4	9931.5 ± 358.8	8.65 ± 1.2	1.57	201.7
Hollow pin. OD 2mm. ID 1mm. 50% w/w fiber. T=0	136.6 ± 11.7	10231.3 ± 1609.2	14.1 ± 1.1	1.37	157.6
Hollow pin. OD 2mm. ID 1mm. 50% w/w fiber. T=5d	100.1 ± 16.5	6913.7 ± 2420.1	10.35 ± 2.11	1.56	158.1

Table 2a: Mean values and standard deviations of the mechanical properties and bulk properties of the implants (n=3).

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Incubation for 5 days in SBF at 50 ^{0}C, which accelerates degradation rate, resulted in a decrease in flexural strength of 54%, 27% and 73% in the full 50% w/w, full 70% w/w and hollow 2mm pins respectively. Incubation for 5 days in SBF at 50 ^{0}C, which accelerates degradation rate, resulted in a decrease in maximum flexural load of 52%, 27% and 71% in the full 50% w/w, full 70% w/w and hollow 2mm pins respectively. Incubation for 5 days in SBF at 50 ^{0}C, which accelerates degradation rate, resulted in a decrease in flexural modulus of 32% and 29% in the full 70% w/w and hollow 2mm 50% w/w pins respectively.

	Fiber diameter range (µm)	**Distance between fibers (µm)**	**Fibers in layer thickness**	**Layer thickness (µm)**	**Distance between layers (µm)**
Full pin. OD 2mm. 50% w/w fiber	10.18-13.5 (Fig 8)	2.80-16.02 (Fig 9)	4 – 6 (Fig 10)	91.09 (Fig 10)	14.35 - 41.59 (Fig 12)
Hollow pin. OD 2mm, ID 1mm. 50% w/w fiber	11-15 (Fig 13)	2.04-10.11 (Fig 14)			11.96-33.6 (Fig 16)

Table 2b: Measured structural parameters relating the reinforcing fibers and biocomposite layers within a biocomposite pin

Area of Entire Cross Section	Sum of Fiber Area	Remaining Area	Percentage of Fiber to Polymer
22579 µm	11043 µm	1.1536e+04 µm	48.90%

Table 2c: Measured volume percentage of fiber as measured from cross-section of biocomposite full pin implant of OD 2mm, 50% w/w fiber (see Fig 11)

Area of Entire Cross Section	Sum of Fiber Area	Remaining Area	Percentage of Fiber to Polymer
14094 μm	9645.14 μm	4448.86 μm	68.43%

Table 2d: Measured volume percentage of fiber as measured from cross-section of biocomposite full plate implant of OD 2mm, ID 1mm, 50% w/w fiber (see Fig 15)

Without wishing to be limited by a single hypothesis, it is believed that reinforcing fiber content, diameter, distribution, and arrangement into layers seen in this example (Example 2) were the cause or at least a significantly contributing factor.

This example also suggests a potential structural difference between different implant part geometries (between a full pin and cannulated pin), where it is optionally possible for reinforcing fiber layers in the biocomposite implant to arrange and align themselves in differential manners depending on the shape of the implant and the forces that the implant is exposed to during its production.

Example #3 – Plates

Below example describes production of thin orthopedic plates with reinforced biocomposite materials. This example demonstrates how different medical implant plates comprised of reinforced biocomposite materials can have different performance properties with regard to flexural modulus and strength, both at time zero (following production) and following simulated degradation, relating to the compositional structure, geometry, and composition of each type of plate.

Materials & Methods

Four types of plate implants, each with a thickness of 2mm, width of 12.8mm and 6 cm length were produced using reinforced composite material. Material composite was comprised of PLDLA 70/30 polymer reinforced with 50% w/w or 70% w/w continuous mineral fibers. Mineral fibers composition was approximately Na_2O 14%, MgO 5.4%, CaO 9%, B_2O_3 2.3%, P_2O_5 1.5%, and SiO_2 67.8% w/w. Testing samples were manufactured by compression molding of multiple layers of composite material into a rectangle mold. Each layer was comprised of the PLDLA polymer

with embedded uni-directionally aligned continuous fibers. Orientation of layers relative to longitudinal axis of implant were 0° (parallel to implant longitudinal axis), 45°, 0°, -45°, 0°, in a repetitive manner according to number of layers in the implant. Each layer was approximately 0.18 mm thick. For the amorphous plates, continuous fibers were cut to small pieces, mixed and molded. Three (3) plate samples were produced for each plate group.

Implant samples were tested in a tensile testing system (220Q1125-95, TestResources, MN, USA) for flexural strength, flexural modulus and maximum flexural load according to modified standard test method , ASTM D790 (Standard Test Methods for Flexural Properties of Unreinforced and Reinforced Plastics and Electrical Insulating Materials, http://www.astm.org/Standards/D790.htm, ASTM International, PA, USA). Testing was conducted initially and following simulated *in vitro* degradation according to modified ASTM F1635,(Standard Test Method for *in vitro* Degradation Testing of Hydrolytically Degradable Polymer Resins and Fabricated Forms for Surgical Implants, http://www.astm.org/Standards/F1635.htm ASTM International, PA, USA) wherein samples were incubated in simulated body fluid (SBF), 142 Na^+, 5 K^+, 1.5 Mg^{2+},2.5 Ca^{2+}, 147.8 Cl^-, 4.2 HCO_3^- , 1 HPO_4^{3-}, 0.5 SO_4^{2-} mol/m^3, for 5 days at a temperature of 50°C, , while shaking at 30 rpm. Mechanical testing was performed using a 5 KN load cell and an appropriate fixture for three point bending testing. Sample span was 40 mm at the beginning of the test and cross head speed was set at 2 mm/min. Dimensions, weight and density of samples were recorded.

Scanning electron microscope (SEM) (FEI Quanta FEG 250, Holland) images were captured for cross-sections of implant samples at several magnifications, with and without Au sputtering, and using either SE or BSE detectors. ImageJ™ (NIH Image Processing Software,http://www.*imagej.nih.gov/ij/*, National Institute of Health, Maryland, USA) was used to count or measure the following parameters:

1. Distance between fibers
2. Distance between layers
3. Number of fibers per layer
4. Fiber diameter

5. Tangential angle to curvature

MATLAB (http://www.mathworks.com/products/matlab/, Mathworks, MA, USA) was used to count or measure the following parameters:

1. Volume distribution of fibers within cross section of implant

Results

Table 3a shows the mechanical performance results of three different types of reinforced composites implant pins produced as described above. The structural properties of these implants are described by the production methods discussed above and their internal compositions are seen in the associated images. Quantification of several parameters related to the internal composition structure of the implants can be seen in table 3b.

Plate Type	**E [MPa]**	**Flexural Strength [MPa]**	**Max Load [N]**	**Density [gr/ml]**	**Volume [mm^3]**
Plate. 50% w/w fiber. T=0	306.9 ± 13.9	15362.1 ± 502.4	285.27 ± 7.7	1.65	1624.8
Plate. 50% w/w fiber. T=5d	127.0 ± 39.1	11063.3 ± 688.8	143.5 ± 41.7	1.6	1786
Plate. 70% w/w fiber. T=0	358.5 ± 142.9	23088.4 ± 2012.5	307.56 ± 121	1.89	1552.0
Plate. 70% w/w fiber. T=5d	83.2 ± 34.3	10806.9 ± 1463.3	115.76 ± 115.8	1.7	1947.7
Plate. Amorphous 50% w/w fiber. T=0	108.1 ± 16.5	8299.7 ± 1276.9	97.4 ± 17.0	1.66	1595.1

Table 3a: Mean values and standard deviations of the mechanical properties and bulk properties of the implants (n=3).

Incubation for 5 days in SBF at 50 ^{0}C, which accelerates degradation rate, resulted in a decrease in flexural modulus of 27% and 53 % in the full 50% w/w and full 70% w/w plates respectively. Incubation for 5 days in SBF at 50 ^{0}C, which accelerates degradation rate, resulted in a decrease in flexural strength of 58 % and 76% in the full 50% w/w and full 70% w/w plates respectively.

Incubation for 5 days in SBF at 50 ^{0}C, which accelerates degradation rate, resulted in a decrease in maximum flexural load of 50 % and 62 % in the full 50% w/w and full 70% w/w plates respectively.

For this geometry and production method it seems that the increase in fiber content from 50% to 70 w/w, increases the initial mechanical strength but accelerates the degradation process.

Having short non oriented fibers as exist in the amorphous plate versus continuously oriented fibers resulted in a decrease of 46 %, 65% and 66% in the modulus, flexural strength and maximum load for a similar density and production conditions.

	Fiber diameter range (µm)	**Distance between fibers (µm)**	**Fibers in layer thickness**	**Layer thickness (µm)**	**Distance between layers (µm)**
Plate. 50% w/w fiber	11.48-13.98 (Fig 17)	2.32 - 9.88 (Fig 18)			
Plate. 70% w/w fiber		3.04 - 20 (Fig 19)	6-10 (Fig 20)	70.03-110.86 (Fig 20)	3.77-15.99 (Fig 21)

Table 3b: Measured structural parameters relating the reinforcing fibers and biocomposite layers within a biocomposite plate

Example #4 – Degradation differences

Below example describes the degradation of orthopedic implants produced with reinforced biocomposite materials. This example demonstrates how different medical implants comprised of reinforced biocomposite materials can differ in performance properties with regards to material loss and swelling ratio following simulated degradation. An absorbable orthopedic implant, used for bone fixation, as intended for the following, ideally needs to retain its strength for the period needed for the bone to heal, and then gradually degrade and lose its strength as it is replaced by bone. Material weight loss is an indication for the rate of degradation. Swelling ratio is an indication for conformational changes, hydrophilicity as well as an indication for porosity. Control of both parameters are important for implant design.

Materials & Methods

Pin and plate implants were produced using reinforced composite material as described in example 1-3. Material composite was comprised of PLDLA 70/30 polymer reinforced with 50% w/w or 70% w/w continuous mineral fibers. Mineral fibers composition was approximately Na_2O 14%, MgO 5.4%, CaO 9%, B_2O_3 2.3%, P_2O_5 1.5%, and SiO_2 67.8% w/w. Testing samples were manufactured by compression molding of multiple layers of composite material into an appropriate mold. Each layer was comprised of the PLDLA polymer with embedded uni-directionally aligned continuous fibers. Orientation of layers relative to longitudinal axis of implant were 0° (parallel to implant longitudinal axis), 45°, 0°, -45°, 0°, in a repetitive manner according to number of layers in the implant. Each layer was approximately 0.18 mm thick. Three (3) implant samples were produced for each group.

Implant samples were weighed initially and following simulated *in vitro* degradation according to a modified ASTM F1635, wherein samples were incubated in simulated body fluid (SBF), 142 Na^+, 5 K^+, 1.5 Mg^{2+},2.5 Ca^{2+}, 147.8 Cl^-, 4.2 HCO_3^- , 1 HPO_4^{3-}, 0.5 SO_4^{2-} mol/m^3, for 5 days at a temperature of 50°C, while shaking at 30 rpm. Samples were then dried in a vacuum desiccator overnight and weighed again. Material percentage loss was calculated as (initial weight -dried

weight)/initial weight *100. Swelling ratio was calculated as (weight at the end of the incubation – dried weight)/dried weight*100.

Results

Table 4 shows the weight measurement results of different types of reinforced composite implants produced as described above.

	T0 [gr]	**5 Days [gr]**	**Dried [gr]**	**Material loss (%)**	**Swelling ratio (%)**
Full pin. OD 6mm. 50% w/w	2.33 ± 0.09	2.43 ± 0.09	2.35 ± 0.09	0.245	4.42
Full pin. OD 6mm. 70% w/w	2.68 ± 0.09	2.79 ± 0.01	2.69 ± 0.01	0.262	4.35
Hollow pin. OD 6mm. ID 3mm. 50% w/w	1.69 ± 0.01	1.81 ± 0.01	1.69 ± 0.01	0.262	7.57
Full pin. OD 2mm. 50% w/w	0.257 ± 0.01	0.273 ± 0.01	0.254 ± 0.01	1.24	7.456
Full pin. OD 2mm. 70% w/w	0.281 ± 0.02	0.317 ± 0.03	0.274 ± 0.02	2.6	15.626
Hollow pin. OD 2mm. ID 1mm. 50% w/w	0.226 ± 0.03	0.246 ± 0.02	0.221 ± 0.02	2.085	11.347
Plate. 50% w/w fiber	2.755 ± 0.01	2.870 ± 0.01	2.75 ± 0.01	0.143	4.353
Plate. 70% w/w fiber	3.158 ± 0.3	3.346 ± 0.3	3.149 ± 0.25	0.312	6.237

Table 4: Mean values and standard deviations of implant weight measurements and calculated material loss and swelling ratio (n=3). Measurements are of the weight at the beginning of the experiment (T0), after degradation of 5 days in SBF at 50 ^{0}C, 30 rpm (5 days) and after dehydration in the desiccator overnight (dried).

Mineral fiber concentration increase from 50% to 70%, in the 2 mm pins and plates, increased the material loss and the swelling ratio over time by ~ 110% and more than 40% respectively. Relative degradation, as measured by relative material loss, seemed to be faster in cannulated implants vs non cannulated designs.

In the 6 mm pins, mineral fiber concentration increase from 50% to 70% also caused an increase in degradation as measured by material loss %. In the 6 mm cannulated pins, the relative degradation increase could also be noted by the increase in swelling ratio of 74% vs the full pins.

EXAMPLE 5 – Mineral Content

In the current example, biocomposite implant samples are demonstrated that comprise 50, 60 and 70% mineral content. These samples have both high mineral content and high mechanical properties.

This example further demonstrates the difference between medical implant mechanical properties at time = 0 and following 5 days of simulated bioabsorption. In many cases, the mechanical properties (including flexural modulus, flexural strength, and maximum load) of the implants with 50% mineral content were lower than the mechanical properties of the corresponding implants with higher mineral content at time = 0. However, after 5 days of simulated bioabsorption, the mechanical properties of implants with higher mineral content (60% or 70%) dropped further than the 50% mineral content implants. As such, the long term performance of the 50% mineral content implant would be improved as compared with the higher mineral content implants. However, an initially stronger implant can be achieved with higher mineral contents.

Methods & Materials

Three types of biocomposite implants were produced: pin of outer diameter 2mm, pin of outer diameter 6mm, and rectangular plates (60 x 27 x 1.5 mm). Each sample was of 7 cm length. Material composite was comprised of PLDLA 70/30 polymer reinforced with 50% w/w, 60% w/w or 70% w/w continuous mineral fibers. Mineral fibers composition was approximately Na_2O 14%, MgO 5.4%, CaO 9%, B_2O_3 2.3%, P_2O_5 1.5%, and SiO_2 67.8% w/w. Testing samples were manufactured by compression molding of multiple layers of composite material into a tubular mold or rectangular mold. Each layer was comprised of the PLDLA polymer with embedded uni-directionally aligned continuous fibers. Orientation of layers relative to longitudinal axis of implant were 0° (parallel to implant longitudinal axis), 45°, 0°, -45°, 0°, in a repetitive manner according to a number of layers in the implant. Each layer was approximately 0.18 mm thick. Three (3) samples were produced for each group. Mechanical properties were tested in a three point bending test.

Results

	2mm pins					
	E [Mpa]		**Flexural Strength [Mpa]**		**Max Load [N]**	
	T0	**5 days**	**T0**	**5 days**	**T0**	**5 days**
50%	11761.0055	11954.8476	273.5469	127.1556	25.6647	12.453
60%	17772.9284	10858.1928	339.9570	95.0317	30.5300	11.1593
70%	14062.1921	9931.4495	290.5704	78.8613	30.1587	8.6517

	Plates					
	E [Mpa]		**Flexural Strength [Mpa]**		**Max Load [N]**	
	T0	**5 days**	**T0**	**5 days**	**T0**	**5 days**
50%	15362.1439	11063.2504	306.8561	127.0402	285.2700	143.5000
70%	23088.3630	10806.9162	358.4756	83.1500	307.5633	115.7633

	6mm pins					
	E [Mpa]		Flexural Strength [Mpa]		Max Load [N]	
	T0	5 days	T0	5 days	T0	5 days
50%	8696.9920	6423.4802	243.6777	118.6093	549.6000	267.8900
70%	14207.5159	6744.9709	224.6186	85.0544	455.0700	209.7467

Example 6 - Additional drawings showing various embodiments

Figure 30 shows a continuous fiber-reinforced tape of the type that can be used to form a layer in a medical implant comprised of continuous fiber-reinforced layers. The top view (3000) shows a single strip of composite tape comprising reinforcement fibers aligned in a single direction within a bioabsorbable polymer matrix. The interspersed reinforcement fibers (3006) within the bioabsorbable polymer matrix (3008) can be seen more clearly in the close-up top view (3002) of the continuous-fiber reinforced composite tape. The reinforcement fibers can be present as separate fibers or in bundles of several reinforcement fibers per bundle. The cross-sectional view of the continuous fiber reinforced tape (3004) shows the bundles of aligned reinforcement fibers (3010) embedded within the bioabsorbable polymer matrix (3012). Fibers preferably do not breach the surface of the bioabsorbable polymer matrix.

Figure 31 shows a cut-away, three-dimensional view of a continuous fiber-reinforced tape (200). The cut-away view shows the aligned reinforcement fibers (202) embedded within the bioabsorbable polymer matrix (204).

Figure 32a shows a top-view of a reinforced bioabsorbable composite sheet (300) comprised of three layers of uni-directional fibers at different angles. Each layer can optionally be comprised of continuous fiber reinforced tapes of the type depicted in

Figure 30. The expanded view (302) shows layers of uni-directional fibers at different angles within an implant. One layer (304) aligned in the longitudinal axis, one layer (306) aligned at an angle to the right of the longitudinal axis, and one layer (308) aligned at an angle to the left of the longitudinal axis.

Figure 32b shows a cut-away view of a reinforced bioabsorbable composite structure (310) comprised of three layers of uni-directional fibers at different angles. One layer (312) aligned in the longitudinal axis, one layer (314) aligned at an angle to the right of the longitudinal axis, and one layer (316) aligned at an angle to the left of the longitudinal axis. Each layer is comprised of reinforced continuous fibers (318) embedded within bioabsorbable polymer matrix (320).

Figure 33 shows the wall of a continuous-fiber reinforced composite medical implant. The implant wall is comprised of two layers of uni-directional continuous-fiber reinforced composite tape layers (402 & 404) aligned at a perpendicular angle to each other. The medical implant wall additional comprises perforations (406) to allow for tissue penetration into or through the implant.

Figure 34 shows a bone filler cage that consists of continuous-fiber reinforced composite medical implant walls (500) that additionally contains perforations (502) to allow tissue and cellular ingrowth into the bone filler material (504) contained within the bone filler cage. The bone filler cage optionally includes a separate door to close the cage (506).

Figure 35 shows a bioabsorbable cannulated screw (600) that is a medical implant comprised of two parts: a continuous-fiber reinforced bioabsorbable composite cylindrical core (602) and bioabsorbable polymer threading (604) that was subsequently molded or 3D printed on top of the continuous-fiber core. This is an example of a bioabsorbable medical implant where a significant amount or majority of the mechanical strength is provided by a continuous-fiber reinforced part that serves as a mechanical support or structure but where additional implant features are comprised of materials that are not continuous fiber reinforced and yet can be molded or printed directly onto the fiber reinforced composite material.

Example 7 – Mineral Content

In the current example, biocomposite implant samples are demonstrated that comprise 50, 60 and 70% mineral content. These samples have both high mineral content and high mechanical properties.

Methods & Materials

Three types of biocomposite implants were produced: pin of outer diameter 2mm, pin of outer diameter 6mm, and rectangular plates (60 x 27 x 1.5 mm). Each sample was of 7 cm length. Material composite was comprised of PLDLA 70/30 polymer reinforced with 50% w/w, 60% w/w or 70% w/w continuous mineral fibers. Mineral fibers composition was approximately Na2O 14%, MgO 5.4%, CaO 9%, B 2 O 3 2.3%, P 2 O 5 1.5%, and SiO 2 67.8% w/w. Testing samples were manufactured by compression molding of multiple layers of composite material into a tubular mold or rectangular mold. Each layer was comprised of the PLDLA polymer with embedded uni-directionally aligned continuous fibers. Orientation of layers relative to longitudinal axis of implant were 0° (parallel to implant longitudinal axis), 45°, 0°, -45°, 0°, in a repetitive manner according to a number of layers in the implant. Each layer was approximately 0.18 mm thick. Three (3) samples were produced for each group. Mechanical properties were tested in a three point bending test.

Results

	2mm pins					
	E [Mpa]		Flexural Strength [Mpa]		Max Load [N]	
	T0	5 days	T0	5 days	T0	5 days
50%	11761.0055	11954.8476	273.5469	127.1556	25.6647	12.453
60%	17772.9284	10858.1928	339.9570	95.0317	30.5300	11.1593
70%	14062.1921	9931.4495	290.5704	78.8613	30.1587	8.6517

	Plates					
	E [Mpa]		Flexural Strength [Mpa]		Max Load [N]	
	T0	5 days	T0	5 days	T0	5 days
50%	15362.1439	11063.2504	306.8561	127.0402	285.2700	143.5000
70%	23088.3630	10806.9162	358.4756	83.1500	307.5633	115.7633

	6mm pins					
	E [Mpa]		Flexural Strength [Mpa]		Max Load [N]	
	T0	5 days	T0	5 days	T0	5 days
50%	8696.9920	6423.4802	243.6777	118.6093	549.6000	267.8900
70%	14207.5159	6744.9709	224.6186	85.0544	455.0700	209.7467

Example 8 – Mineral Content and Sustained Strength

In the current example, biocomposite implant samples are demonstrated that comprise 58% and 68% mineral content. These samples have both high mineral content and high mechanical properties.

Methods & Materials

Biocomposite rectangular plate implants were produced of dimensions 12.7x60x2.0mm. Material composite was comprised of PLDLA 70/30 polymer reinforced with 58% w/w or 68% w/w mineral fibers. Mineral fibers composition was approximately Na_2O 14%, MgO 5.4%, CaO 9%, B_2O_3 2.3%, P_2O_5 1.5%, and SiO_2 67.8% w/w. Testing samples were manufactured by compression molding of composite material into a rectangular mold. Reinforcing mineral fibers were of a chopped nature, with fiber segment lengths predominately in the range of 5-10mm. Plate weight was 2.75g on average for each plate. Ten (10) samples were produced for each group. Mechanical properties were tested in a three-point bending test according to ASTM D790, with 5 samples from each of the 58% and 68% group being tested at time zero (t = 0 days) and 5 samples from each of the 58% and 68% group being tested after 5 days of simulated *in vitro* degradation according to modified ASTM F1635 (t=5 days at 37 deg C, 60 rpm) in PBS.

Mechanical testing was performed using a 5 KN load cell and an appropriate fixture for three-point bending testing. Sample span was 40 mm at the beginning of the test and cross head speed was set at 2 mm/min. Dimensions and weight of the samples were recorded.

Results

2mm 58% fiber plate from chopped raw material		
	Average Maximum load [N]	Average Flexural strength [MPa]
T0	59.6 +/- 12.8	70.35
T5	40.9 +/- 7.7	48.28

2mm 68% fiber plate from chopped raw material		
	Average Maximum load [N]	Average Flexural strength [MPa]
T0	53.3 +/- 7.1	62.97
T5	30.9 +/- 4.4	36.49

58% mineral plates had a slightly higher Flexural strength at T0 than 68% plates. After 5 days in PBS Solution under the same conditions the Flexural strength of the 58% plate decreased by 32% while the flexural strength of the 68% plate decreased by 42%. Though this test was performed after only a few days of simulated degradation, there is a clear trend to suggest that increasing fiber contents above 60 % will reduce flexural strength and increase the mechanical strength loss rate over time.

It will be appreciated that various features of the invention which are, for clarity, described in the contexts of separate embodiments may also be provided in combination in a single embodiment. Conversely, various features of the invention which are, for brevity, described in the context of a single embodiment may also be provided separately or in any suitable sub-combination. It will also be appreciated by persons skilled in the art that the present invention is not limited by what has been particularly shown and described hereinabove. Rather the scope of the invention is defined only by the claims which follow.

Throughout the specification and claims, unless the context requires otherwise, the word “comprise” or variations such as “comprises” or “comprising”, will be understood to imply the inclusion of a stated integer or group of integers but not the exclusion of any other integer or group of integers.

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What is claimed is:

1. A medical implant comprising a biocomposite, said biocomposite comprising a polymer and a plurality of reinforcement mineral fibers, wherein a weight percentage of a mineral composition within the biocomposite medical implant is in the range of 40-65%, wherein an average diameter of said fibers is in a range of 3-30 microns; wherein the reinforcing fibers are fiber segments with an average fiber segment length in the range of 0.5-20 mm; wherein a residual monomer content in the medical implant following production is less than 3%; wherein the mineral composition is provided by a reinforcing mineral fiber made from the mineral composition.

2. The implant of claim 1, wherein said implant comprises a silica-based mineral compound, optionally, wherein said silica-based mineral compound has at least one oxide composition in at least one of the following mol. % ranges:

Na_2O: 11.0-19.0mol.%

CaO: 9.0 -14.0 mol.%

MgO: 1.5 - 8.0 mol.%

B_2O_3: 0.5 - 3.0 mol.%

Al_2O_3: 0 - 0.8 mol.%

P_2O_3: 0.1 - 0.8 mol.%

SiO_2: 67 - 73 mol.%; or

wherein said silica-based mineral compound has at least one oxide composition in at least one of the following mol. % ranges:

Na_2O: 12.0 - 13.0 mol.%

CaO: 9.0 - 10.0 mol.%

MgO: 7.0 - 8.0 mol.%

B_2O_3: 1.4 - 2.0mol.%

P_2O_3: 0.5 - 0.8mol.%

SiO_2: 68 - 70 mol.%.

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3. The implant of claim 1 or claim 2, wherein the density of the biocomposite composition is between 1 to 2 g/mL, between 1.2 to 1.9 g/mL, or between 1.4 to 1.8 g/mL.

4. The implant of any one of claims 1-3, wherein the diameter of a majority of reinforcing fibers is in the range of 5-20 μm, or wherein the diameter is in the range of 4-16 μm or wherein the diameter is in the range of 9-14 μm.

5. The implant of any one of claims 1-4, wherein the average fiber segment length is in the range of 1-15 mm, 3-10 mm, or 4-8 mm.

6. The implant of any one of claims 1-5, wherein said biocomposite comprises mineral fibers which are embedded in a polymer matrix; wherein said polymer comprises lactide, glycolide, caprolactone, valerolactone, carbonates, trimethylene carbonate, tetramethylene carbonate, dioxanones, 1,4- dioxanone, δ-valerolactone, l,dioxepanones,1,4-dioxepan-2-one and 1,5-dioxepan-2-one, ethylene glycol, ethylene oxide, esteramides, γ- ydroxyvalerate, β-hydroxypropionate, alpha-hydroxy acid, hydroxybuterates, poly (ortho esters), hydroxy alkanoates, tyrosine carbonates, polyimide carbonates, polyimino carbonates poly (bisphenol A-iminocarbonate), poly (hydroquinone-iminocarbonate, polyurethanes, polyanhydrides, polymer drugs, polydiflunisol, polyaspirin, protein therapeutics, sugars; starch, cellulose and cellulose derivatives, polysaccharides, collagen, chitosan, fibrin, hyaluronic acid, polypeptides, proteins, poly (amino acids), polylactides (PLA), poly-L-lactide (PLLA), poly-DL-lactide (PDLLA); polyglycolide (PGA); copolymers of glycolide, glycolide/trimethylene carbonate copolymers (PGA/TMC); other copolymers of PLA, lactide/tetramethylglycolide copolymers, lactide/trimethylene carbonate copolymers, lactide/d-valerolactone copolymers, lactide/c-caprolactone copolymers, L-lactide/DL-lactide copolymers, glycolide/L-lactide copolymers (PGA/PLLA), polylactide-co-glycolide; terpolymers of PLA, lactide/glycolide/trimethylene carbonate terpolymers, lactide/glycolide/ ε-caprolactone terpolymers, PLA/polyethylene oxide copolymers; polydepsipeptides; unsymmetrically - 3,6-substituted poly-1,4-dioxane-2,5-diones; polyhydroxyalkanoates; polyhydroxybutyrates (PHB); PHB/b-hydroxyvalerate copolymers (PHB/PHV); poly-b-hydroxypropionate (PHPA); poly-p-dioxanone (PDS); poly-d-valerolactone - poly-ε-capralactone,

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poly(ε-caprolactone-DL-lactide) copolymers; methylmethacrylate-N-vinyl pyrrolidone copolymers; polyesteramides; polyesters of oxalic acid; polydihydropyrans; polyalkyl- 2-cyanoacrylates; polyurethanes (PU); polyvinylalcohol (PVA); polypeptides; poly-b-malic acid (PMLA): poly-b-alkanbic acids; polycarbonates; polyorthoesters; polyphosphates; poly(ester anhydrides); and mixtures thereof; copolymers and mixtures thereof.

7. The implant of claim 6, wherein said polymer is selected from the group consisting of PLLA, PDLA, PGA, PLGA, PCL, PLLA-PCL and a combination thereof, optionally wherein said PLLA is used in said polymer matrix and said matrix comprises at least 30% PLLA, at least 50% PLLA, or at least 70% PLLA, and/or wherein said PDLA is used in said polymer matrix and said matrix comprises at least at least 5% PDLA, at least 10% PDLA, or at least 20% PDLA.

8. The implant of any one of claims 1-7, wherein the implant comprises a plurality of layers, each layer has a directional fiber orientation, and wherein said fiber orientation alternates between adjacent layers such that each adjacent layer is of a different angle, wherein said angle difference between layers is between 15 to 75 degrees, 30 to 60 degrees, or 40 to 50 degrees.

9. The implant of any one of claims 1-8, wherein the implant has a flexural modulus exceeding 12 GPa and flexural strength exceeding 180 MPa after 5 days of simulated physiological degradation, or wherein the implant has a flexural modulus exceeding 10 GPa and flexural strength exceeding 120 MPa after 5 days of simulated physiological degradation.

10. The implant of any one of claims 1-9, wherein said implant has strength retention of Elastic Modulus above 10 GPa after 8 weeks implantation and flexural strength above 150 MPa after 8 weeks implantation.

11. The implant of any one of claims 1-10, wherein a moisture content of the implant following production is less than 1%, less than 0.4%, or less than 0.2%.

12. The implant of any one of claims 1-11, wherein said residual monomer content is less than 2% or less than 1%.

13. The implant of any one of claims 1-12, wherein the implant is selected from the groups including bone fixation plates, intramedullary nails, joint (hip, knee, elbow) implants, spine

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implants, and other devices for such applications such as for fracture fixation, tendon reattachment, spinal fixation, and spinal cages.

14. The implant of claim 13, adapted to a threaded implant.

15. The implant of claim 14, wherein an outer layer of the threaded implant is directionally aligned such that a direction of the fibers approximates a helix angle of threading of the threaded implant.

16. A method of treatment for an orthopedic application in a subject in need of treatment thereof, comprising implanting to the subject the medical implant of any one of claims 1-15.

17. The method of treatment of claim 16, wherein said implanting to the subject comprises performing structural fixation for a load-bearing purpose within the subject.

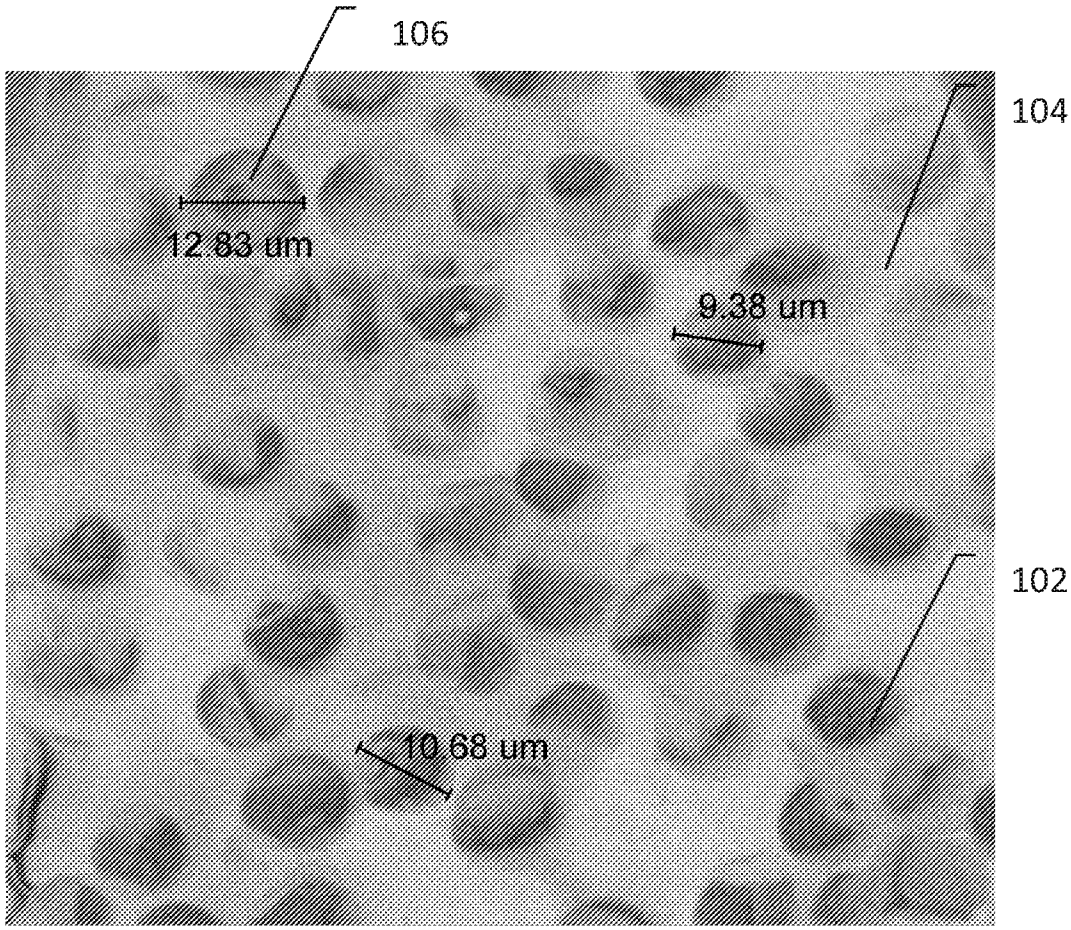

Fig 1

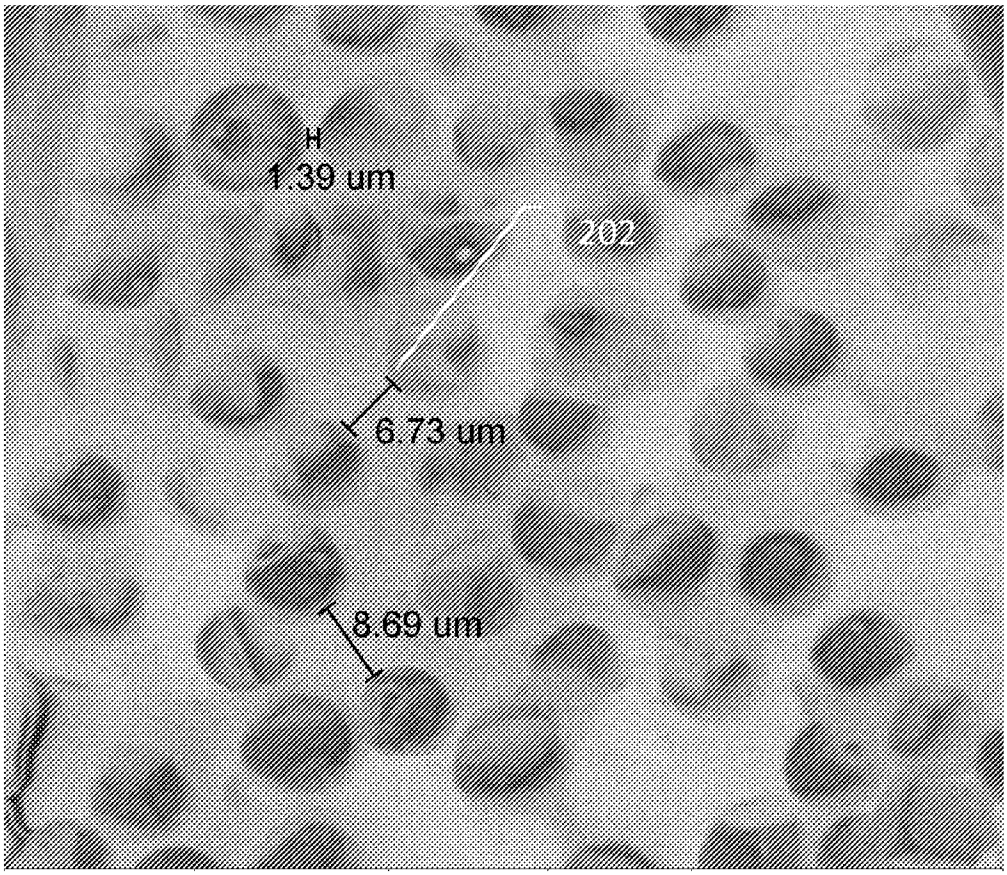

Fig 2

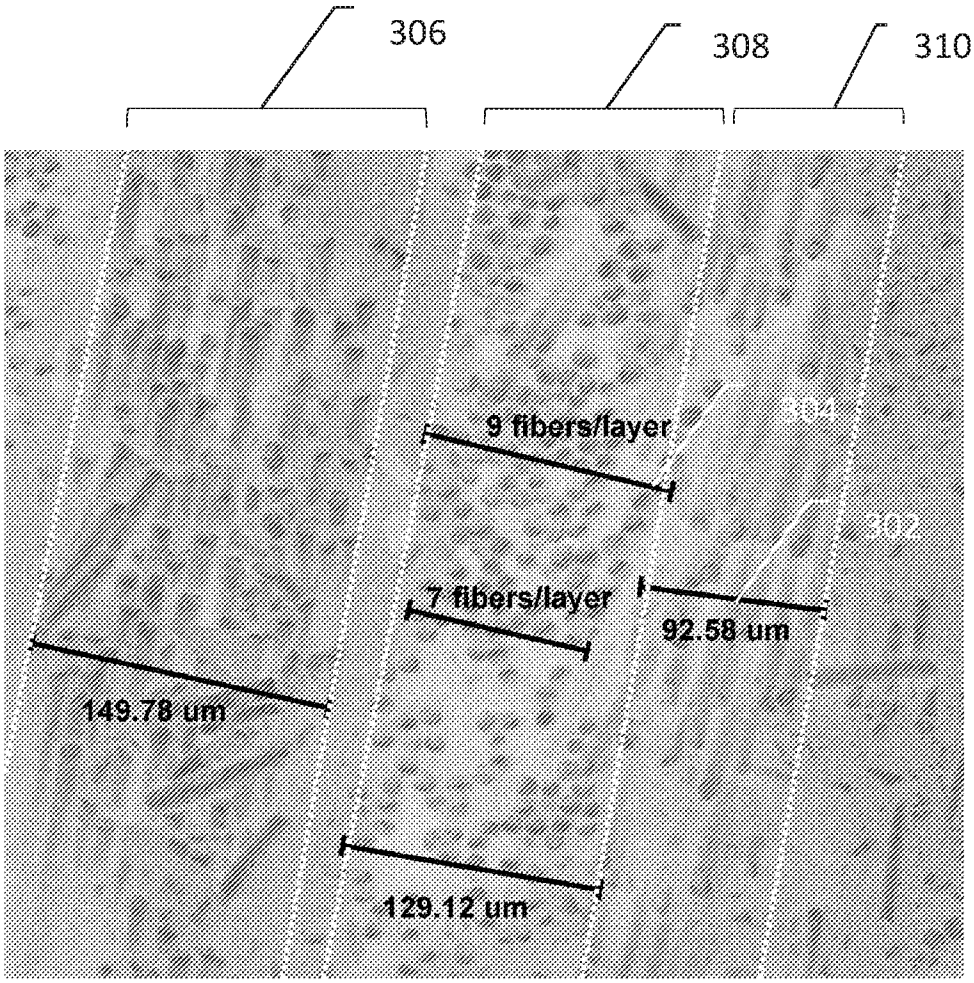
306
308
310
9 fibers/layer
304
302
7 fibers/layer
92.58 um
149.78 um
129.12 um

Fig 3

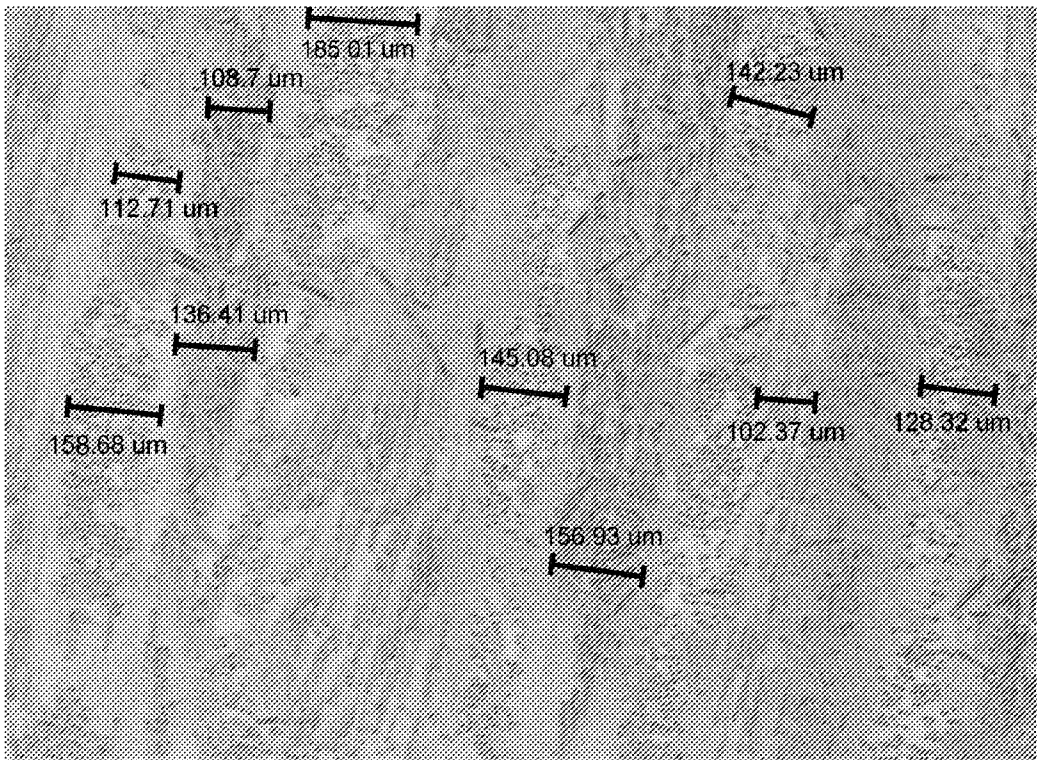
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Fig 4

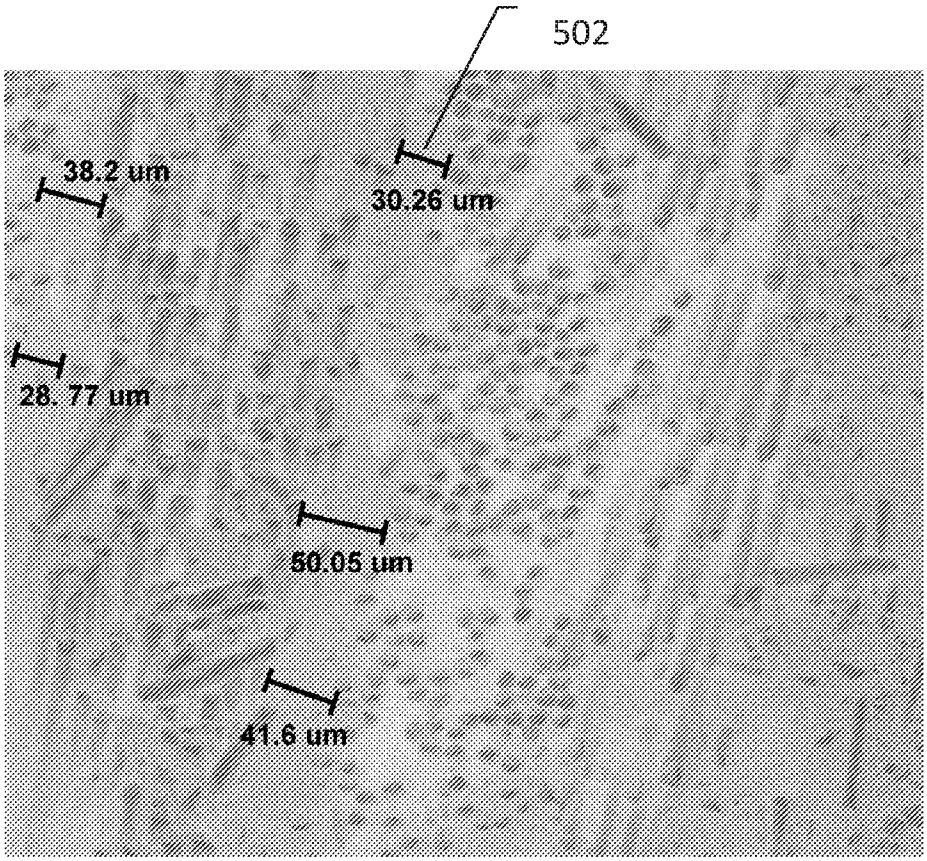


Fig 5

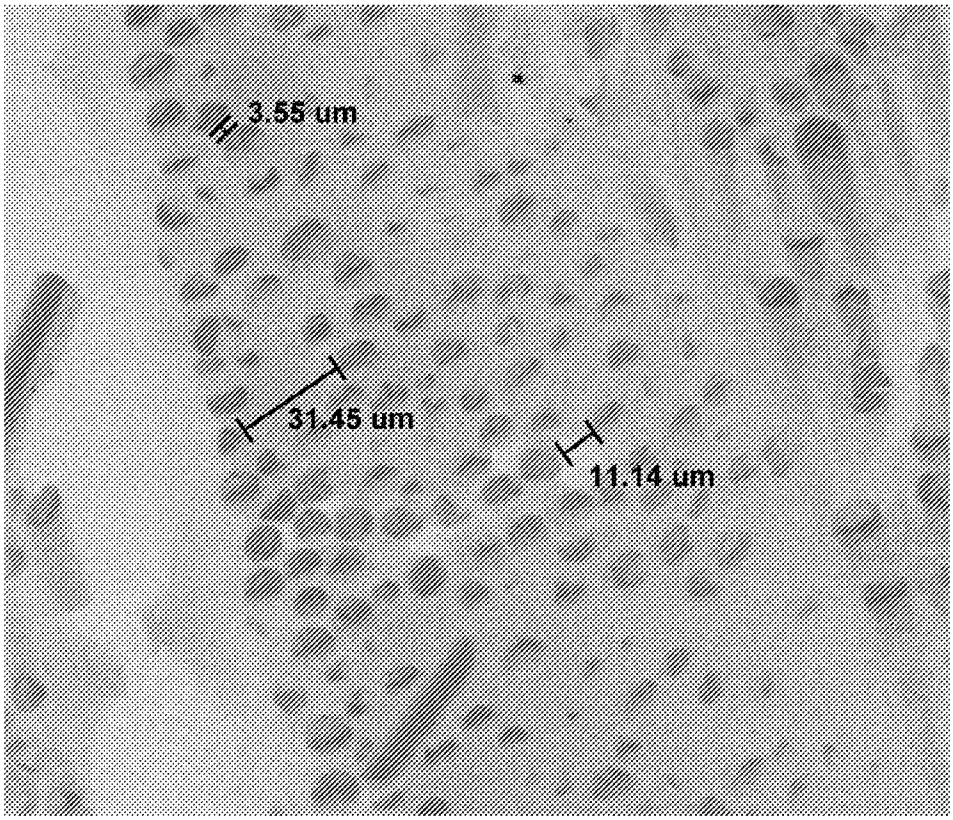


Fig 6

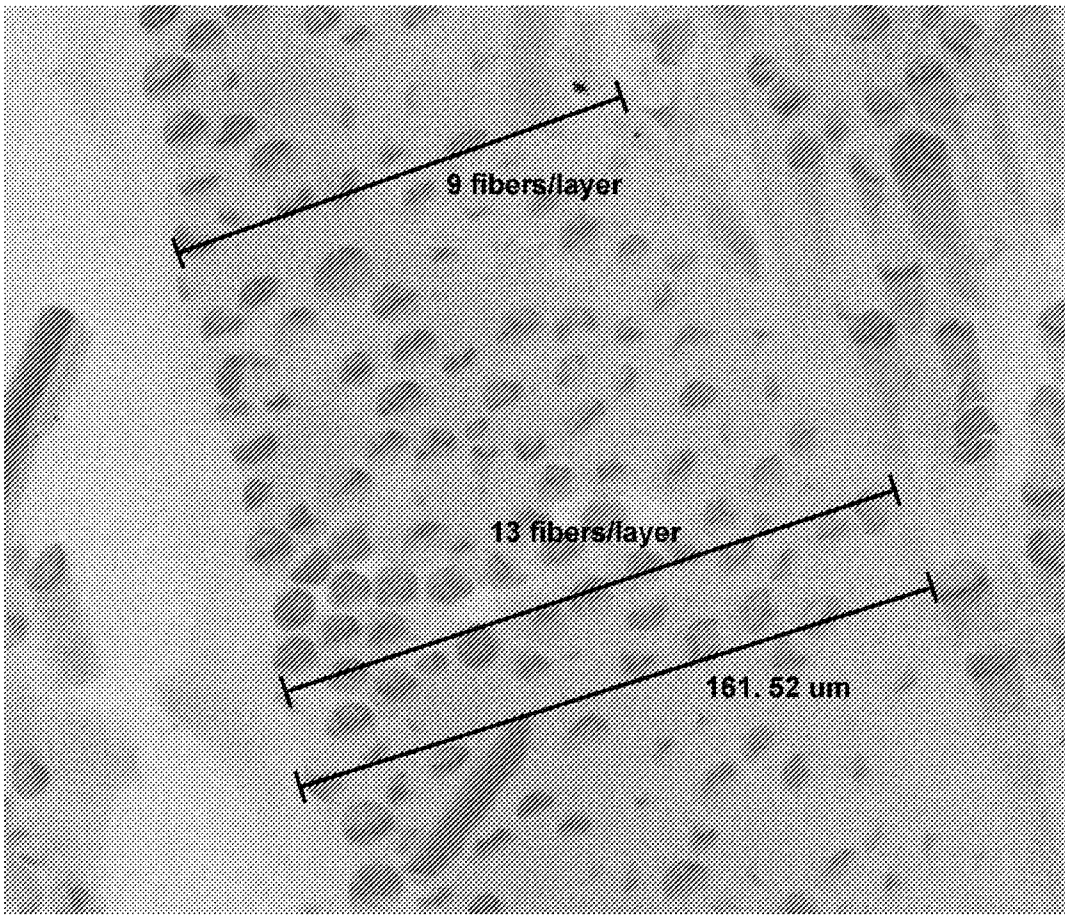


Fig 7

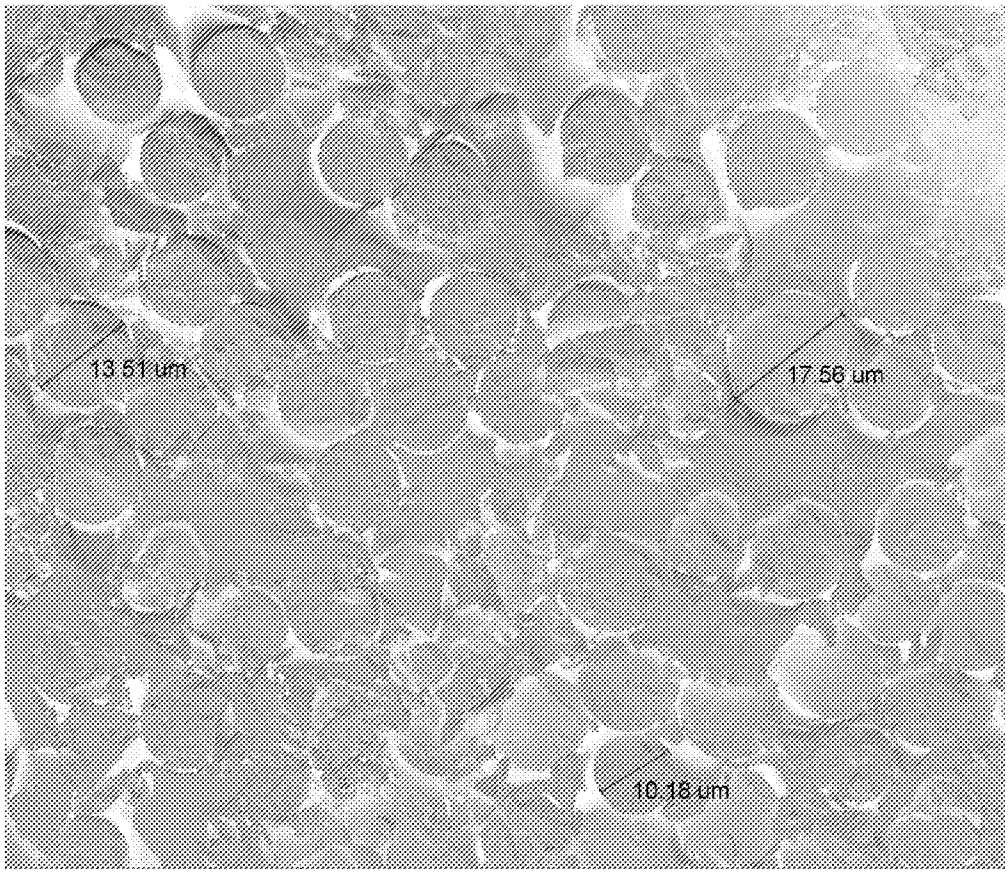


Fig 8

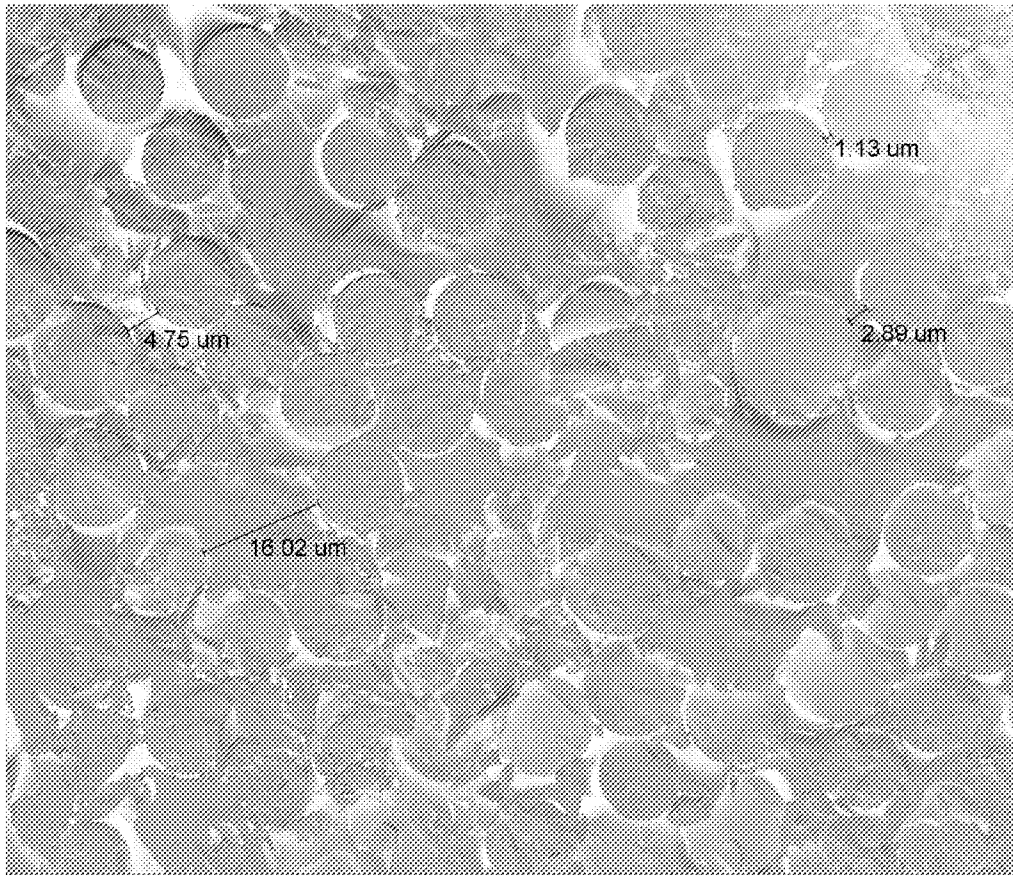


Fig 9

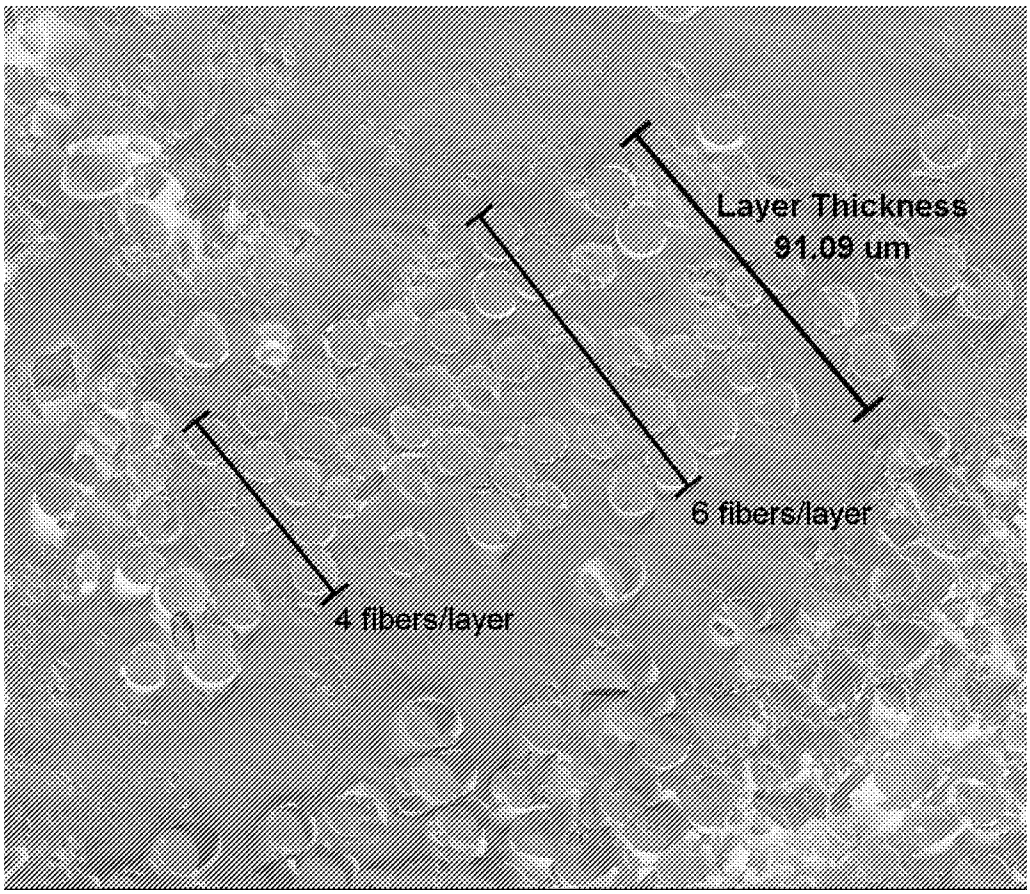


Fig 10

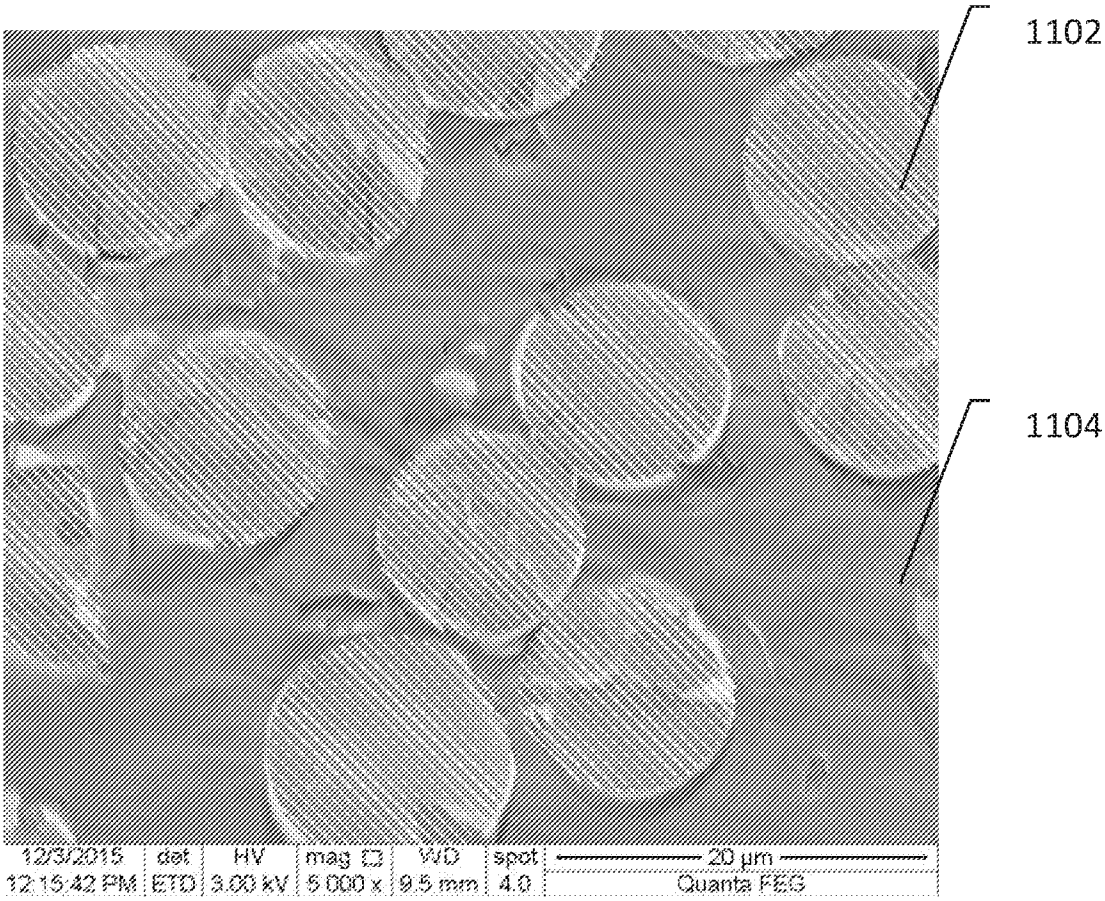

1102
1104
12/3/2015 12:15:42 PM
det ETD
HV 3.00 kV
mag 5 000 x
WD 9.5 mm
spot 4.0
20 µm
Quanta FEG

Fig 11

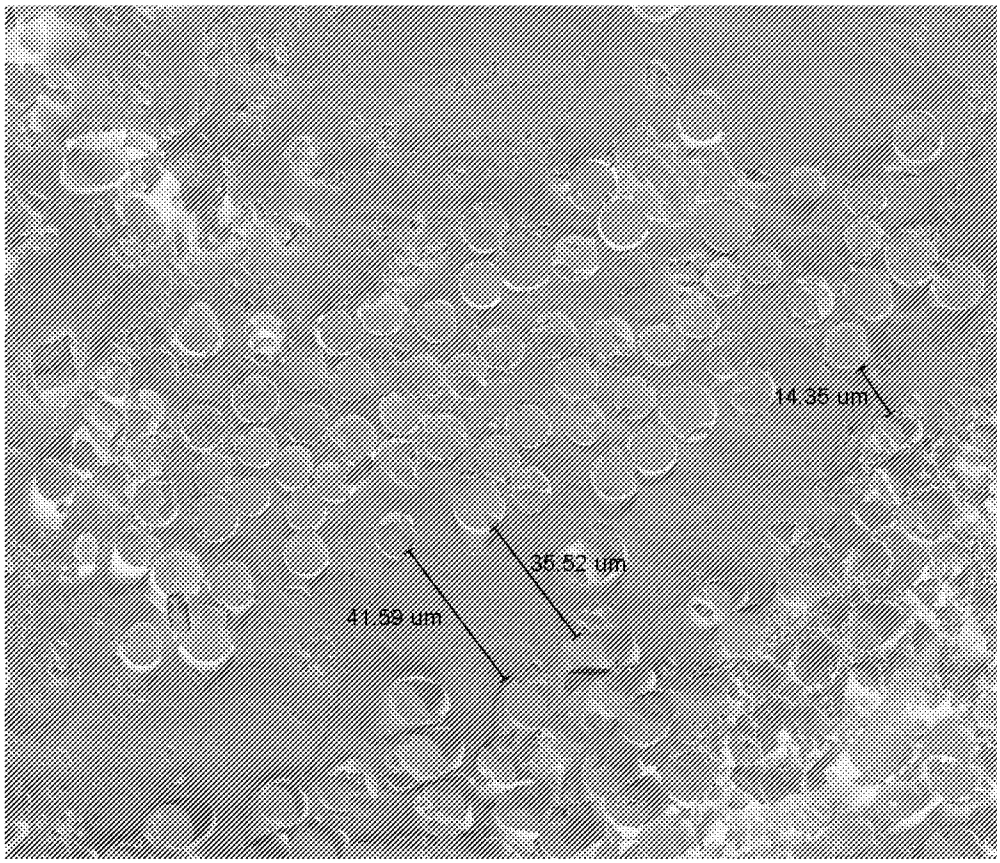


Fig 12

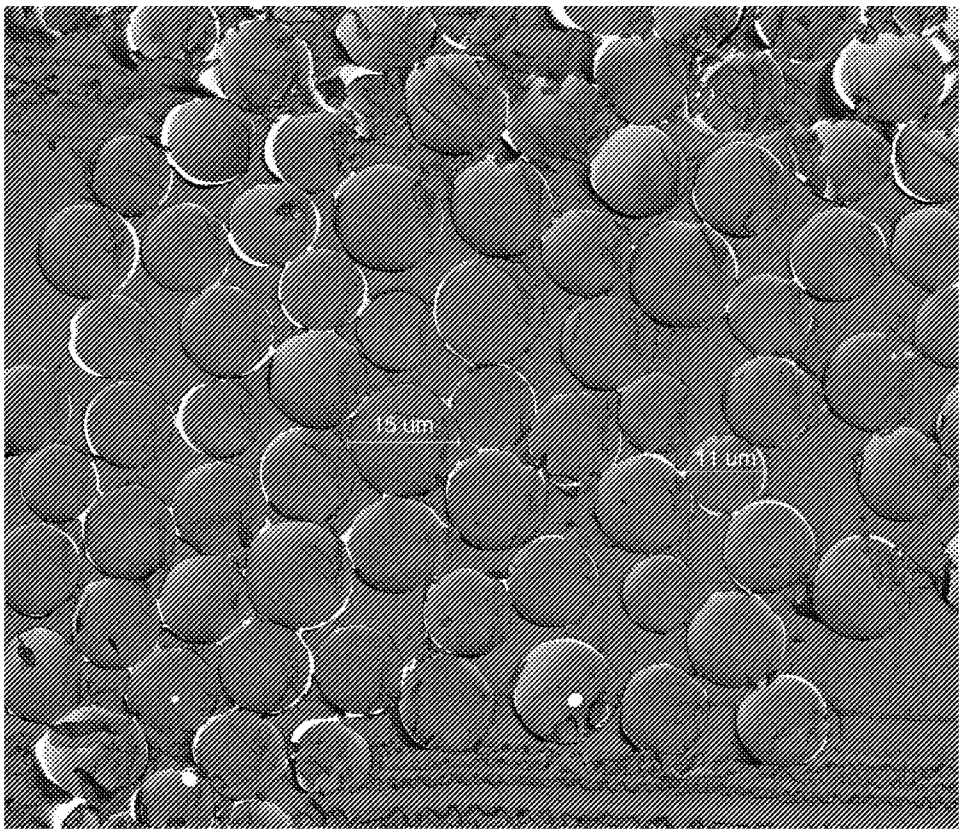


Fig 13

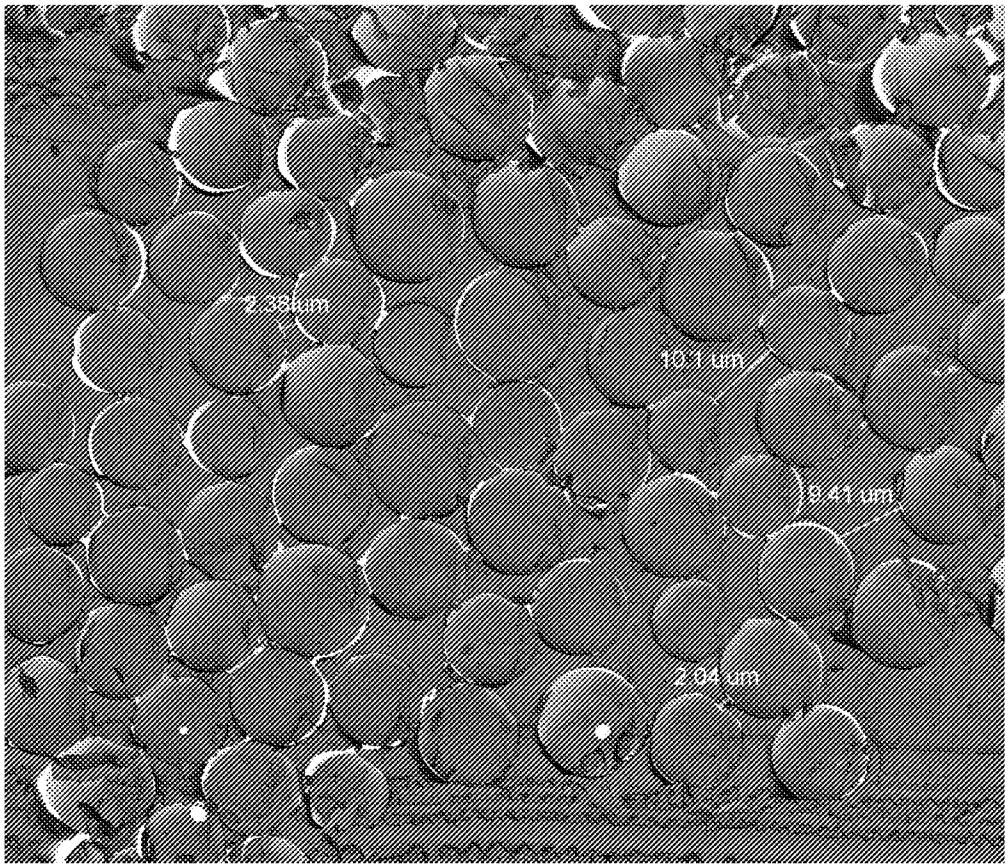


Fig 14

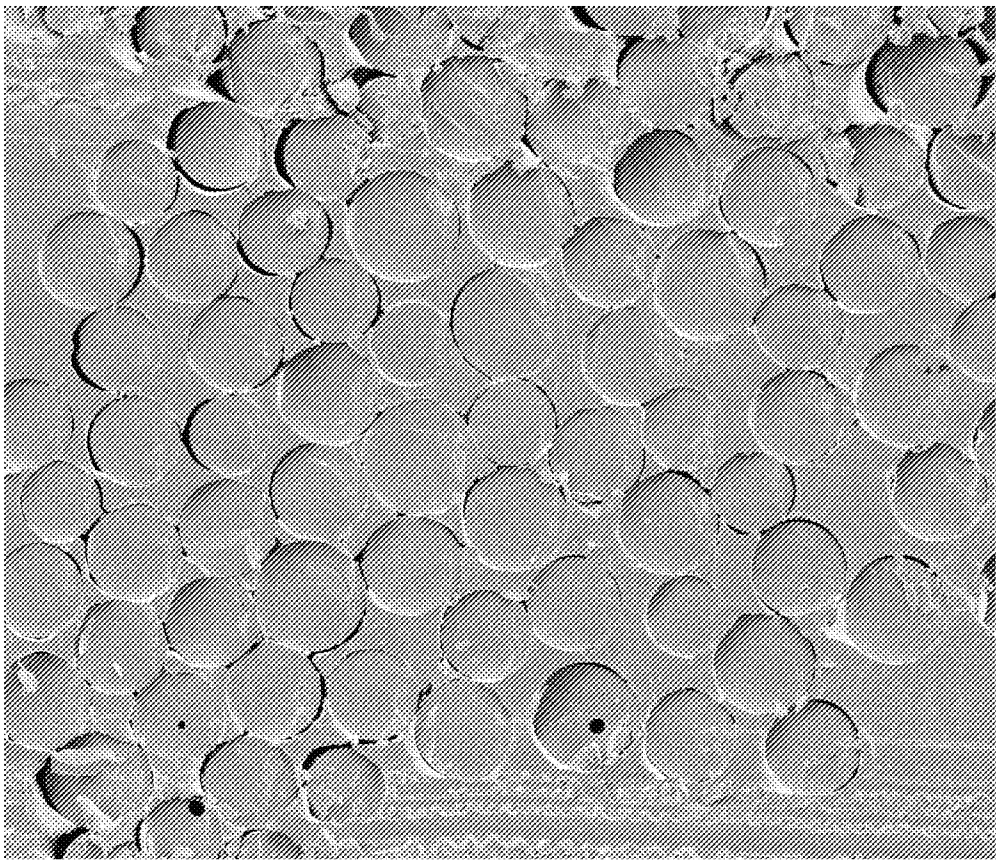


Fig 15

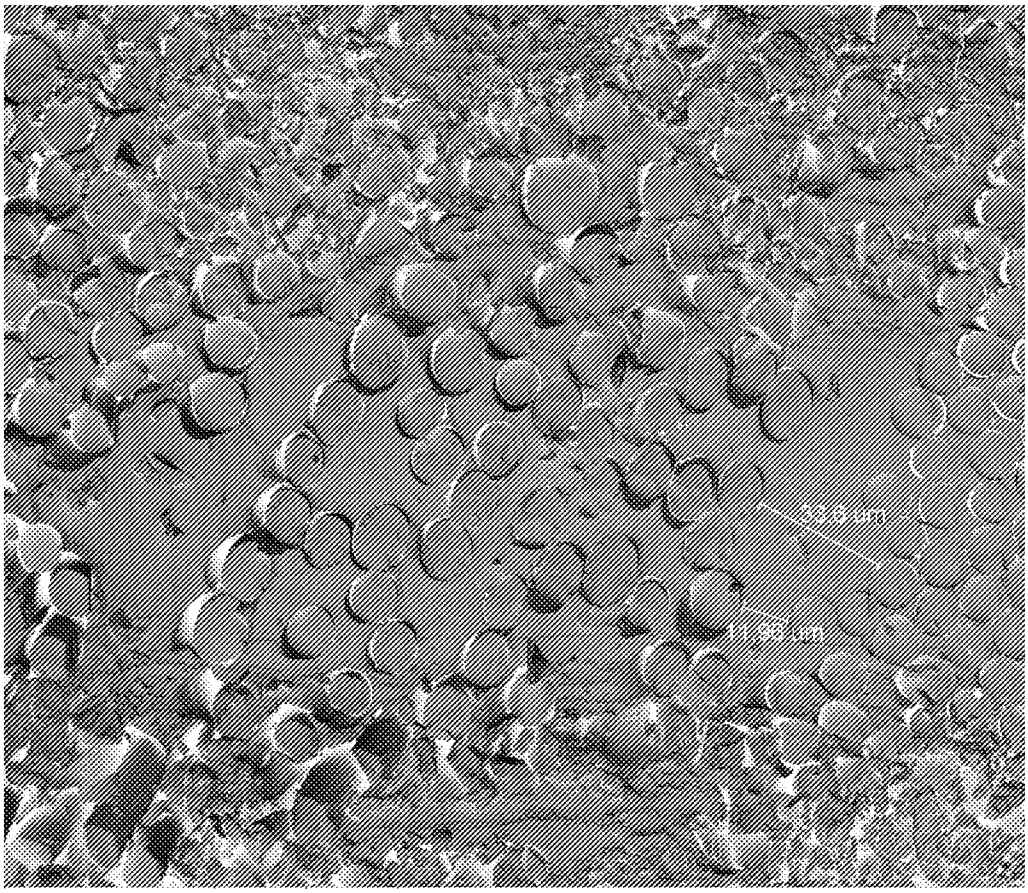


Fig 16

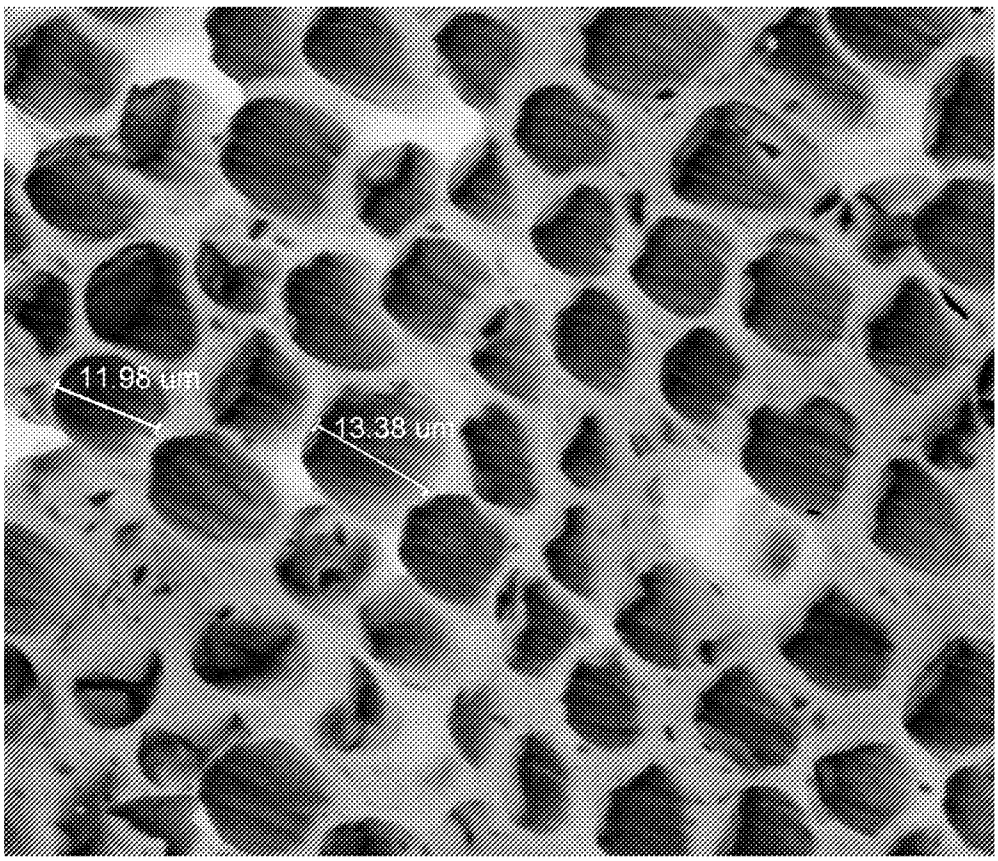


Fig 17

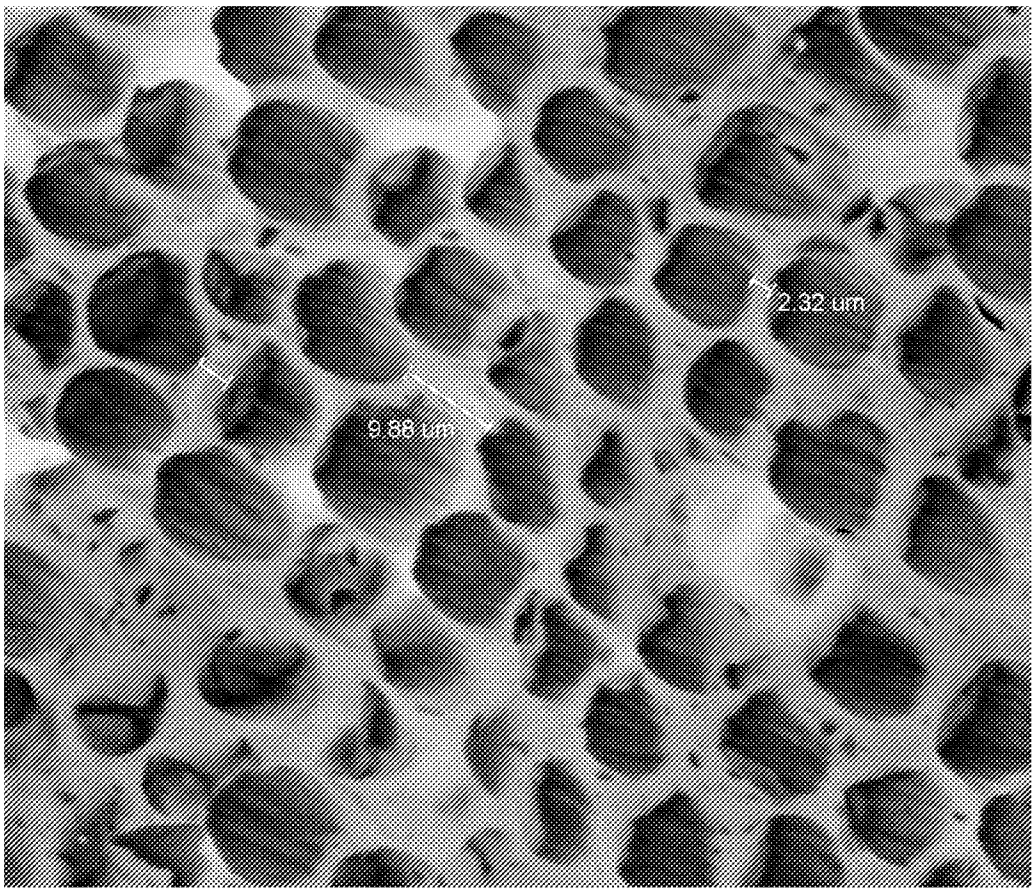


Fig 18

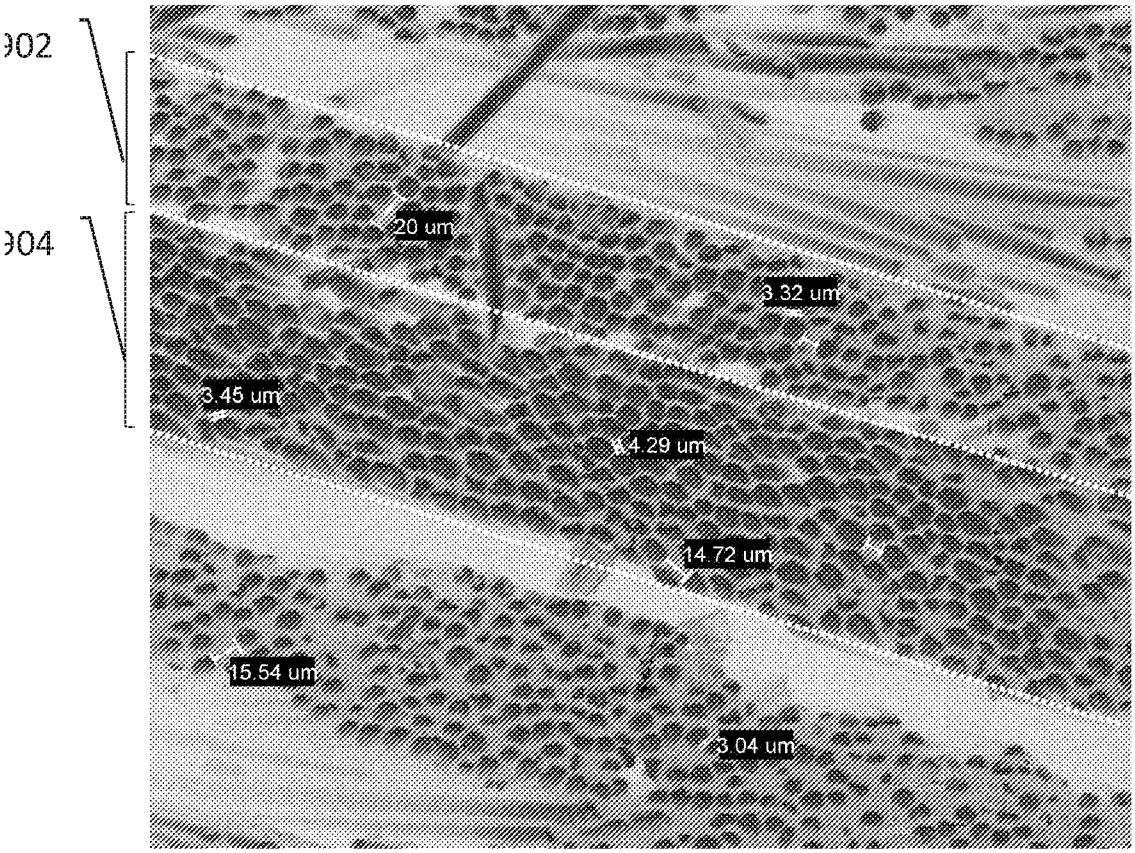


Fig 19

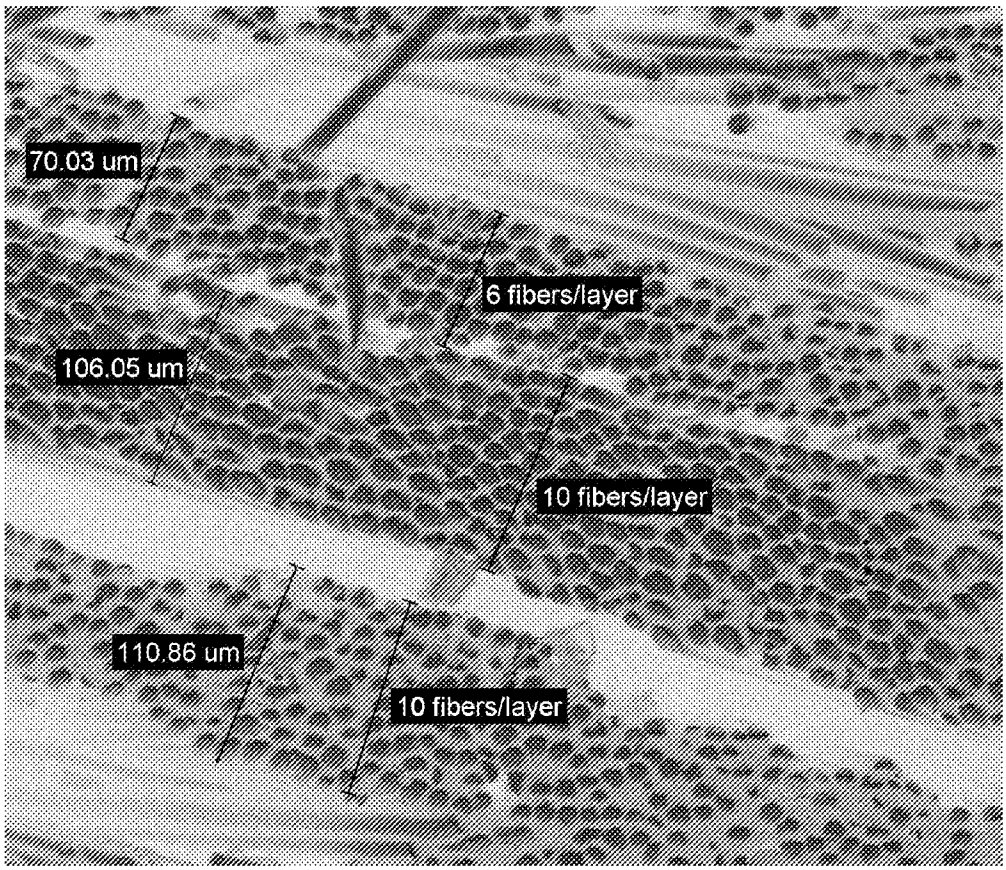

70.03 um
6 fibers/layer
106.05 um
10 fibers/layer
110.86 um
10 fibers/layer

Fig 20

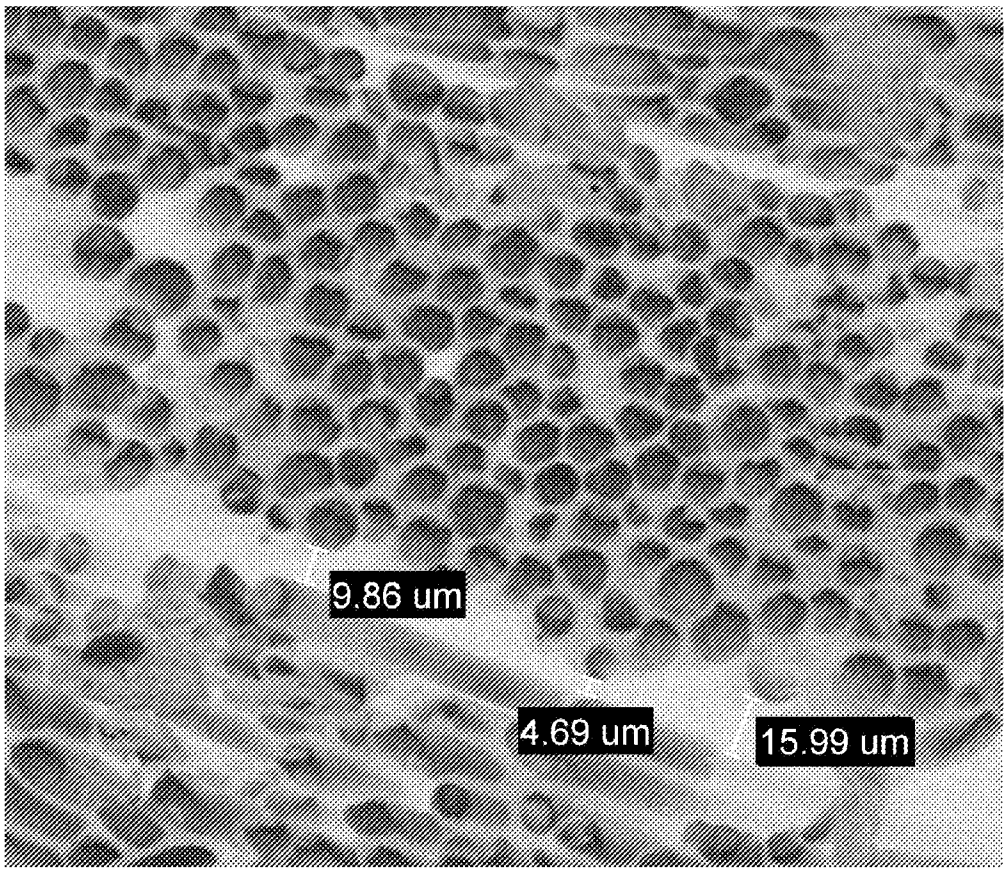


Fig 21

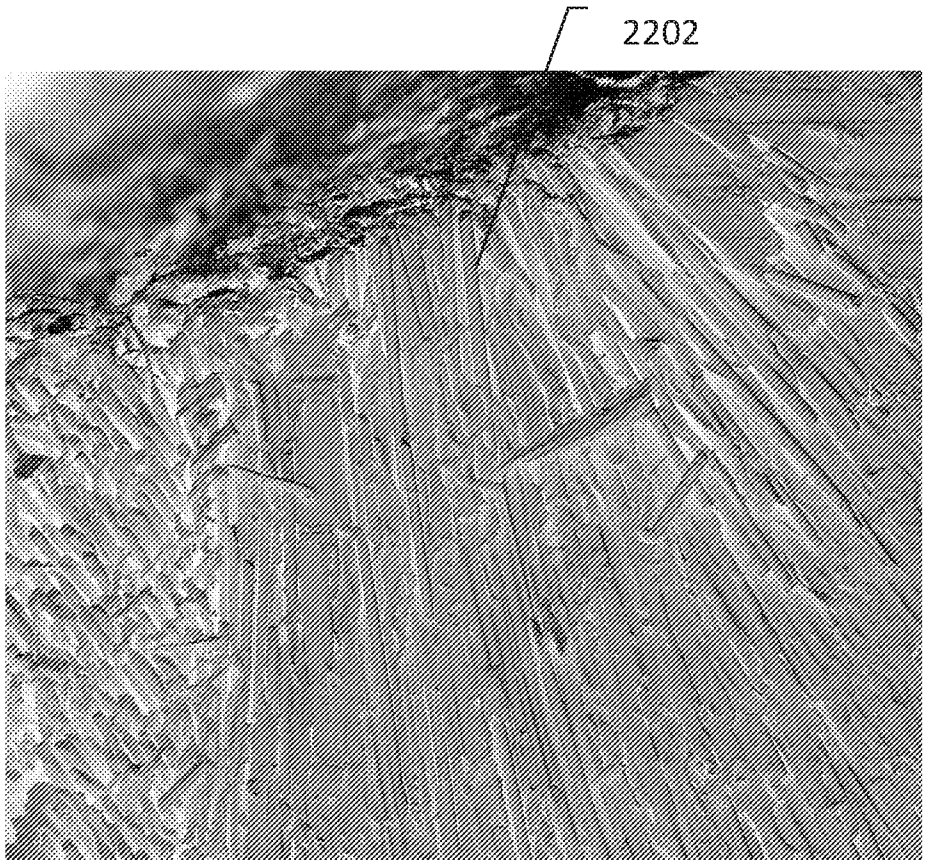


Fig 22

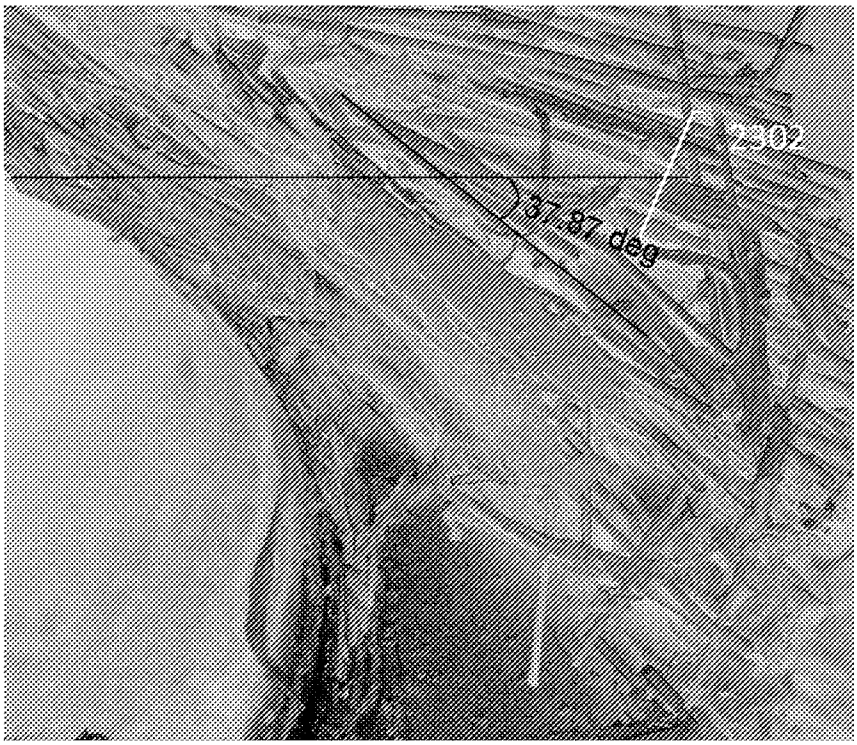


Fig 23

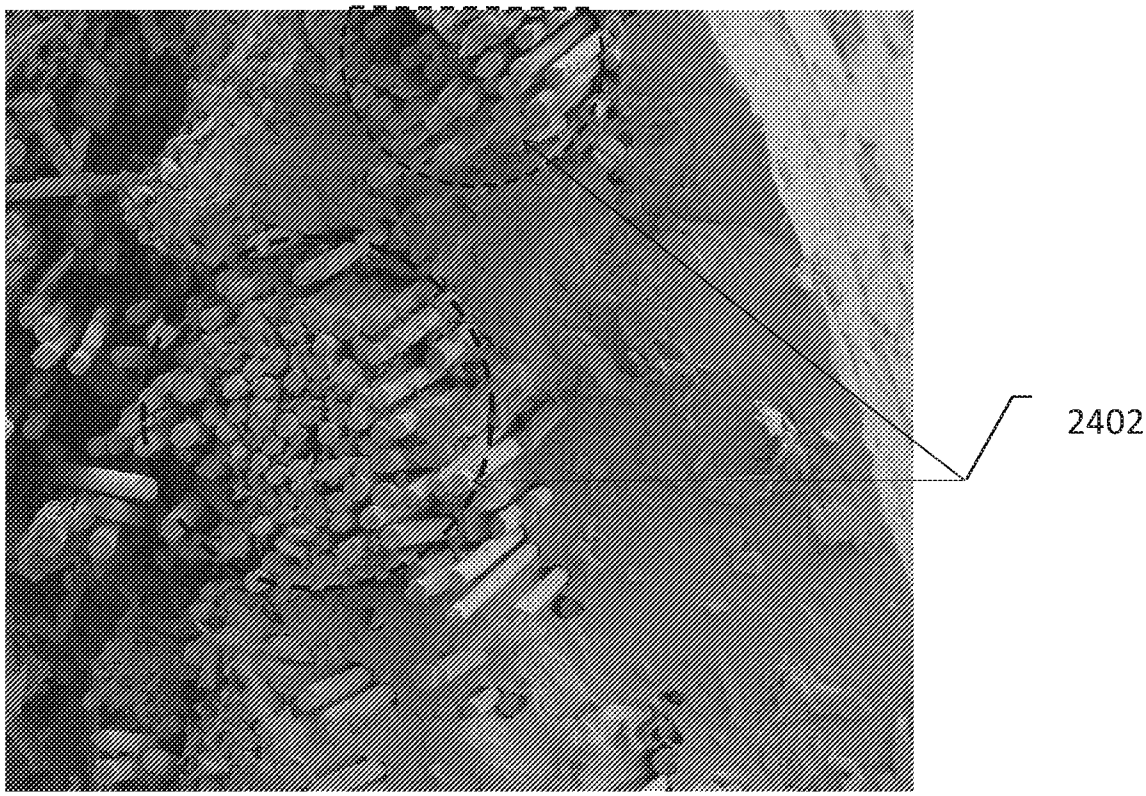


Fig 24

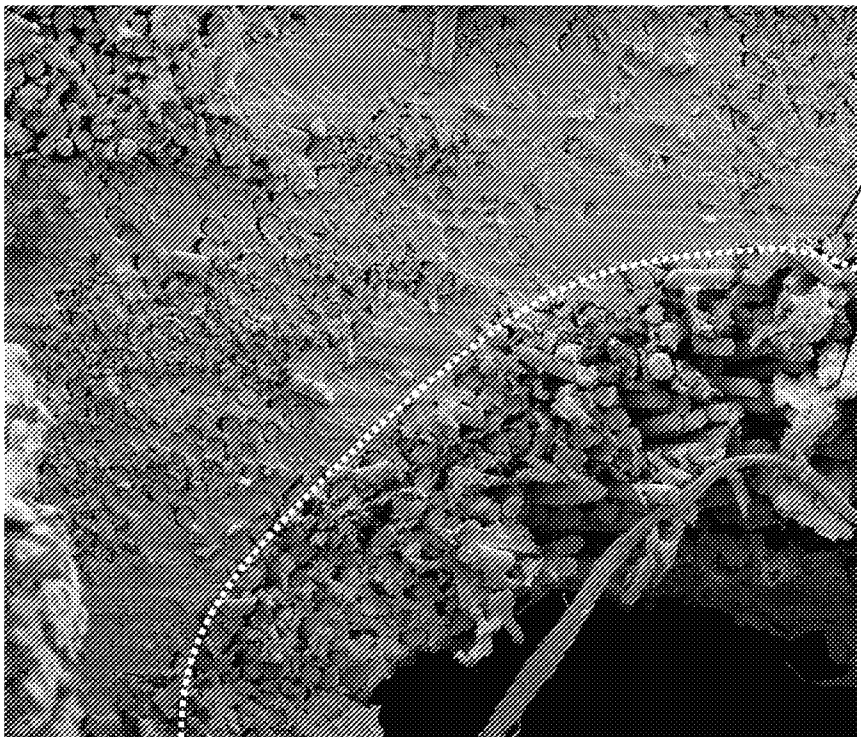


Fig 25

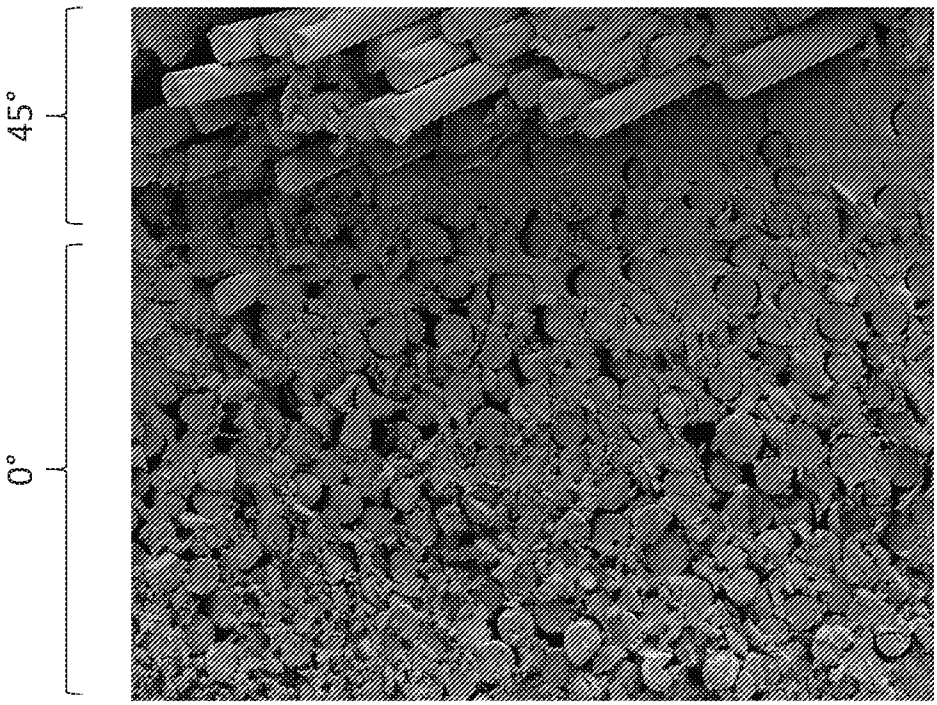
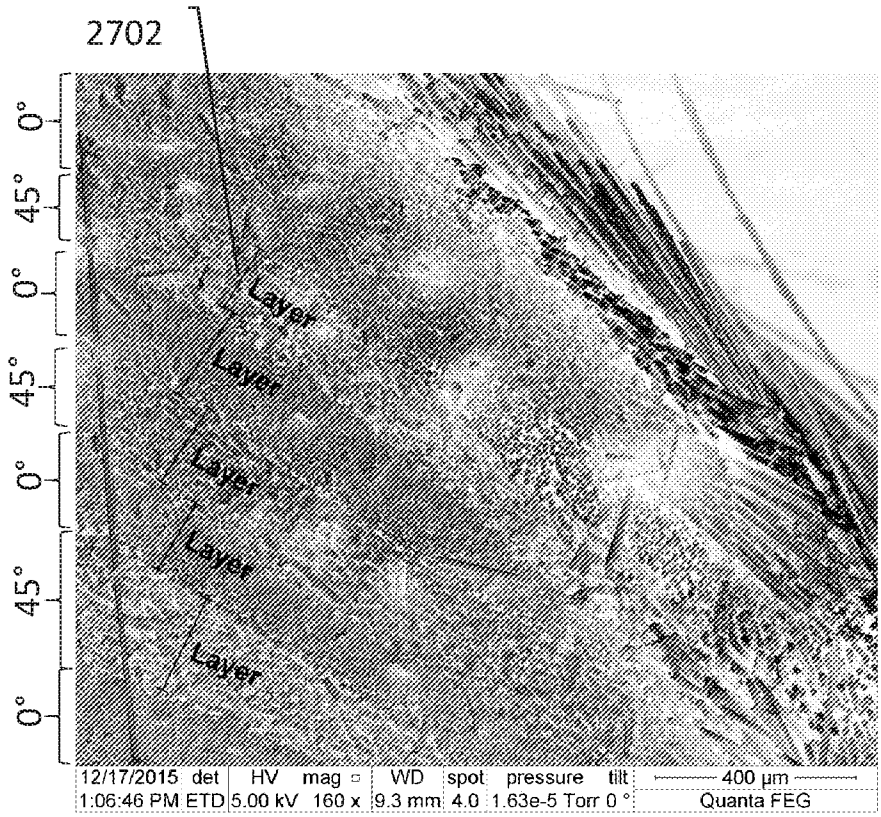


Fig 26


2702
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45°
0°
45°
0°
45°
0°
Layer
Layer
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Layer
12/17/2015 det HV mag WD spot pressure tilt 400 µm
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Fig 27

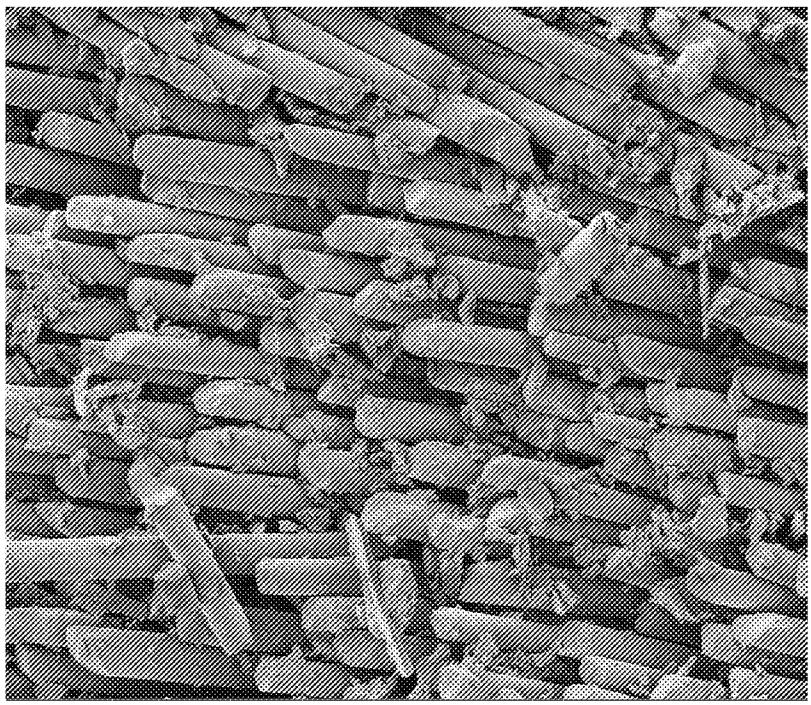

Fig 28

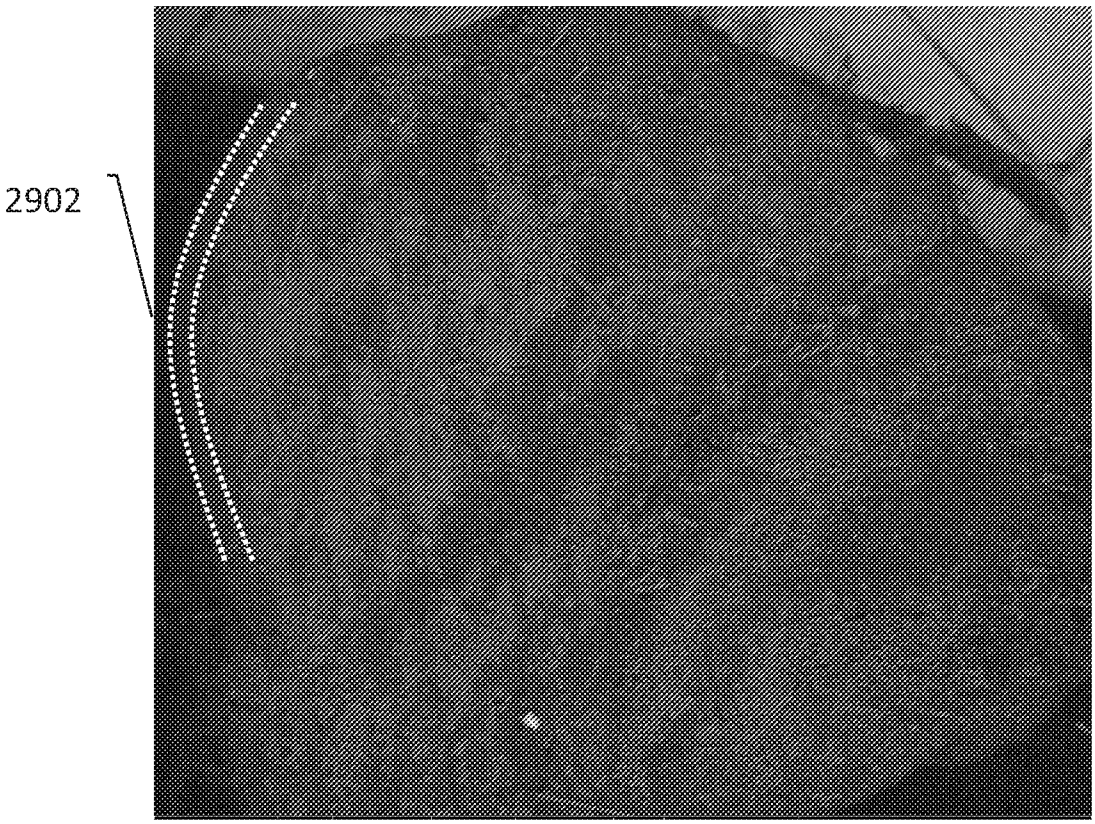

Fig 29

Figure 30. Continuous fiber reinforced composite tape

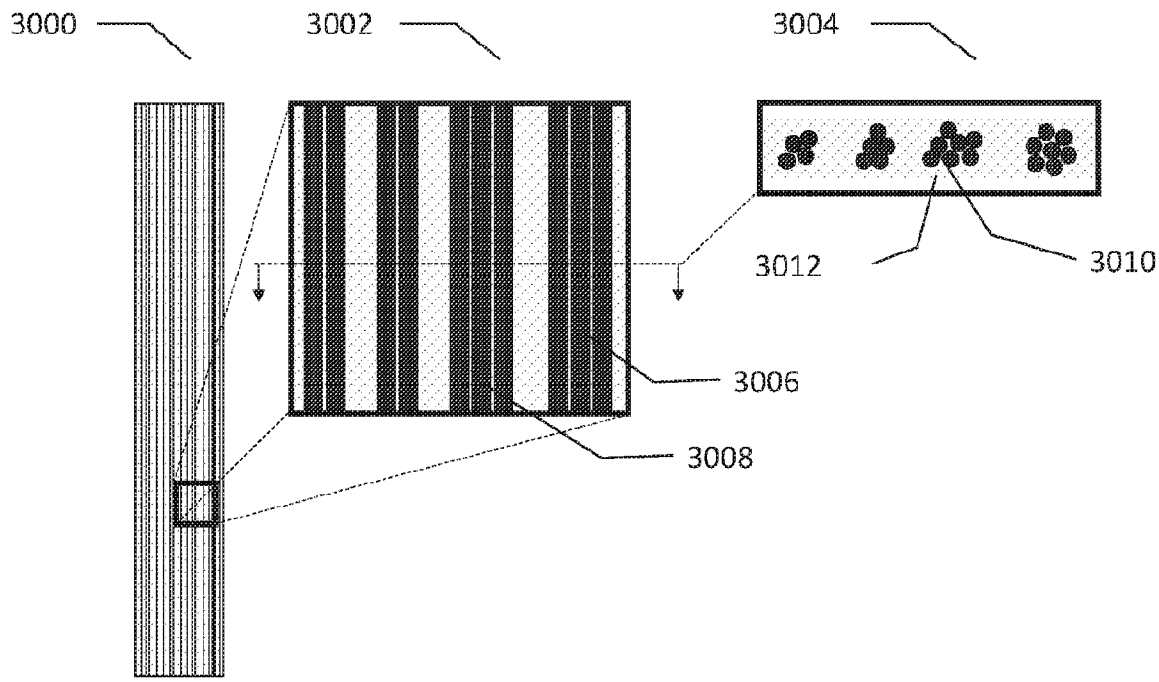

Figure 31. Cut-away, three-dimensional view of continuous fiber reinforced composite tape

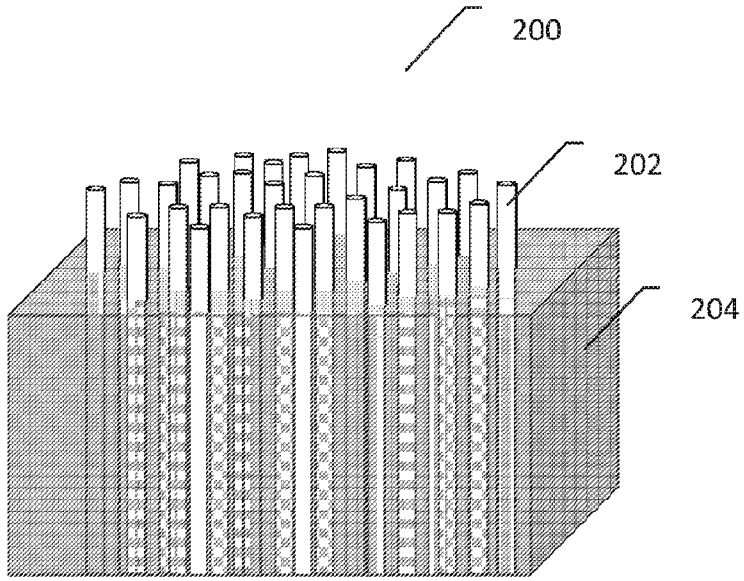

Figure 32a. Continuous-fiber reinforced sheet structure wherein sheet is comprised of multiple layers, each aligned at an angle to each other.

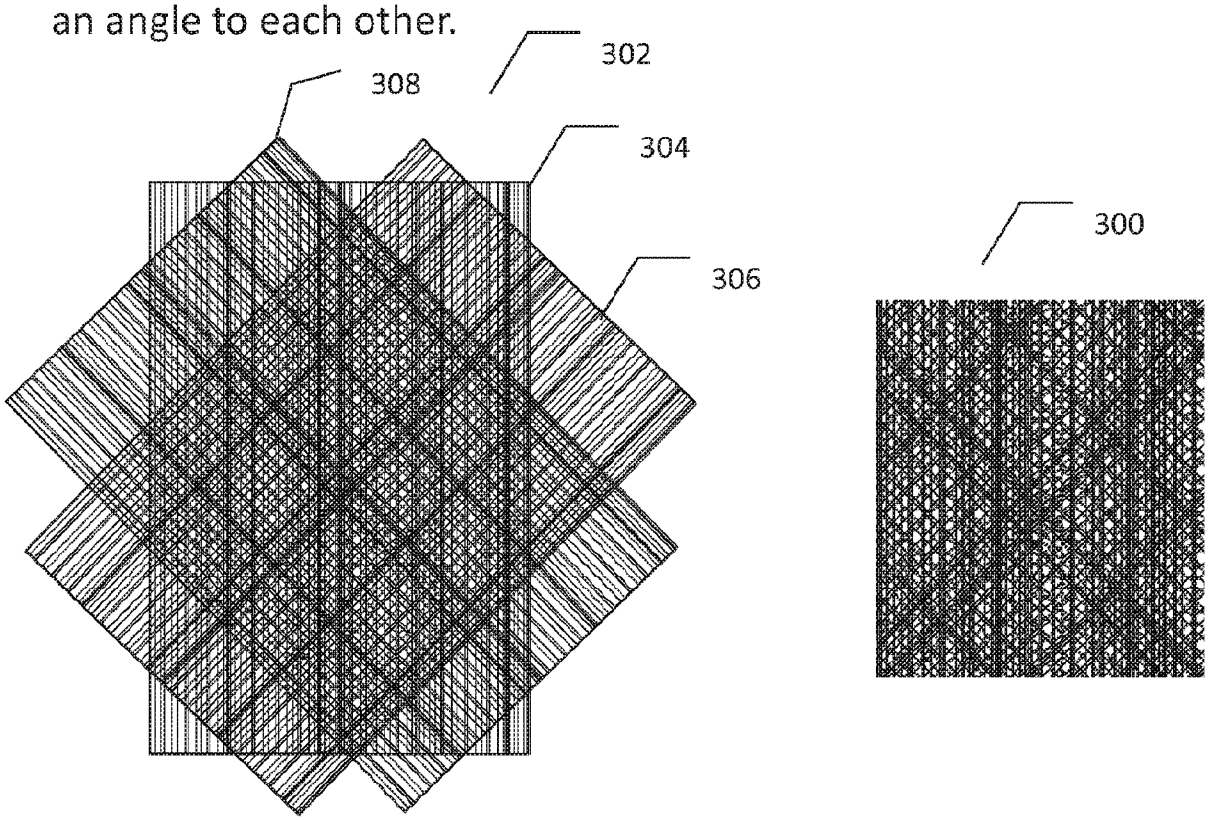

Figure 32b. Cut-away view of continuous-fiber reinforced sheet structure wherein sheet is comprised of multiple layers, each aligned at an angle to each other.

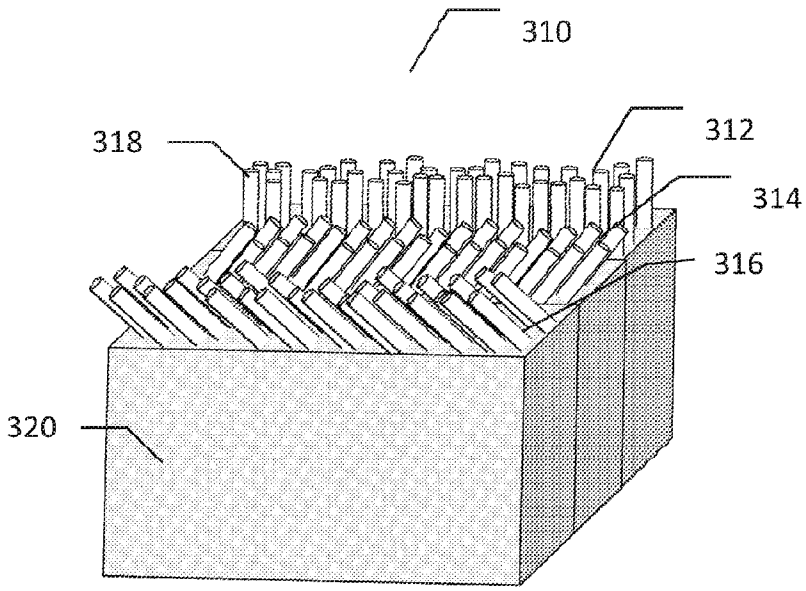

Figure 33. Continuous fiber reinforced medical implant wall with perforations to allow for tissue penetration

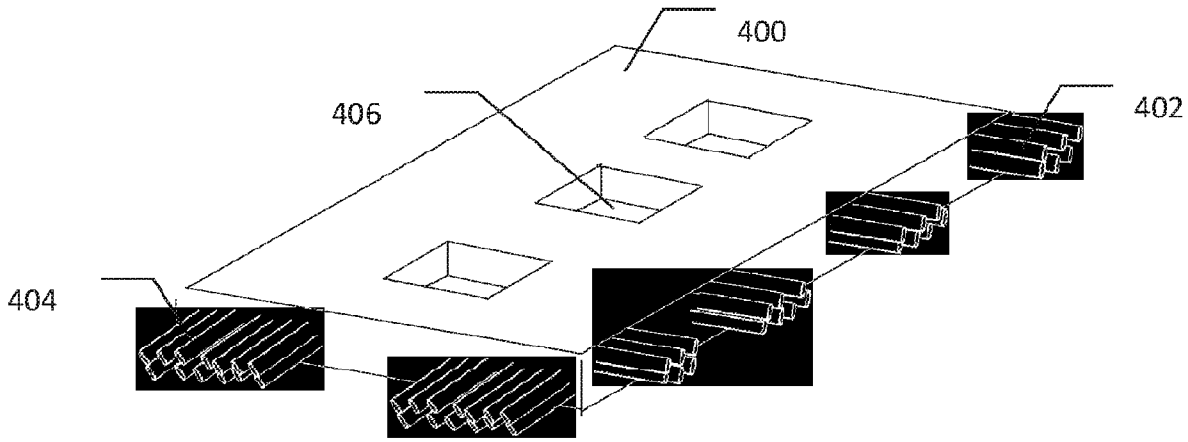

Figure 34. Bone filler cage

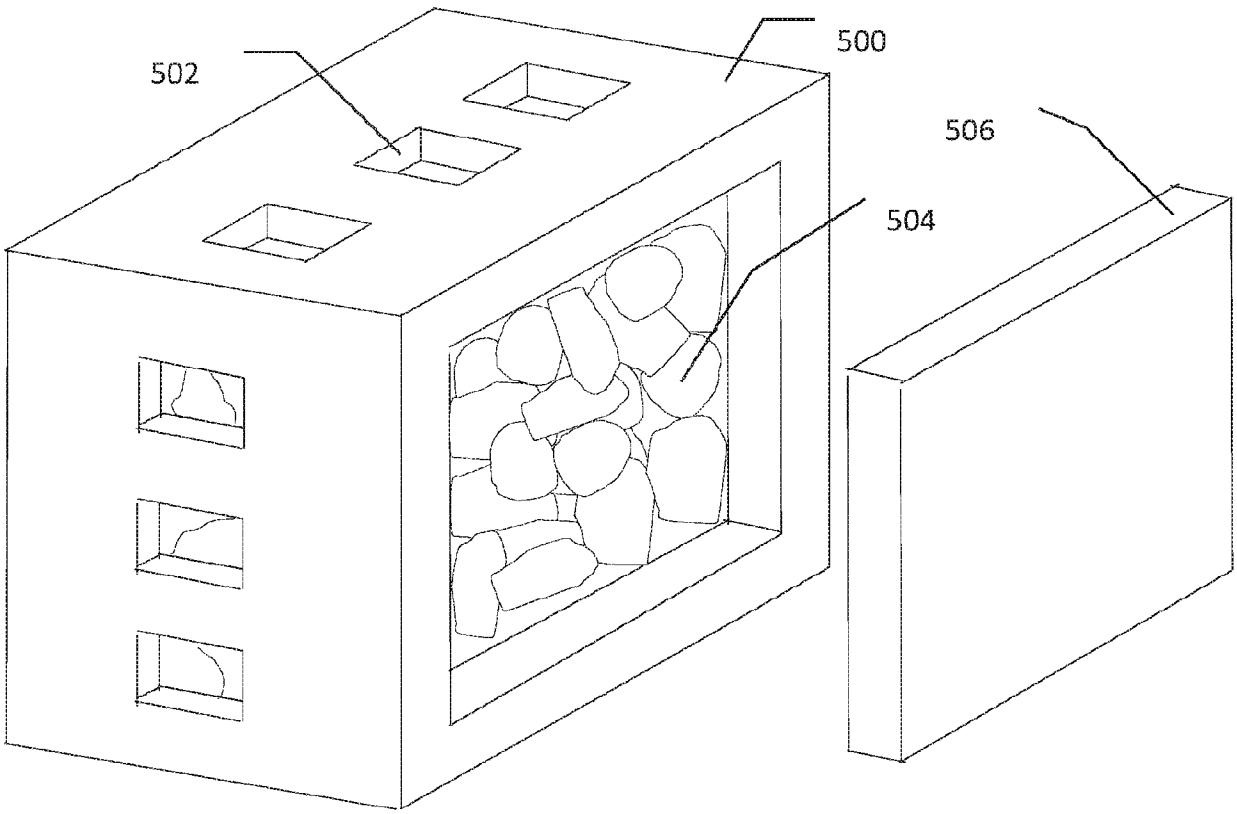

Figure 35. Continuous-fiber reinforced frame / backbone with non-reinforced polymer molded or printed on top