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(54) Title: POLYMERIC, DEGRADABLE DRUG-ELUTING STENTS AND COATINGS

(57) Abstract: Absorbable stents and absorbable stent coatings have been developed with improved properties. These devices preferably comprise biocompatible copolymers or homopolymers of 4-hydroxybutyrate, and optionally poly-L-lactic acid and other absorbable polymers and additives. Compositions of these materials can be used to make absorbable stents that provide advantageous radial strengths, resistance to recoil and creep, can be plastically expanded on a balloon catheter, and can be deployed rapidly *in vivo*. Stent coatings derived from these materials provide biocompatible, uniform coatings that are ductile, and can be expanded without the coating cracking and/or delaminating and can be used as a coating matrix for drug incorporation.



WO 2007/092418 A2

**POLYMERIC, DEGRADABLE DRUG-ELUTING
STENTS AND COATINGS**

CROSS-REFERENCE TO RELATED APPLICATIONS

The present application claims priority to U.S. Serial No. 60/747,144 filed May 12, 2006, U.S. Serial No. 60/765,840, filed February 7, 2006, and U.S. Serial No. 60/765,808 filed February 7, 2006.

Field of the Invention

The present invention generally relates to absorbable polymer compositions that can be used to prepare absorbable stents, and absorbable stent coatings.

Background of the Invention

Stents are currently used in a range of medical applications normally to prevent re-occlusion of a vessel after a procedure to dilate the vessel. Examples include cardiovascular, urology, and gastroenterology stents, with the former being by far the largest market. Generally, stents are made from permanent materials, such as metal alloys or non-absorbable thermoplastics, and can additionally incorporate special coatings and drugs to improve their performance *in vivo*. These coatings, for example, include a number of polymer coating materials for metallic stents, as well as a variety of active agents, such as agents that are anti-inflammatory or immunomodulators, antiproliferative agents, agents which affect migration and extracellular matrix production, agents which affect platelet deposition or formation of thrombus, and agents that promote vascular healing and re-endothelialization. Notably the coatings of currently marketed stents are made from permanent materials.

While the incorporation of certain active agents in coatings on the surfaces of coronary metal stents has been demonstrated to retard restenosis, it has been reported that the polymer coatings that are left after elution of the drug can present a serious risk of late thrombosis (Virmani, R et al., Coron Artery Dis. 2004;15(6):313-8). It has also been reported that the polymeric coating materials of drug-eluting stents may cause hypersensitivity reactions in the patient treated with such coated stents (Nebeker, J.R. et al., J Am Coll

Cardiol 2006;47:175– 81). Thus, there is a need to develop new stent coating materials that can be used to deliver drugs without the risk of late thrombosis and hypersensitivity reactions.

Furthermore, although permanent metal stents are used widely in coronary stenting applications, and their use in peripheral stenting is growing rapidly, there remain several drawbacks to the use of permanent materials to manufacture these stents (Colombo, A et al., Circulation. 2000 25;102(4):371-3, Erne, P. et al., Cardiovasc Intervent Radiol. 2005). First, metal stents are not compatible with certain methods of medical imaging, such as MRI and CT scanning systems. Second, metal stents can cause complications if the patient subsequently needs coronary artery bypass surgery, or other surgical intervention, requiring manipulation of a stented vessel. Third, the use of permanent stents can result in long-term compliance mismatches between the metal stent and the stented vessel, and fourth, in certain peripheral applications, catastrophic failure of metal stent struts has been reported.

It should also be noted that permanent stents used in urology applications to temporarily relieve obstruction in a variety of benign, malignant, and post-traumatic vessel conditions are prone to rapid encrustation (Shaw G.L. et al., Urol Res. 2005 Feb;33(1):17-22). Such encrustation often necessitates removal of the stent. Removal, however, requires an additional procedure, and can be difficult and painful because of tissue in-growth. The use of a degradable implant would eliminate this clinical problem.

To address the disadvantages associated with the use of permanent materials in stents and stent coatings, there have been several reports describing the use of absorbable materials to make stents and stent coatings. U.S. Patent Nos. 5,059,211 and 5,306,286 to Stack *et al.* describe the use of absorbable materials to make stents. Stack, however, does not describe which specific absorbable materials a person skilled in the art would use to make an absorbable stent, or the properties necessary to make such stents.

U.S. Patent No. 5,935,506 to Schmitz *et al.* describes a method to manufacture an absorbable stent from poly-3-hydroxybutyrate (P3HB).

U.S. Patent No. 6,045,568 to Igaki *et al.* describes absorbable stents manufactured from knitting yarns of polylactic acid (PLA), polyglycolic acid (PGA), polyglactin (P(GA-co-LA)), polydioxanone (PDS), polyglyconate (a block co-polymer of glycolic acid and trimethylene carbonate, P(GA-co-TMC)), and a copolymer of glycolic acid or lactic acid with ϵ -caprolactone (P(GA-co-CL) or P(LA-co-CL)).

Laaksovirta *et al.* describe a self-expandable, biodegradable, self-reinforced stent from P(GA-co-LA) for use in urethral applications (*J Urol.* 2003 Aug;170(2 Pt 1):468-71).

The potential use of polyanhydride and polyorthoester polymers to manufacture absorbable stents has also been described by Tanguay, J.F. *et al.* Current Status of Biodegradable Stents, *Cardiology Clinics*, 12:699-713 (1994).

WO 98/51812 to Williams *et al.* discloses methods to remove pyrogens from polyhydroxyalkanoates, and the fabrication of stents with these depyrogenated materials. WO 99/32536 to Martin *et al.* and WO 00/56376 to Williams *et al.* disclose methods to prepare polyhydroxyalkanoates with controlled degradation rates, and the fabrication of stents with these materials.

Van der Giessen *et al.* (Marked Inflammatory Sequelae to Implantation of Biodegradable and Nonbiodegradable Polymers in Porcine Coronary Arteries, *Circulation*, 94:1690-1697 (1996)) evaluated coatings of a copolymer of glycolic acid and lactic acid (P(GA-co-LA)), polycaprolactone (PCL), poly-3-hydroxybutyrate-co-3-hydroxyvalerate (P(3HB-co-3HV)), a polyorthoester, and a polyethyleneoxide-polybutylene terephthalate on metal stents, and reported that the coatings induced marked inflammatory reactions within the coronary artery.

Despite some progress towards the development of absorbable stents and stent coatings, there is currently no coronary or peripheral stent device comprising an absorbable material approved for general sale in the United

States or Europe. This is partly because of the highly demanding requirements of an absorbable material used for medical stenting applications and the shortcomings of the currently available materials. Further improvements to existing materials that are considered desirable, or required, include the following elements: (i) an absorbable stent or stent coating that is biocompatible, does not create a risk of late stage thrombosis, and provides long-term vessel patency; (ii) an absorbable stent that has sufficient radial strength (or hoop strength) to prevent the collapse of the vessel wall or stent; (iii) an absorbable polymer composition that when processed into a stent or stent coating can be expanded *in vivo*, from a suitably low profile form to the desired diameter without surface or strut cracking or similar types of mechanical failure; (iv) an absorbable stent or permanent stent coated with an absorbable polymer that can be dilated sufficiently fast *in vivo* to allow deployment of the stent without risk to the patient, and using a reasonable inflation pressure if the stent is delivered using a balloon catheter; (v) an absorbable stent that does not recoil significantly after deployment; (vi) an absorbable stent that is sufficiently resistant to creep to be effective; (vii) an absorbable stent with strut thicknesses that are relatively low in profile once the stent is implanted, and with edges that are smooth; (viii) an absorbable stent coating that can be applied in a uniform manner, without defects such as web formation between struts, and a method for such application; (ix) an absorbable stent, and/or a stent coated with an absorbable material, where the struts are not susceptible to fracture after implantation, and the risk of vessel perforation is eliminated; (x) an absorbable stent that does not interfere with medical scanning systems, such as MRI and CT; (xi) an absorbable stent, and a stent coated with an absorbable material, that protects against an inflammatory response, limits smooth muscle cell proliferation, and neointimal hyperplasia after implantation, stimulates positive remodeling of the vessel wall, and eliminates long-term compliance mismatches between the stent and the vessel wall; (xii) an absorbable stent, and/or a stent coated with an absorbable material, that is sufficiently flexible to allow delivery to the desired location without strut fracture or kinking, which can conform to

the shape of the affected body lumen; (xiii) an absorbable stent that contains a contrast agent, radiopaque markers, or similar material that allows the stent to be imaged using conventional scanning techniques; (xiv) an absorbable coating that adheres sufficiently strongly to a metal stent, maintains its integrity following stent expansion and does not delaminate; (xv) an absorbable stent and a coated permanent stent that can be loaded with one or more drugs or co-drugs (for example, on the inside or surface of the stent or coating) to improve the performance of the stent by controlled delivery of the drug(s), including agents that are anti-inflammatory or immunomodulators, antiproliferative agents, drugs which affect migration and extracellular matrix production, drugs which affect platelet deposition or formation of thrombus, and drugs that promote vascular healing and re-endothelialization, and that also allow larger drug loadings; (xvi) an absorbable stent and/or a stent coated with an absorbable material that can be mounted onto a catheter, and subsequently delivered *in vivo* without causing damage to the stent; (xvii) an absorbable stent, and an absorbable coating on a stent, that is absorbed *in vivo* over a time period that allows positive remodeling of the vessel wall, does not prematurely fail due to fatigue, and results in long-term vessel patency; (xviii) an absorbable stent that does not shorten in an undesirable manner upon expansion and deployment; (xix) an absorbable stent or permanent stent coated with an absorbable material that can be sterilized without detrimental loss of properties, for example, by irradiation or exposure to ethylene oxide; (xx) an absorbable stent and a coated metal stent that can be loaded with one or more drugs to improve the performance of the stent by controlled delivery of the drug(s), where the method of polymer degradation (e.g. surface erosion or bulk degradation) allows for delivery of large drugs such as proteins; (xxi) an absorbable stent and a coated permanent stent that can be loaded with one or more drugs to improve the performance of the stent by controlled delivery of the drug(s), where the low-acidity polymer degradation products (of the stent or stent coating) allows for delivery of large drugs such as proteins without drug denaturation; (xxii) an absorbable material for use in stents that has a glass transition

temperature below body temperature, a melt temperature above 50°C, and a shelf-life of at least one to three years.

Any discussion of documents, acts, materials, devices, articles or the like which has been included in the present specification is solely for the purpose of providing a context for the present invention. It is not to be taken as an admission that any or all of these matters form part of the prior art base or were common general knowledge in the field relevant to the present invention as it existed before the priority date of each claim of this application.

It is therefore an object of a preferred embodiment of this invention to provide absorbable compositions that can be used to develop improved absorbable stents, and absorbable stent coatings.

It is another object of a preferred embodiment of this invention to provide improved absorbable stents, and stents coated with absorbable materials.

It is a further object of a preferred embodiment of this invention to provide methods for preparing improved absorbable stents and stents coated with absorbable materials.

It is a yet still further object of a preferred embodiment of this invention to provide methods for the delivery of the absorbable stents and stents coated with absorbable materials.

Summary of the Invention

Throughout this specification the word "comprise", or variations such as "comprises" or "comprising", will be understood to imply the inclusion of a stated element, integer or step, or group of elements, integers or steps, but not the exclusion of any other element, integer or step, or group of elements, integers or steps.

In a first aspect, the invention provides an absorbable biocompatible stent comprising a polymer composition, wherein the composition comprises

(a) one or more polymers selected from the group consisting of poly-4-hydroxybutyrate, copolymers of 4-hydroxybutyrate, and mixtures thereof and

(b) one or more polymers selected from the group consisting of polylactides, copolymers of lactic acid, and mixtures thereof,

wherein the composition comprises between 60 and 98% by weight of a polylactide, wherein the stent is plastically expandable at normal body temperature, and is of a first diameter sufficient to be retained upon a balloon catheter for placement within a body lumen, and is expandable to a second diameter sufficient to be retained within the body lumen.

In another aspect, the invention provides a method of making the stent of the first aspect, the method comprising fabricating a tube or stent blank using a technique selected from the group consisting of solution based dip-coating, casting, and combinations thereof or a technique selected from the group consisting of extrusion and
5 injection molding.

In yet another aspect, the invention provides a method of deploying the stent of the first aspect, comprising mounting the stent on a delivery system and balloon-expanding the stent inside the body lumen with a balloon pressure of 4 to 16 bar.

In yet another aspect, the invention provides a metal stent comprising an
10 absorbable polymeric coating, wherein the polymeric coating comprises (a) one or more polymers selected from the group consisting of poly-4-hydroxybutyrate, copolymers of 4-hydroxybutyrate, and mixtures thereof and (b) one or more polymers selected from the group consisting of polylactides copolymers of lactic acid, and mixtures thereof, wherein the composition comprises between 60 and 98% by weight of
15 a polylactide, wherein the stent coating is plastically expandable at normal body temperature without cracking or delaminating and is of a first diameter sufficient to be retained upon a balloon catheter for placement within a body lumen, and is expandable to a second diameter sufficient to be retained within the body lumen.

Absorbable compositions and stents, and absorbable coatings for stents, with
20 improved properties and performance, and methods for making these materials and devices, have been developed. These compositions and devices are preferably derived from a biocompatible homopolymer and/or copolymer(s) of 4-hydroxybutyrate, and combinations of these materials with other absorbable materials. Absorbable stents are most preferably derived from compositions comprising a homopolymer or
25 copolymer(s) of 4-hydroxybutyrate with a polymer of lactic acid, with or without plasticizers.

Different methods may be used to apply the absorbable stent coatings. Most preferably, the coatings are applied from solution by spraying. Different methods may be used to prepare the absorbable stents. A preferred method comprises forming a tube
30 by solution dipping or extrusion, injection molding or micro injection molding, and cutting the tube with a laser to form the stent. The stent may be used as manufactured or expanded *in vivo*, for example, using an expandable balloon catheter.

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The absorbable stent coatings provide devices with thin coatings on the stent struts, without the formation of web-like structures between the struts, and stents that can be expanded quickly without the coating cracking,

delaminating, or losing its structural integrity. The absorbable stent coatings are biocompatible, degrade to less acidic metabolites by mechanisms that include surface erosion (minimizing the risk of particulate breaking away from the stent surface), elongate up to 1,000% of their original length, adhere to the stent, can be mounted to a catheter and deployed without damage to the coating, degrade over a period of up to about one year, can be sterilized by irradiation or treatment with ethylene oxide, and can be loaded or coated with drugs for controlled release. The stents are flexible, and more compliant with the vessel wall; have sufficient radial strength and strength retention to permit positive remodeling for long-term patency; and have a radial recoil of less than 10%, and more preferably less than 6%; can be expanded rapidly *in vivo*, without cracking or other mechanical failure, preferably in less than five minutes, and more preferably in less than one minute, using a balloon pressure of 4 to 16 bar, more preferably 8 bar, and can be delivered to the desired location without strut fracture, kinking, or damage to the vessel wall; do not exhibit any significant creep at 100 mmHg for 7 days, do not shorten significantly upon expansion; can be constructed with smooth strut edges with strut thicknesses of less than 300 μm , more preferably 160 μm or less for coronary applications, and 250-270 μm for peripheral applications; can contain contrast agents, radiopaque markers or similar material to allow the imaging of the stent *in vivo*, and can also be loaded and/or coated with therapeutic, prophylactic or diagnostic agents, including, but not limited to, agents that are anti-inflammatory or immunomodulators, antiproliferative agents, drugs which affect migration and extracellular matrix production, drugs which affect platelet deposition or formation of thrombus, and drugs that promote vascular healing and re-endothelialization, at low or high drug loadings; can be sterilized, for example, by gamma-irradiation, electron-beam irradiation or ethylene oxide. In the specific case of coronary applications, the absorbable compositions can be used to prepare absorbable stents that can be expanded *in vivo* from an inner diameter of approximately 1-1.4 mm to 3-4 mm in about one minute. Larger absorbable stents can also be made for use, for example, in

peripheral and urology applications. A preferred internal diameter for peripheral applications is 2.0 to 2.8 mm with a wall thickness of 250-270 μm .

Brief Description of the Drawings

Figure 1 is the chemical structure of poly-4-hydroxybutyrate (P4HB, TephafLEX® biomaterial).

Figure 2 shows some of the known biosynthetic pathways for the production of P4HB. Pathway enzymes are: 1. Succinic semialdehyde dehydrogenase; 2. 4-hydroxybutyrate dehydrogenase; 3. diol oxidoreductase; 4. aldehyde dehydrogenase; 5. Coenzyme A transferase; and 6. PHA synthetase.

Figure 3 is a graph showing the accelerated decrease of molecular weight (Mw) of a polymer blend material of high molecular weight PLLA and P4HB with a mass ratio of 78:22 % in comparison to pure PLLA as a function of *in vitro* incubation time in Sørensen buffer (pH = 7.4) at 37 °C.

Figure 4 is a graph showing the *in vitro* drug release profiles of metallic stents coated with the following different compositions of a P4HB matrix incorporated with the immunomodulator rapamycin as a base coat, and coated with a top coat of pure P4HB as a diffusion barrier for retarded drug release: stent 85:15-1 – polymer/drug ratio 85:15 (w/w), base coat thickness = 20 μm , no top coat; stent 85:15-2 – polymer/drug ratio 85:15 (w/w), base coat thickness = 20 μm , top coat thickness = 5 μm ; stent 85:15-3 – polymer/drug ratio 85:15 (w/w), base coat thickness = 20 μm , top coat thickness = 15 μm ; stent 85:15-4 – polymer/drug ratio 85:15 (w/w), base coat thickness = 10 μm , top coat thickness = 10 μm .

Figure 5 is a graph showing the *in vitro* drug release profiles of metallic stents coated with the following different compositions of a P4HB matrix incorporated with the immunomodulator rapamycin as a base coat, and coated with a top coat of pure P4HB as a diffusion barrier for retarded drug release: stent 40:60-1 – polymer/drug ratio 40:60 (w/w), base coat thickness = 5 μm , top coat thickness = 10 μm ; stent 70:30-1 – polymer/drug ratio 70:30 (w/w), base coat thickness = 5 μm , top coat thickness = 10 μm ;

stent 85:15-4 – polymer/drug ratio 85:15 (w/w), base coat thickness = 10 μm , top coat thickness = 10 μm .

Detailed Description of the Invention

Absorbable stents and stents coated with absorbable materials have been developed that have improved properties.

I. Definitions

“Poly-4-hydroxybutyrate” as generally used herein means a homopolymer comprising 4-hydroxybutyrate units. It may be referred to herein as P4HB or TephafLEX® biomaterial (manufactured by Tephaf, Inc., Cambridge, MA).

“Copolymers of poly-4-hydroxybutyrate” as generally used herein means any polymer comprising 4-hydroxybutyrate with one or more different hydroxy acid units.

“Copolymers of lactic acid” as generally used herein means any polymer comprising lactic acid with one or more different hydroxy acid units.

“Molecular weight” as used herein, unless otherwise specified, refers to the weight average molecular weight (Mw) as opposed to the number average molecular weight (Mn).

“Blend” as generally used herein means a macroscopically homogeneous mixture of two or more different species of polymer.

“Absorbable” or “degradable” as generally used herein means the material is broken down in the body and eventually eliminated from the body.

“Biocompatible” as generally used herein means the biological response to the material or device is appropriate for the device’s intended application *in vivo*. Any metabolites of these materials should also be biocompatible.

II. Compositions

A. Absorbable polymers

The stents and stent coatings may be formed from absorbable polymers, such as poly-4-hydroxybutyrate (P4HB), and copolymers thereof,

such as poly-4-hydroxybutyrate-co-3-hydroxybutyrate (P(4HB-co-3HB)). In a preferred embodiment, the absorbable stents are formed from combinations of P4HB and/or copolymers thereof, with a second absorbable material. A preferred second absorbable material is a polyhydroxy acid, preferably polylactic acid (PLA), and even more preferably poly-L-lactic acid (PLLA) (such as Resomer® L214 available from Boehringer Ingelheim). Copolymers of lactic acid may also be used as the second absorbable material, including copolymers with glycolic acid.

Tepha, Inc. of Cambridge, MA produces poly-4-hydroxybutyrate (P4HB) and copolymers thereof using transgenic fermentation methods. Poly-4-hydroxybutyrate is a strong, pliable thermoplastic polyester that is produced by a fermentation process (see U.S. Patent No. 6,548,569 to Williams *et al.*). Despite its biosynthetic route, the structure of the polyester is relatively simple (Figure 1). The polymer belongs to a larger class of materials called polyhydroxy-alkanoates (PHAs) that are produced by numerous microorganisms (Steinbüchel, A. Polyhydroxyalkanoic acids, *Biomaterials*, 123-213 (1991); Steinbüchel A., et al. Diversity of Bacterial Polyhydroxyalkanoic Acids, *FEMS Microbial. Lett.* 128:219-228 (1995); and Doi, Y. *Microbial Polyesters* (1990)). In nature these polyesters are produced as storage granules inside cells, and serve to regulate energy metabolism. They are also of commercial interest because of their thermoplastic properties, and relative ease of production. Several biosynthetic routes are currently known for producing P4HB, as shown in Figure 2. Chemical synthesis of P4HB has been attempted, but it has not been possible to produce the polymer with a sufficiently high molecular weight necessary for most applications (Hori, Y., et al., *Polymer* 36:4703-4705 (1995)).

Tepha, Inc. (Cambridge, MA) produces P4HB and related copolymers for medical use, and has filed a Device Master File with the United States Food and Drug Administration (FDA) for P4HB. Related copolymers include 4-hydroxybutyrate copolymerized with 3-hydroxybutyrate or glycolic acid

(U.S. patent application publication number US 2003/0211131 by Martin & Skraly, U.S. Patent No. 6,316,262 to Huisman *et al.*, and U.S. Patent No. 6,323,010 to Skraly *et al.*). Tephra has also filed a Device Master File with the United States FDA for copolymers containing 3-hydroxybutyrate and 4-hydroxybutyrate. Methods to control molecular weight of PHA polymers are disclosed by U.S. Patent No. 5,811,272 to Snell *et al.*, and methods to purify PHA polymers for medical use are disclosed by U.S. Patent No. 6,245,537 to Williams *et al.* PHAs with degradation rates *in vivo* of less than one year are disclosed by U.S. Patent No. 6,548,569 to Williams *et al.* and WO 99/32536 to Martin *et al.* Other applications of PHAs are reviewed in Williams, S.F., *et al.*, *Polyesters, III*, 4:91-127 (2002), and other applications specific to P4HB are reviewed in Martin *et al.* Medical Applications of Poly-4-hydroxybutyrate: A Strong Flexible Absorbable Biomaterial, *Biochem. Eng. J.* 16:97-105 (2003).

B. Other Stent Components

i. Plasticizers

The absorbable material composition used to produce stents and stent coatings may comprise other materials in addition to the polymers described above. In a preferred method of the invention, a plasticizer may be introduced into the absorbable material prior to forming the stent or stenting coating. Preferred plasticizers are biocompatible. A particularly preferred plasticizer is triethylcitrate (TEC).

ii. Therapeutic, Prophylactic, and Diagnostic Agents

In addition to incorporating plasticizers into the absorbable material, it can be advantageous to incorporate one or more therapeutic, prophylactic or diagnostic agents ("agent") into the stent, either by loading the agent(s) into the absorbable material prior to processing, and/or coating the surface of the stent with the agent(s). The rate of release of agent may be controlled by a number of methods including varying the following: the ratio of the absorbable material to the agent, the molecular weight of the absorbable material, the composition of the agent, the composition of the absorbable polymer, the coating thickness, the number of coating layers and their

relative thicknesses, and/or the agent concentration. Top coats of polymers and other materials, including absorbable polymers, may also be applied to active agent coatings to control the rate of release. For example, P4HB (TephaFLEX® biomaterial from Tepha, Inc.) may be applied as a top coat on a metallic stent coated with P4HB comprising an active agent, such as rapamycin, to retard the release of rapamycin.

Exemplary therapeutic agents include, but are not limited to, agents that are anti-inflammatory or immunomodulators, antiproliferative agents, agents which affect migration and extracellular matrix production, agents which affect platelet deposition or formation of thrombus, and agents that promote vascular healing and re-endothelialization, described in Tanguay *et al.* Current Status of Biodegradable Stents, *Cardiology Clinics*, 12:699-713 (1994), J. E. Sousa, P. W. Serruys and M. A. Costa, *Circulation* 107 (2003) 2274 (Part I), 2283 (Part II), K. J. Salu, J. M. Bosmans, H. Bult and C. J. Vrints, *Acta Cardiol* 59 (2004) 51. Examples of antithrombin agents include, but are not limited to, Heparin (including low molecular heparin), R-Hirudin, Hirulog, Argatroban, Efegatran, Tick anticoagulant peptide, and Ppack.

Examples of antiproliferative agents include, but are not limited to, Paclitaxel (Taxol), QP-2, Vincristin, Methotrexat, Angiopeptin, Mitomycin, BCP 678, Antisense c-myc, ABT 578, Actinomycin-D, RestenASE, 1-Chlor-deoxyadenosin, PCNA Ribozym, and Celecoxib.

Examples of anti-restenosis agents include, but are not limited to, immunomodulators such as Sirolimus (Rapamycin), Tacrolimus, Biorest, Mizoribin, Cyclosporin, Interferon γ 1b, Leflunomid, Tranilast, Corticosteroide, Mycophenolic acid and Biphosphonate.

Examples of anti-migratory agents and extracellular matrix modulators include, but are not limited to, Halofuginone, Propyl-hydroxylase-Inhibitors, C-Proteinase-Inhibitors, MMP-Inhibitors, Batimastat, Probucol.

Examples of antiplatelet agents include, but are not limited to, heparin. Examples of wound healing agents and endothelialization

promoters include vascular epithelial growth factor ("VEGF"), 17β -Estradiol, Tkase-Inhibitors, BCP 671, Statins, nitric oxide ("NO")-Donors, and endothelial progenitor cell ("EPC")-antibodies.

Besides coronary applications, drugs and active agents may be incorporated into the stent or stent coating for other indications. For example, in urological applications, antibiotic agents may be incorporated into the stent or stent coating for the prevention of infection. In gastroenterological and urological applications, active agents may be incorporated into the stent or stent coating for the local treatment of carcinoma.

It may also be advantageous to incorporate in or on the stent a contrast agent, radiopaque markers, or other additives to allow the stent to be imaged *in vivo* for tracking, positioning, and other purposes. Such additives could be added to the absorbable composition used to make the stent or stent coating, or absorbed into, melted onto, or sprayed onto the surface of part or all of the stent. Preferred additives for this purpose include silver, iodine and iodine labeled compounds, barium sulfate, gadolinium oxide, bismuth derivatives, zirconium dioxide, cadmium, tungsten, gold, tantalum, bismuth, platinum, iridium, and rhodium. These additives may be, but are not limited to, micro- or nano-sized particles or nano particles. Radio-opacity may be determined by fluoroscopy or by x-ray analysis.

III. Methods of Making Absorbable Stents

The stents described herein can be fabricated from solution processes, such as dip coating and casting, melt processes such as extrusion and injection molding, and combinations thereof.

In a preferred method, an absorbable stent may be prepared as follows. A polymer, such as P4HB or copolymer thereof, is optionally mixed in a predetermined ratio with a second absorbable polymer, such as PLLA, and if desired, a plasticizer, such as triethylcitrate (TEC), and/or other additives, in a suitable solvent, to prepare a viscous solution of a predetermined concentration. A rod or mandrel of predetermined diameter is then repeatedly dipped into the viscous solution and removed so as to build

up layers of the absorbable material composition on the rod, by precipitation of the material as the solvent is evaporated, with the layer previously deposited only being partially dissolved. Successive dipping of the rod is repeated until the desired thickness of material is built up on the rod, whereupon the rod is withdrawn to yield a circular tube, known as the stent blank, that may be trimmed or coated further as desired.

In an alternative method to prepare the stent blank, a tube of predefined dimensions may be melt extruded from a blend of P4HB or copolymers thereof, optionally with a second absorbable polymer, such as PLLA, and if desired a plasticizer, and /or other additives.

In a further alternative method to prepare the stent blank, a tube of predefined dimensions may be injection molded or micro system injection molded from a blend or composition of P4HB or copolymer thereof with the second absorbable polymer, such as PLLA, and if desired a plasticizer, and /or other additives.

In a preferred method, the dimensions of the stent blank for coronary application are an external diameter of approximately 1.3 mm, and a wall thickness of approximately 150 μm .

The stent blank may then be cut to form the stent. In a preferred method, the stent is cut with a laser according to a predefined stent design. Examples of suitable stent designs are described by Grabow et al. (J Biomech Eng. 2005 Feb;127(1):25-31) and Sternberg et al. (Urologe A. 2004 Oct;43(10):1200-7). In a preferred embodiment, a CO₂ laser, Excimer laser, or a femtosecond laser is used to cut the stent blank.

Another alternative method to prepare the stent blank, is to prepare a fiber by injection molding or extrusion of a composition of P4HB or copolymer thereof, optionally with a second absorbable polymer, such as PLLA, and if desired, a plasticizer and /or other additives. The fiber may be reinforced by solid-state drawing. The stent can then be fabricated from a single fiber or multiple fibers, which may be wound, knitted, braided, woven or welded to a tubular structure or to form a tubular structure.

Additives may, if desired, be added to the stent or stent blank at different steps of the fabrication process. Such additives can include radiopaque materials and/or active agents.

Also, stent coatings may be added to the stent after stent fabrication. Such coatings can include radiopaque materials and/or active agents.

Absorbable stents prepared according to these methods are characterized by the following properties: biocompatibility; potentially reduced risk of late stage thrombosis and restenosis; low profile; rapid deployment *in vivo*; maintenance of structural integrity after expansion; radial strength and strength retention; limited recoil after deployment; resistance to creep; elimination of struts that could potentially fracture over the long-term; radio-opaque, if desired; good compliance match between stent and vessel wall; flexibility and low profile to permit delivery of the stent through small vessels and along restricted and tortuous paths; ability to load drug of choice; ability to positively remodel the vessel wall for long-term patency; compatibility with imaging systems, such as CT and MRI; maintenance of length upon expansion; compatibility with several sterilization options, including gamma-irradiation, electron beam irradiation, and treatment with ethylene oxide; degradation that can include, but is not limited to, surface erosion in addition to bulk degradation; lower acidity degradation products and, ability to dilate the stent *in vivo* sufficiently quickly to allow the deployment of the stent without risk to the patient, and using only a reasonable amount of pressure.

It is notable in particular that absorbable stents and stent coatings can be plastically deformed at normal body temperature, and in reasonable operative times (for example, less than 5 minutes and more preferably less than 1 minute). They do not require the use of thermo-mechanical expansion or stent designs that rely on the use of non-plastic deformation of the stent. For example, US Patent No. 5,670,161 to Healy *et al.* describes biodegradable stents made from copolymers of L-lactide and caprolactone that are not plastically expandable at normal body temperature, but can be expanded by using thermo-mechanical expansion. Attempting to expand

these stents (and other stents made from compositions of degradable materials that are not plastically expandable) in one minute or less causes the stents to fracture. Although not wishing to be bound by theory, this may be due to the brittle or glassy characteristics of the stent composition. Tamai *et al.* Circulation, 2000; (102) 399-404 also describes the need to heat a PLLA stent (the Igaki-Tamai stent) to 50°C in order to expand the stent in 13 seconds. At normal body temperature, expansion is reported to take 20 minutes. Zeltinger *et al.* Biomaterials Forum, 2004 First Quarter, 8, 9 and 24, reported that the expansion of absorbable stents, prepared for example from polylactides (poly-L-lactic acid), with the use of a heated balloon, represented additional risks to the patient, and reported that these stents have therefore not been commercialized. In one approach to overcome the inability to plastically expand stiff, rigid absorbable polymers, this group employed a new stent design, based upon a slide and lock ratchet mechanism. Thus, prior approaches to develop absorbable stents have sought ways using heat to expand rigid, stiff absorbable polymers or polymer compositions, or to eliminate the need for plastic deformation of the polymer composition by using stent designs, such as slide and ratchet or self-expanding designs that do not require compositions that are plastically expandable. In contrast, the stents made from the compositions described herein can be expanded without the use of heat.

The specific compositions described herein are highly advantageous in permitting the deployment of the stent *in vivo* for coronary applications in approximately one minute, using a reasonable inflation pressure and at normal body temperature, yet still providing an absorbable stent with high radial strength and strength retention, acceptable recoil and creep, flexibility to contour to the vessel wall, ability to remodel the vessel wall and degrade over time, and all based upon a low profile design. The specific compositions described herein can be designed to degrade more rapidly than stents made from polymers or copolymers comprising lactic acid. For example, Ormiston *et al.* (Catheter Cardiovasc. Interv. 2006; (69) 128-131) has reported that absorbable stents made from poly-L-lactic acid (PLLA) degrade very slowly

over a period of 2-3 years. In contrast, a specific composition described herein of PLLA comprising 22% P4HB (TephaFLEX® biomaterial from Tepha, Inc.) can be fabricated into an absorbable stent that degrades much faster. At 48 weeks, less than 20% of the original molecular weight of this blend remains compared to almost 50% for a PLLA-derived stent. The rate of degradation may be further adjusted by manipulation of the percentage of P4HB in the P4HB/PLLA blend.

IV. Method of Coating a Stent with An Absorbable Polymer Composition

In a preferred method, a stent may be coated with an absorbable polymer as follows. A polymer, such as P4HB or copolymer thereof, which may optionally incorporate a second absorbable polymer and/or additives, is dissolved at a known concentration in a volatile solvent. The solution is then sprayed onto the stent to be coated in a uniform manner to provide an even surface coating of the stent. Evaporation of the solvent produces a film coating on the surface of the stent. The process may be repeated to build up the thickness of the coating. The concentration of the solution, application time, drying time, position and rotation of the stent, and number of applications, may be adjusted to create the desired coating thickness, and also to yield a coated stent where the coating only evenly coats the struts, and does not form web structures between struts. Additionally, dip coating or whirl sintering methods may be used to apply the coating.

Stents coated according to these methods are characterized by the following properties: good biocompatibility; a uniform coating that is maintained upon expansion of the stent, adheres well to the stent surface and does not delaminate or crack upon expansion; a coating that degrades partly by surface erosion, in addition to bulk erosion, and is thus less likely to cause thrombosis as a result of small fragments of coating being released from the stent surface; a coating that is less likely to cause an inflammatory response; a coating that can be loaded with a drug or is compatible with surface coating by a drug; and a stent coating that can be sterilized by irradiation or treatment with ethylene oxide.

Due to the ductility and high elongation to break of P4HB, and copolymers thereof, stent coatings derived from these materials, and applied using the methods described herein, form exceptionally good coatings that maintain their structural integrity after stent expansion, as evidenced by SEM (Scanning Electron Microscopy). This is advantageous when compared to more brittle materials of limited ductility and low elongation to break.

V. Stent Deployment

The stents described herein can be deployed *in vivo* by any means appropriate to their design, such as self-expansion, a combination of self-expansion and balloon-expansion, or balloon-expansion without self-expansion. A preferred method of delivery is to mount the stent onto a balloon catheter, insert the stent system into the body at the desired position for delivery, and expand the balloon in a pressure range of 4 to 16 bar, more preferably 8 bar, to locate the stent against the luminal wall in the desired position.

Due to the greater flexibility, relatively low profile, and small diameter of the absorbable stents described herein, it may be possible to deploy these stents in positions that require navigation of difficult and narrow paths. The stents may be used for coronary, peripheral, urological, gastroenterological, neurological, esophageal and tracheal applications.

The present invention will be further understood by reference to the following non-limiting examples.

EXAMPLES

Example 1: Absorbable Coronary Stent from a Dip-Coated Stent Blank

Polymer tubes with an inner diameter of 1.0 or 1.4 mm were fabricated by dip-coating of stainless steel male cores into a 2 % w/w solution of a preferred composition of 70 % PLLA (Resomer® L214 from Boehringer Ingelheim Pharma), 20 % P4HB (TephaFLEX® biomaterial from Tepha Inc., Mw 300-600K) and 10 % TEC in chloroform. The dip-coating procedure was repeated until a mean wall thickness of $160 \pm 10 \mu\text{m}$ of the polymer tubes was achieved. Afterwards, the polymer tubes were

removed from the cores and washed twice in methanol and twice in water for 24 h each for solvent removal.

The polymer tubes were then machined with a CO₂ laser for the manufacture of balloon-expandable coronary stents with nominal dimensions in the dilated state of 3.0 and 3.5 mm diameter and various lengths from 10 – 25 mm, as established by SEM.

The stents were deployed with a balloon catheter that was inflated to 8 bar within 1 minute. The stents exhibited a recoil between 2 – 10 % upon balloon deflation and a collapse pressure of 0.3 – 0.7 bar. In contrast to other material compositions, such as a composition of P3HB, P4HB and TEC, no strut cracking was observed, as established by comparing detailed electron micrographs of struts of an absorbable polymeric stent made from a blend of poly-3-hydroxybutyrate (P3HB), poly-4-hydroxybutyrate (P4HB) and triethylcitrate (TEC) (70/20/10 % w/w/w) and from a blend of PLLA, P4HB and TEC (70/20/10 % w/w/w) after deployment. In contrast to the P3HB/P4HB/TEC stent, the PLLA/P4HB/TEC stent exhibits no strut cracking. SEM shows cracking of a stent composition of P3HB, P4HB and TEC dilated slowly over 7 minutes, compared to a P4HB, PLLA and TEC composition dilated much more rapidly (in 1 minute) which shows no cracking.

Example 2: Absorbable Coronary Stent From An Extruded Stent Blank

Polymer tubes with an inner diameter of 1.0 or 1.4 mm and a wall thickness of 150 µm were fabricated by extrusion of a preferred composition of 78 % PLLA (Resomer® L214 from Boehringer Ingelheim Pharma) and 22 % P4HB (TephaFLEX® biomaterial from Tepha Inc., Mw 300-600K).

The polymer tubes were then machined with a CO₂ or excimer laser for the manufacture of balloon-expandable coronary stents with nominal dimensions in the dilated state of 3.0 and 3.5 mm diameter and various lengths from 10 – 25 mm.

Example 3: Absorbable Polymeric Matrix for a Drug-Eluting Stent Coating

A 0.3 % w/w solution of P4HB (TephaFLEX® biomaterial from Tepha, Inc., Mw 300-600K) in chloroform was prepared. Metallic coronary stents were spray-coated with this solution until a mean coating layer thickness of 15-20 micrometer was achieved. After 24 h storage under vacuum to remove the chloroform, the stents were mounted on standard balloon catheters and afterwards deployed to a nominal diameter of 3.5 mm. Detailed electron micrographs of metallic stent struts coated with P4HB before and after stent dilation illustrate the smoothness and integrity of the coating before and after balloon expansion.

Example 4: Permanent Drug-Eluting Stent with Absorbable Polymeric Coating Matrix and Incorporated Antiproliferative Immunosuppressant (low dose)

A 0.3 % w/w solution of P4HB (TephaFLEX® biomaterial from Tepha, Inc., Mw 300-600K) and rapamycin (70/30 % w/w, polymer/drug) in chloroform was prepared. Metallic coronary stents were spray-coated with this solution until a mean coating layer thickness of 15-20 micrometer was achieved. After 24 h storage under vacuum to remove the chloroform, the stents were mounted on standard balloon catheters and afterwards deployed to a nominal diameter of 3.5 mm. Rapamycin as the active agent was released from the coating.

Example 5: Permanent Drug-Eluting Stent with Absorbable Polymeric Coating Matrix and Incorporated Antiproliferative Immunosuppressant (high dose)

A 0.3 % w/w solution of P4HB (TephaFLEX® biomaterial from Tepha, Inc., Mw 300-600K) and rapamycin (40/60 % w/w, polymer/drug) in chloroform was prepared. Metallic coronary stents were spray-coated with this solution until a mean coating layer thickness of 15-20 micrometer was achieved. After 24 h storage under vacuum to remove the chloroform, the stents were mounted on standard balloon catheters and afterwards deployed

to a nominal diameter of 3.5 mm. Rapamycin as the active agent was released from the coating.

Example 6: Absorbable Drug-Eluting Stent with Incorporated Antiproliferative Immunosuppressant (low dose)

Polymer tubes with an inner diameter of 2.8 mm were fabricated by dip-coating of stainless steel male cores into a 2 % w/w solution of a preferred composition of 70 % PLLA (Resomer® L214 from Boehringer Ingelheim Pharma), 20 % P4HB (TephaFLEX® biomaterial from Tepha Inc., Mw 300-600K) and 10 % TEC in chloroform. The dip-coating procedure was repeated until a mean wall thickness of $250 \pm 20 \mu\text{m}$ of the polymer tubes was achieved. Afterwards, the polymer tubes were removed from the cores and washed twice in methanol and twice in water for 24 h each for solvent and TEC removal.

The polymer tubes were then machined with a CO₂ laser for the manufacture of balloon-expandable peripheral vascular stents with nominal dimensions in the dilated state of 6.0 mm diameter and various lengths from 15–25 mm.

The stents were then spray-coated with a 0.3 % w/w solution of P4HB and rapamycin (70:30 % w/w, polymer/drug) in chloroform. After 24 h storage under vacuum to remove the chloroform, the stents were mounted on balloon catheters. The stents were deployed to a nominal I.D. of 6.0 mm with a balloon catheter, which was inflated to 8 bar within 1 minute. The stents exhibited a recoil of approximately 5 % upon balloon deflation and a collapse pressure greater than 0.6 bar. Rapamycin as the active agent was released from the coating.

Example 7: Absorbable Drug-Eluting Stent with Incorporated Antiproliferative Immunosuppressant (high dose)

Polymer tubes with an inner diameter of 2.8 mm were fabricated by dip-coating of stainless steel male cores into a 2 % w/w solution of a preferred composition of 70 % PLLA (Resomer® L214 from Boehringer Ingelheim Pharma), 20 % P4HB (TephaFLEX® biomaterial from Tepha Inc., Mw 300-600K) and 10 % TEC in chloroform. The dip-coating

procedure was repeated until a mean wall thickness of $250 \pm 20 \mu\text{m}$ of the polymer tubes was achieved. Afterwards, the polymer tubes were removed from the cores and washed twice in methanol and twice in water for 24 h each for solvent and TEC removal.

The polymer tubes were then machined with a CO_2 laser for the manufacture of balloon-expandable peripheral vascular stents with nominal dimensions in the dilated state of 6.0 mm diameter and various lengths from 15–25 mm.

The stents were then spray-coated with a 0.3 % w/w solution of P4HB and rapamycin (40/60 % w/w, polymer/drug) in chloroform. After 24 h storage under vacuum to remove the chloroform, the stents were mounted on balloon catheters. The stents were deployed to a nominal I.D. of 6.0 mm with a balloon catheter which was inflated to 8 bar within 1 minute. The stents exhibited a recoil of approximately 5 % upon balloon deflation and a collapse pressure greater than 0.6 bar. Rapamycin as the active agent was released from the coating by diffusion and also supported by the polymer degradation.

Example 8: Accelerated *in vitro* degradation behavior of an absorbable peripheral stent from a polymer blend material of high molecular weight PLLA and P4HB

Balloon-expandable absorbable peripheral stents with nominal dimensions in the dilated state of 6.0 mm x 25 mm were manufactured from a polymer blend of high molecular weight PLLA (Resomer® L214 from Boehringer Ingelheim Pharma), and P4HB (TephaFLEX® biomaterial from Tepha Inc., Mw 300-600K) with a mass ratio of 78/22 % or from pure high molecular weight PLLA (Resomer® L214). The stents were deployed with balloon catheters and then incubated *in vitro* in Sørensen buffer solution at 37°C to evaluate hydrolytic degradation *in vitro*.

After 0/2/4/8/12/24/48 weeks, stents were removed from storage and analyzed by gel permeation chromatography (GPC) to determine molecular weight. Figure 3 shows the accelerated decrease of molecular weight of the

polymer blend material of high molecular weight PLLA and P4HB in comparison to pure PLLA.

Example 9: *In vitro* drug release kinetics of permanent drug-eluting stents with absorbable polymeric coating matrix and incorporated antiproliferative immunosuppressant rapamycin showing the influence of base coat thickness, and of top coat thickness on the release profile

A 0.3 % w/w solution of P4HB (TephaFLEX® Biomaterial from Tepha, Inc., Mw 300-600K) and the immunomodulator rapamycin of 85:15 % w/w in chloroform was prepared. Metallic coronary stents were spray-coated with this solution as a base coat, until a coating layer thickness of 10-20 micrometer was achieved, which is equivalent to a drug content of 1-2 μg per mm^2 of stent surface area. The stents were then spray-coated with a 0.3 % w/w solution of P4HB until a top-coat thickness of 5-15 micrometer was achieved in order to establish different diffusion barriers for retarded drug release. After 24 h storage *in vacuo* to remove the chloroform, the stents were mounted on standard balloon catheters and afterwards deployed to a nominal diameter of 3.5 mm. The stents were then stored in 2 ml of a 0.9 % sodium chloride solution and incubated at 37 °C. After different time points, aliquots were taken from the elution medium for analysis of released drug, the elution medium was changed, and the stents were put back to storage. The aliquots were analyzed by HPLC.

Figure 4 displays the *in vitro* drug release profiles of sample stents, showing the effect of top coat and base coat thicknesses on the drug release profile at a constant drug concentration. The use of a top-coat retarded the drug release (stent 85:15-1 compared to stent 85:15-2). The use of a thicker top-coat retarded the drug release further (stents 85:15-1 and 85:15-2 compared to stent 85:15-3). The use of a thinner base coat reduces the amount of drug released (stent 85:15-1 compared to stent 85:15-4).

Example 10: *In vitro* drug release kinetics of permanent drug-eluting stents with absorbable polymeric coating matrix and incorporated antiproliferative immunosuppressant rapamycin showing the influence of drug content, and of base coat thickness on the release profile

A 0.3 % w/w solution of P4HB (TephaFLEX® biomaterial from Tepha, Inc., Mw 300-600K) and the immunomodulator rapamycin of 85:15 % w/w (polymer/drug), or 70:30 % w/w, or 40:60 % w/w in chloroform was prepared. Metallic coronary stents were spray-coated with either of these solutions as a base coat, until a coating layer thickness of 5-10 micrometer was achieved, which is equivalent to a drug content of 1-2 μg per mm^2 of stent surface area. The stents were then spray-coated with a 0.3 % w/w solution of P4HB until a top coat thickness of 10 micrometer was achieved in order to establish a diffusion barrier to retard drug release. After 24 h storage *in vacuo* to remove the chloroform, the stents were mounted on standard balloon catheters and afterwards deployed to a nominal diameter of 3.5 mm. The stents were then stored in 2 ml of a 0.9 % sodium chloride solution and incubated at 37°C. After different time points, aliquots were taken from the elution medium, the elution medium was changed, and the stents were put back to storage. The aliquots were analyzed by HPLC.

Figure 5 displays *in vitro* drug release profiles of sample stents, showing the increasing drug release at higher drug concentrations at a constant top-coat thickness. The elution rate and the total amount of drug released increases as the concentration of drug in the basecoat coat increases (stent 85:15-4 compared to stent 70:30-1 and stent 40:60-1).

Example 11: Safe mounting of an absorbable coronary stent onto a balloon catheter

Balloon-expandable absorbable stents with an inner diameter of 1.4 mm in the undilated state and a length of 10 mm were manufactured from a polymer blend of high molecular weight PLLA and P4HB (TephaFLEX® biomaterial from Tepha, Inc., Mw 300-600K) by melt extrusion followed by laser cutting. The polymeric stents were mounted without crimping on dedicated balloon catheter systems with nominal dimension of 3.5 mm x 10

mm in the dilated state. The balloon catheter system contained an inner support tube in the balloon region underneath the stent to enhance stent retention. The inner support tubing was made from an elastomeric material and provided an interference fit to hold the mounted stent in place. The diameter or durometer of the elastic support tubing could be modified to adjust the resistance of the interference fit and modify the stent retention. The dislodgment force of the stent systems were tested using a universal testing machine.

A mean stent dislodgement force of 2 N, and a maximum dislodgement force of greater than 5 N were measured. Without the inner support tubing, the dislodgment force was less than 0.3 N and would not be suitable for intravascular deployment without some method to hold the stent in place.

CLAIMS:

1. An absorbable biocompatible stent comprising a polymer composition, wherein the composition comprises
 - (a) one or more polymers selected from the group consisting of poly-4-hydroxybutyrate, copolymers of 4-hydroxybutyrate, and mixtures thereof and
 - (b) one or more polymers selected from the group consisting of polylactides, copolymers of lactic acid, and mixtures thereof,wherein the composition comprises between 60 and 98% by weight of a polylactide, wherein the stent is plastically expandable at normal body temperature, and is of a first diameter sufficient to be retained upon a balloon catheter for placement within a body lumen, and is expandable to a second diameter sufficient to be retained within the body lumen.
2. The stent of claim 1 wherein the stent can be expanded within a body lumen in less than 5 minutes
3. The stent of any one of claims 1-2 wherein the stent has a recoil of less than 10%.
4. The stent of any one of claims 2-3 wherein the stent can be expanded using a balloon pressure of 4-16 bar.
5. The stent of any one of claims 1-4 wherein the stent has a collapse pressure of at least 0.1 bar.
6. The stent of claim 5 wherein the stent has a collapse pressure of between 0.3-0.7 bar.
7. The stent of claim 1 wherein the stent does not creep at 100 mmHg for 7 days.
8. The stent of any one of claims 1-7 wherein the stent comprises struts and the struts do not kink or fracture during expansion.
9. The stent of any one of claims 1-8 wherein the stent does not significantly shorten upon expansion.
10. The stent of claim 8 wherein the thickness of the struts is less than 300 μ m.
11. The stent of any one of claims 1-10 wherein the first diameter of the stent is at least 1 mm, and the second diameter is at least 3 mm.

12. The stent of claim 11 wherein the stent can be expanded from the first diameter to the second diameter in less than one minute using a balloon pressure of 4-16 bar.
13. The stent of any one of claims 1-12 wherein the stent degrades in less than
5 two years.
14. The stent of claim 13 wherein the molecular weight of the stent decreases more than 50% in 40 weeks.
15. The stent of any one of claims 1-14 wherein the stent degrades in less than one year.
- 10 16. The stent of any one of claims 1-15 wherein the stent comprises between 2 and 40 % by weight of a homopolymer and/or copolymers of 4-hydroxybutyrate.
17. The stent of any one of claims 1-15 wherein the stent comprises between 75 and 95 % by weight of polylactides.
18. The stent of any one of claims 1-15 wherein the stent comprises:
15 (a) between 2 and 40 % by weight of homopolymer and/or copolymers of 4-hydroxybutyrate, and
(b) between 60 and 98% by weight of polylactides.
19. The stent of any one of claims 1-18 wherein the polymer composition further comprises a plasticizer.
- 20 20. The stent of claim 19 wherein the plasticizer is present between 0 to 10% by weight of the polymer.
21. The stent of any one of claims 19-20 wherein the plasticizer is triethylcitrate.
22. The stent of any one of claims 1-21 wherein the stent further comprises one
25 or more active agents.
23. The stent of claim 22 wherein the stent further comprises a top coat.
24. The stent of claim 22 wherein the active agent is selected from the group consisting of anti-inflammatory agents, immunomodulators, antiproliferatives, cytostatics, anti-migratory agents, extracellular matrix modulators, vascular healing
30 promoters, endothelialiation promoters, anticoagulants, antibiotics, anti-tumor agents, anti-cancer agents, and combinations thereof.

25. The stent of claim 22 wherein the agent is incorporated into the stent by a technique selected from the group consisting of spray coating, dip coating, whirl sintering, and combinations thereof.

26. The stent of claim 24 wherein the active agent(s) are released from the stent
5 *in vivo*.

27. The stent of any one of claims 1-26 wherein the stent further comprises one or more radio-opaque and/or contrast substances.

28. A method of mounting any one of the stents of claims 1-27 comprising mounting the stent on a balloon catheter wherein the stent dislodgement force is greater
10 than 1 N.

29. A method of making the stent of any one of claims 1-27, the method comprising fabricating a tube or stent blank using a technique selected from the group consisting of solution based dip-coating, casting, and combinations thereof or a technique selected from the group consisting of extrusion and injection molding.

15 30. The method of claim 29 wherein the stent blank consists of one or more fibers, which are fabricated in a melt based process including extrusion or injection molding, and which are optionally reinforced by a drawing process.

31. The method claim 29 or 30 wherein the stent is manufactured by laser machining of the tube or stent blank.

20 32. The method claim 30 wherein the stent is manufactured by winding, knitting, braiding, weaving or welding of one or several fiber blanks into a tubular stent structure.

33. The method of any one of claims 29-32 comprising applying an absorbable polymer coating to a tube or stent blank.

25 34. The method of claim 33, wherein the coating is applied by a technique selected from the group consisting of spray coating, dip coating, and whirl sintering.

35. A method of deploying the stent of any one of claims 1-27, comprising mounting the stent on a delivery system and balloon-expanding the stent inside the body lumen with a balloon pressure of 4 to 16 bar.

36. The stent of any one of claims 1-27, wherein the stent is used in a coronary, peripheral, urological, gastroenterological, neurological, esophageal or tracheal procedure.

37. The stent of any one of claims 1-27, wherein the stent is used in the local
5 administration of an active agent.

38. The stent of claim 37 wherein the active agent is selected from the group consisting of an anti-tumor agent, anti-cancer agent, an antibiotic agent, and combinations thereof.

39. A metal stent comprising an absorbable polymeric coating, wherein the
10 polymeric coating comprises (a) one or more polymers selected from the group consisting of poly-4-hydroxybutyrate, copolymers of 4-hydroxybutyrate, and mixtures thereof and (b) one or more polymers selected from the group consisting of polylactides, copolymers of lactic acid, and mixtures thereof, wherein the composition comprises between 60 and 98% by weight of a polylactide, wherein the stent coating is
15 plastically expandable at normal body temperature without cracking or delaminating and is of a first diameter sufficient to be retained upon a balloon catheter for placement within a body lumen, and is expandable to a second diameter sufficient to be retained within the body lumen.

40. The stent of claim 39 wherein the polymeric coating comprises poly-4-
20 hydroxybutyrate.

41. The stent of claim 39 or 40 wherein the polymeric coating degrades in less than two years.

42. The stent of any one of claims 39-41 wherein the polymeric coating comprises a copolymer containing 4-hydroxybutyrate monomeric units.

25 43. The stent of any one of claims 39-42 wherein the stent further comprises one or more active agents.

44. The stent of any one of claims 39-43 wherein the stent further comprises a top coat.

45. The stent of any one of claims 39-44 wherein the agent is selected from the
30 group consisting of anti-inflammatory agents, immunomodulators, antiproliferatives, cytostatics, anti-migratory agents, extracellular matrix modulators, vascular healing

promoters, endothelialiation promoters, anticoagulants, antibiotics, anti-tumor agents, anti-cancer agents, and combinations thereof.

46. The stent of claim 43 wherein the active agent(s) are released from the stent *in vivo*.

5 47. The stent of any one of claims 39-46 wherein the stent is used in a coronary, peripheral, urological, gastroenterological, neurological, esophageal or tracheal procedure.

48. The stent of any one of claims 39-46 wherein the stent is used in the local administration of an active agent.

10 49. The stent of claim 48 wherein the active agent is selected from the group consisting of anti-tumor agents, anti-cancer agents, antibiotic agents, and combinations thereof.

50. The stent of any one of claims 1 to 27 or 39 to 49, wherein the stent is capable of being sterilized by gamma-irradiation, electron beam, or treatment with
15 ethylene oxide.

51. A method of making the stent of any one of claims 39 to 50, wherein the coating is applied in a spray coating process, dip coating process, and/or a whirl sintering process.

52. The method of claim 51 wherein the coating does not form a web-like
20 structure on the stent.

53. A stent formed from the method of claim 52.

54. A method of deploying the stent of any one of claims 39 to 50, wherein the stent is mounted on a delivery system and balloon-expanded with a balloon pressure of 4 to 16 bar.

25 55. The method of deploying the stent of claim 54, wherein the stent is mounted on a delivery system and is self-expanded.

56. A method to accelerate the degradation of an absorbable stent comprising blending an absorbable polymer with poly-4-hydroxybutyrate or copolymers thereof.

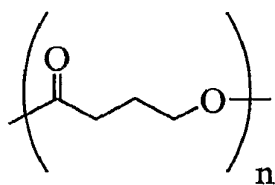


FIG. 1

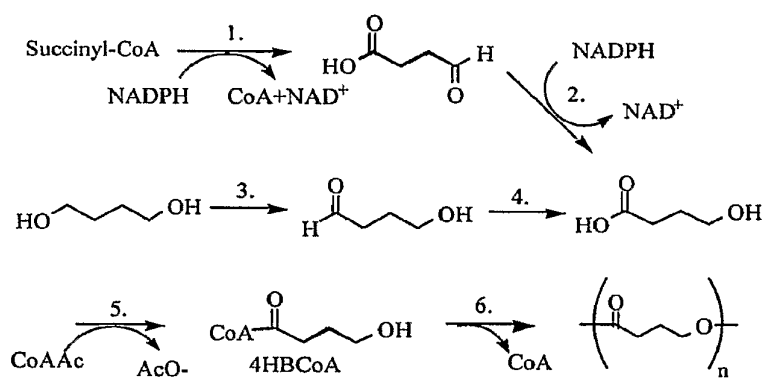


FIG. 2

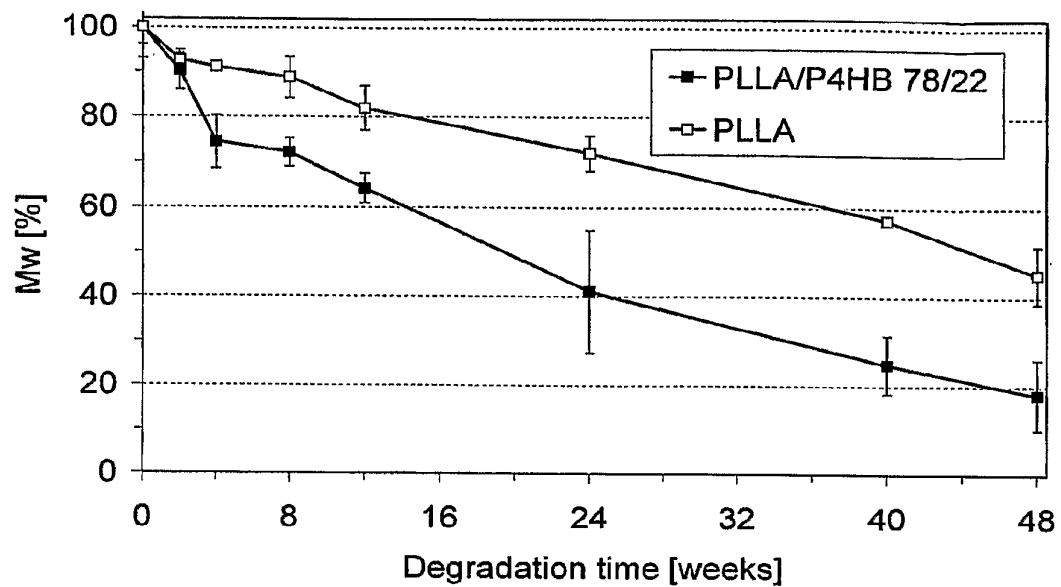


FIG. 3

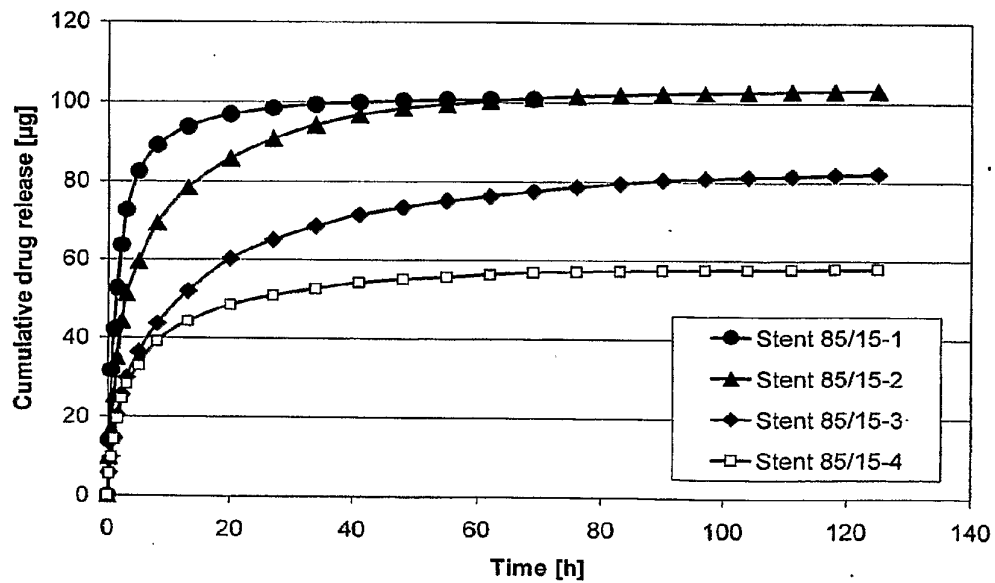


FIG. 4

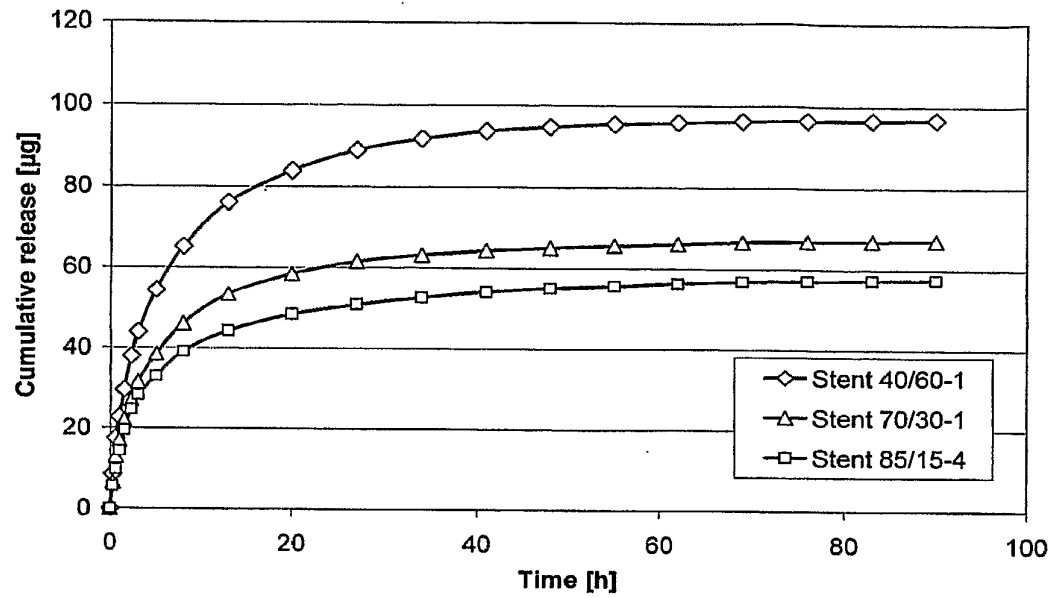


FIG. 5