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(54) **METHODS OF MINIMIZING STENT CONTRACTION FOLLOWING DEPLOYMENT**

(75) Inventor: **Patrick Sabaria**, Saint Nom La Breteche (FR)

Correspondence Address:
TAROLLI, SUNDHEIM, COVELL & TUMMINO L.L.P.
1300 EAST NINTH STREET, SUITE 1700
CLEVEVLAND, OH 44114 (US)

(73) Assignee: **ARTERIAL REMODELING TECHNOLOGIES S.A.**, Noisy Le Roi (FR)

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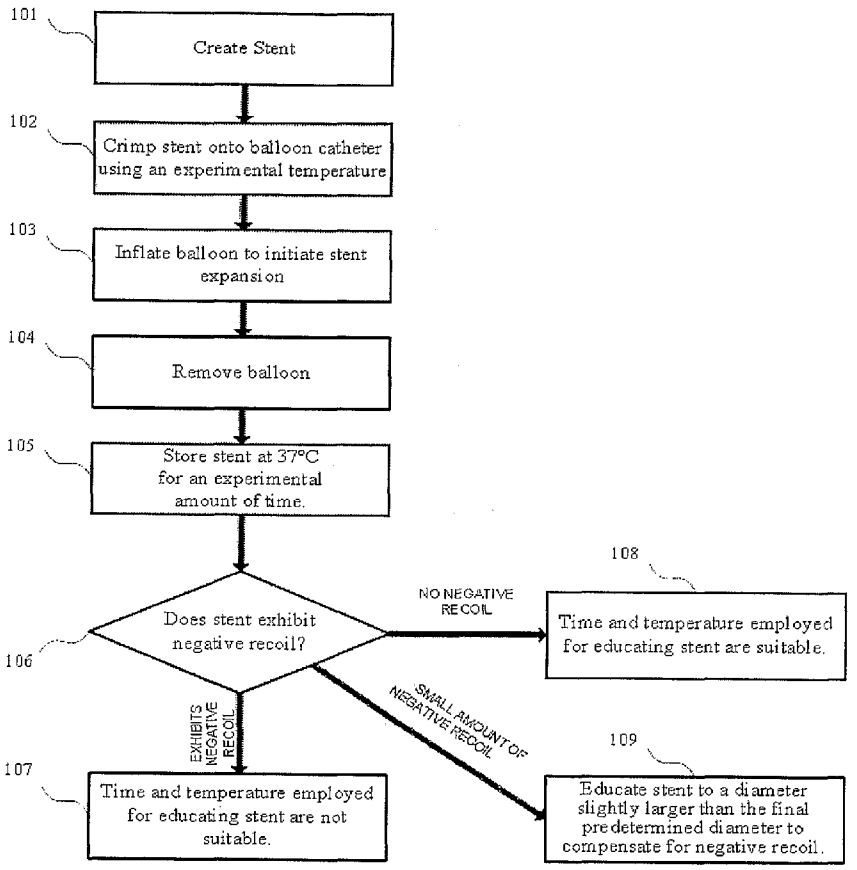
(57) **ABSTRACT**

The present invention provides methods for fabricating a stent using a preheating stage. The inventors have found a fabrication methods that result in the same and/or better product quality stent using a single step process performed at a temperature of below 65° C., more preferably below 60° C., most preferably below 55° C. Stent fabrication under such reduced temperature conditions reduces the exposure of the stent to adverse temperature conditions, thereby enabling the greater retention of the polymer's memory. Further, upon expansion, the stent does not contract to a smaller diameter but instead remains at a constant diameter or increases to a larger diameter.



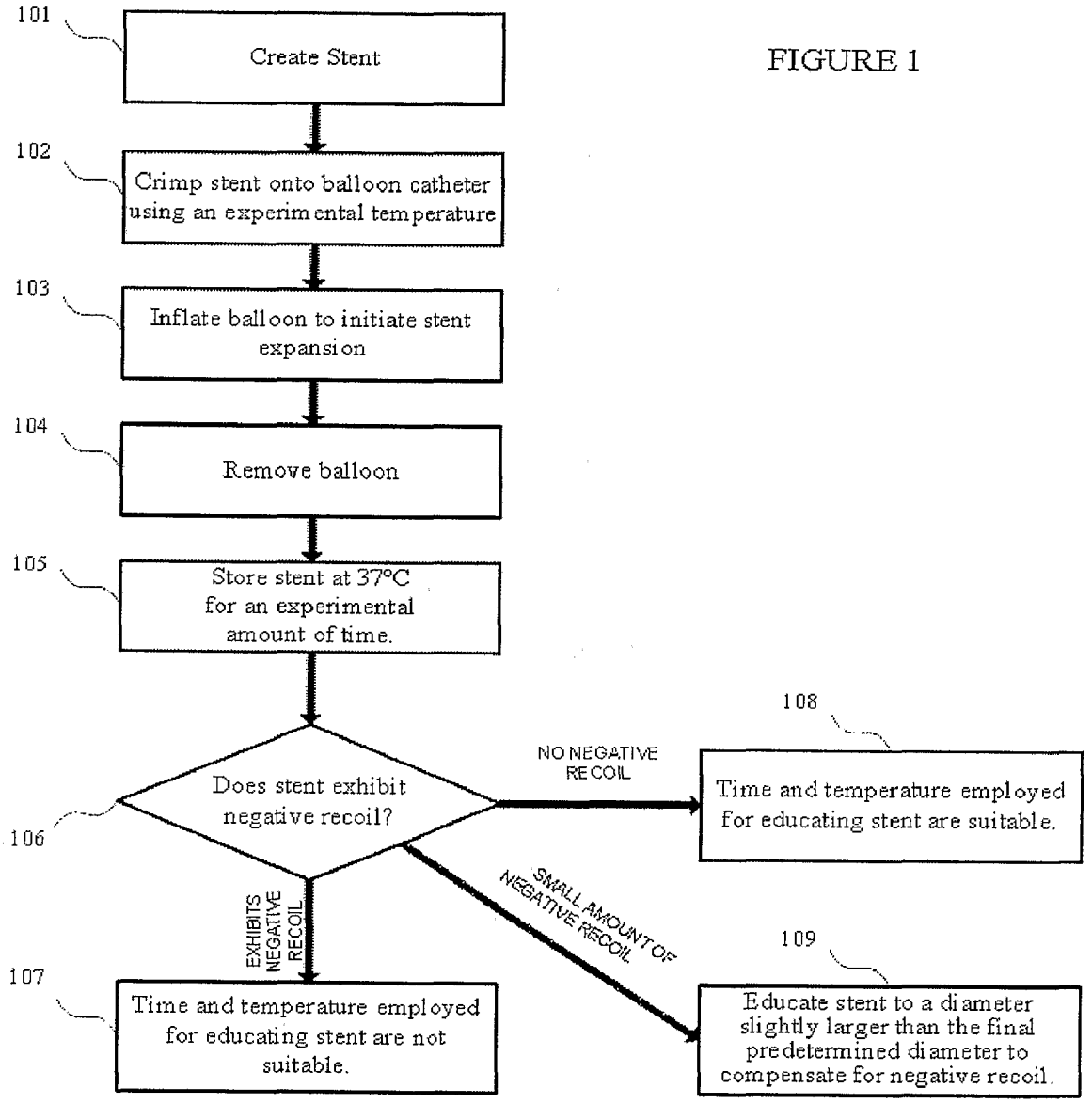

101
Create Stent
102
Crimp stent onto balloon catheter using an experimental temperature
103
Inflate balloon to initiate stent expansion
104
Remove balloon
105
Store stent at 37°C for an experimental amount of time.
106
Does stent exhibit negative recoil?
NO NEGATIVE RECOIL
108
Time and temperature employed for educating stent are suitable.
EXHIBITS NEGATIVE RECOIL
SMALL AMOUNT OF NEGATIVE RECOIL
107
Time and temperature employed for educating stent are not suitable.
109
Educate stent to a diameter slightly larger than the final predetermined diameter to compensate for negative recoil.

FIGURE 1

METHODS OF MINIMIZING STENT CONTRACTION FOLLOWING DEPLOYMENT

BACKGROUND OF THE INVENTION

[0001] The use of stents in various surgical, interventional cardiology, and radiology procedures has quickly become accepted as experience with stent devices accumulates and as the advantages of stents become more widely recognized. Stents are often used in body lumens to maintain open passageways such as the prostatic urethra, the esophagus, the biliary tract, intestines, and various coronary arteries and veins, as well as more remote cardiovascular vessels such as the femoral artery.

[0002] Stents are often used to treat atherosclerosis, a disease in which vascular lesions or plaques consisting of cholesterol crystals, necrotic cells, lipid pools, excess fiber elements and calcium deposits accumulate in the walls of an individual's arteries. One of the most successful procedures for treating atherosclerosis is to insert a deflated balloon within the lumen, adjacent the site of the plaque or atherosclerotic lesion. The balloon is then inflated to put pressure on and "crack" the plaque. This procedure increases the cross-sectional area of the lumen of the artery. Unfortunately, the pressure exerted also traumatizes the artery, and in 30-40% of the cases, the vessel either gradually renarrows or recloses at the locus of the original stenotic lesion. This renarrowing is known as restenosis

[0003] A common approach to prevent restenosis is to deploy a metallic stent to the site of the stenotic lesion. Although metallic stents have the mechanical strength necessary to prevent the retractile form of restenosis, their presence in the artery can lead to biological problems including vasospasm, compliance mismatch, and even occlusion. Moreover, there are inherent, significant risks from having a metal stent permanently implanted in the artery, including erosion of the vessel wall. The stents may also migrate on occasion from their initial insertion location raising the potential for stent induced blockage. Metal stents, especially if migration occurs, cause irritation to the surrounding tissues in a lumen. Also, since metals are typically much harder and stiffer than the surrounding tissues in a lumen, this may result in an anatomical or physiological compliance mismatch, thereby damaging tissue or eliciting unwanted biologic responses. In addition, the constant exposure of the stent to the blood can lead to thrombus formation within the blood vessel. Stents also allow the cellular proliferation associated with the injured arterial wall to migrate through the stent mesh, where the cells continue to proliferate and eventually lead to the narrowing of the vessel. Further, metal stents typically have some degree of negative recoil. Finally, metallic stents actually prevent or inhibit the natural vascular remodeling that can occur in the organism by rigidly tethering the vessel to a fixed, maximum diameter.

[0004] Because of the problems of using a metallic stent, others have recently explored use of bioabsorbable and biodegradable materials stents. The conventional bioabsorbable or bioresorbable materials from which such stents are made are selected to absorb or degrade over time. This degradation enables subsequent interventional procedures such as restenting or arterial surgery to be performed. It is also known that some bioabsorbable and biodegradable materials tend to have excellent biocompatibility characteristics, especially in comparison to most conventionally used biocompatible metals. Another advantage of bioabsorbable and biodegradable stents is that the mechanical properties can be designed to substantially eliminate or reduce the stiffness and hardness that is often associated with metal stents. This is beneficial because the metal stent stiffness and hardness can contribute to the propensity of a stent to damage a vessel or lumen. Examples of novel biodegradable stents include those found in U.S. Pat. No. 5,957,975, which is incorporated by reference in its entirety.

[0005] There are, however, still problems with many biodegradable stents. For example, testing in animals has shown that biodegradable stents still suffer from multiple complications, including relaxation-related negative recoil, lack of sufficient radial strength, difficulty in deployment and distal migration of the entire stent or portions thereof and formation of an occlusive thrombus within the lumen of the stent.

[0006] Accordingly, it is desirable to have a new stent that overcomes the disadvantages of the current stent designs. A polymer-based stent that exhibits little to no relaxation-related negative recoil when implanted in the blood vessel or duct of a mammalian subject is desirable. Indeed, it is preferred that the stent have a positive recoil. It is also desirable to have a polymer-based stent assembly that does not require a mechanical restraint to prevent the stent from expanding when stored at room temperature. To achieve these goals, the stent is fabricated using several heating steps. For instance, in a typical fabrication there is at least one preheating stage performed prior to the cutting procedure. Crimping step is performed as a two-step process at a temperature of 65° C. This method, however, has the drawback in that the multiple heating steps alter the stent "memory" of the ideal final diameter.

[0007] The inventors have found a novel method of stent fabrication that decreases the time the stent is exposed to adverse temperature condition, thereby enabling greater memory retention of the polymers diameter.

SUMMARY OF THE INVENTION

[0008] The present invention provides methods for fabricating a stent using a preheating stage. As opposed to using a two or more step heating process, the inventors have found a preferable embodiment using new fabrication methods that result in the same and/or better product quality stent using a single step process performed at a temperature of below 65° C., more preferably below 60° C., most preferably below 55° C. In certain embodiments, a temperature below about 50° C. is most preferred. Stent fabrication under such reduced temperature conditions permits technicians to avoid burning their hands but more importantly, results in a reducing the exposure of the stent to adverse temperature conditions, thereby enabling the greater retention of the polymer's memory.

[0009] Maintaining the stent at a temperature of fabrication as described also provides a beneficial result by being below the glass transition temperature of the polymeric material. Under the previously employed procedure(s), the stent would be expanded to balloon nominal diameter of: coronary balloon from typically 2.0 mm to 4.5 mm; vascular peripheral balloon (PTA) from typically from 3 mm to more than 20 mm depending on balloon diameter. For example, for a 3 mm balloon, the stent would be expanded to 3 mm and when the balloon was removed, it would contract to a diameter of 2.7, followed by a slow expansion to the final desired diameter of 3.2 mm. Under the presently employed procedure at T=zero, the stent is expanded to 3 mm when deployed. Following

initial deployment the stent does not contract, but instead remains at a diameter of 3 mm and gradually over time expands to the desired diameter of 3.2 mm. Furthermore, this deployment to a final desired diameter is essentially balloon inflation independent.

DETAILED DESCRIPTION

Definitions

[0010] "Bioresorbable polymer" as used herein refers to a polymer whose degradation by-products can be bio-assimilated or excreted via natural pathways in a human body.

"Crimping" as used herein refers to a process that involves radial pressing on a polymeric cylindrical device having slits, or openings in the wall thereof to allow a decrease in the diameter of the device without substantially affecting the thickness of the wall or struts of the cylindrical device. Such process, typically also results in an increase in length of the cylindrical device.

"Degradable polymer" or "biodegradable polymer" as used herein refers to a polymer that breaks down into monomers and oligomers when placed in a human body or in an aqueous solution and maintained under conditions of temperature, osmolality, pH, etc., that mimic physiological media preferably without involving enzymatic degradation to minimize the risk of triggering the antigenantibody defense system of the human body.

"Final predetermined shape and diameter" as used herein refers to the desired diameter, length, design and wall thickness of a stent that has been deployed to a target site in a vessel, particularly a blood vessel, duct, or tube in a mammalian subject, particularly a human subject.

"Negative recoil" as used herein refers to an undesirable decrease in the size or diameter of an expanded stent after initial deployment.

"Positive recoil" as used herein refers to an increase in the size or diameter of a stent that has been educated to have a desired final diameter but has not been fully expanded to the desired final diameter.

"Relaxation-related recoil" as used herein refers to the slow change in dimensions of a polymeric device due to a time-dependent slow rearrangement of molecule conformations according to a well-known behavior of viscoelastic polymeric matters. Such rearrangement is due to thermal agitation that slowly leads the polymeric material to a thermodynamic equilibrium typical of the storage conditions when it has been processed under different environmental conditions. Relaxation is very slow below Tg, i.e. when the matter is in the glassy state.

"Tg" or "glass transition temperature" as used herein refers to the temperature at which a polymer changes from a rubbery state to a glassy state and vice versa.

[0011] The present invention provides a stent fabrication method that only requires a single step process performed at the temperature of below 65° C., more preferably below 60° C., most preferably below 55° C. In certain embodiments, a temperature below about 50° C. is most preferred. The process that results in the same and/or better quality product that methods that require more than one heating steps. Maintaining the stent at a temperature of the invention or below also provides a beneficial result by avoiding the glass transition temperature of the polymeric material. This new stent fabrication method also results in the final stent having no negative recoil when the stent is deployed into a mammalian body. Indeed, the final stent has positive recoil when deployed.

[0012] Under the previously employed procedure(s), the stent would be expanded to 3 mm and when the balloon was removed, it would contract to a diameter of 2.7 mm, followed by a slow expansion to the final desired diameter of 3 mm. Under the presently employed procedure at T=zero, the stent is expanded to 3 mm when deployed. Following initial deployment the stent does not contract, but instead remains at a diameter of 3 mm. Further, over several days, the stent may further expand to a larger desired final diameter, say, of approximately 3.2 mm. In one embodiment, a balloon is used merely as a carrier for the stent through the body. Further, the deployment of the stent into the body may use the balloon as a carrier that also initiates the stent expansion. It is also contemplated that the deployment of the stent into the body may be balloon inflation independent. In addition, it is contemplated that the stent final diameter be larger than that of the deploying balloon diameter.

I. Stent Fabrication and Properties

[0013] The stents may be formed from any biodegradable, biocompatible, bioresorbable polymer, preferably a thermoplastic polymer. As used herein, a bioresorbable polymer is one whose degradative products are metabolized in vivo or excreted from the body via natural pathways. Preferably, the stent of the present assembly is formed from a degradable and bioresorbable polymer having a Tg at least 8 degrees above 37° C., preferably at least 20 degrees above 37° C. The polymer of the stent can be a homopolymer or a copolymer. Preferably, the stent is formed from a thin layer of one or more amorphous, bioresorbable polymers, i.e., the polymers used to form the stent preferably are not crystalline. It is also preferred that the polymers used to form the stent do not generate crystalline residues upon degradation in vivo. It is also contemplated that the chains of the polymer may be or may not be cross-linked. Light cross-linking, however, is acceptable provided that thermal and viscoelastic characteristics that allow education, crimping, and deployment of the device are sufficiently maintained.

[0014] Appropriate biodegradable polymers may include, but are not limited to, poly(L-lactide), polyglycolide, poly(D, L-lactide), copolymers of lactide and glycolide, polycaprolactone, polyhydroxyvalerate, polyhydroxybutyrate, polytrimethylenecarbonate, polyorthoesters, polyanhydrides, and polyphosphazenes. Examples of the types of polymers that are suitable for the stent of the present invention include, but are not limited to, lactic acid-based stereocopolymers (PLAx copolymers composed of L and D units, where X is the percentage of L-lactyl units) (55<Tg<60), copolymers of lactic and glycolic acids (PLAxGAy, where X, the percentage of L-lactyl units, and Y, the percentage of glycolyl units, are such that the Tg of the copolymer is above 45° C.), and Poly(lactic-co-glycolic-co-gluconic acid) where the OH groups of the gluconyl units can be more or less substituted (pLAx-GayGLx, where X, the percentage of L-lactyl units, and Y, the percentage of glycolyl units, and Z the percentage of gluconyl units are such that the Tg of the terpolymer is above 45° C.). Other suitable polymers include, but are not limited to, polylactic acid (PLA), polyglycolic acid (PGA) polyglactin (PLAGA copolymer), polyglyconate (copolymer of trimethylene carbonate and glycolide, and a copolymer of polyglycolide or lactide acid or polylactic acid with ϵ-caprolactone),

provided that the polymer has a glass transition temperature, Tg, of at least 45° C. or greater.

[0015] In one preferred embodiment, the stent comprises a polylactic acid stereocopolymer produced from L and DL lactides. The polymer is designated herein as "PLAX" where X represents the percentage of the L-lactic acid units in the mixture of monomers used to prepare the lactides. Preferably X is in the range of 10 to 90, more preferably 25 to 75. In another preferred embodiment, the stent comprises a polylactic acid, glycolic acid copolymer produced from L and DL lactides and glycolides. The polymer is designated herein as "PLAXGAY" where Y represents the percentage of glycolic acid units in the mixture of monomers used to prepare the copolymers. Preferably, the copolymers do not contain glycolyl repeating units since such units are known to be more inflammatory than lactyl repeating units. Preferably, the polymers are prepared using Zn metal or Zn lactate as initiator. To ensure good initial mechanical properties of the stent, the molecular weight of the polymer in the region having the second in vivo lifetime is preferably above 20,000 daltons, more preferably 100,000 daltons or larger. The polydispersity, I=Mw/Mn, is preferably below two and should not greatly reflect the presence of low molecular weight oligomers smaller than 2,000 daltons as determined by size exclusion chromatography. Optionally, the polymeric layer used to make the stent may be impregnated with an anticoagulant agent, such as heparin, anti-oxidants, such as vitamin E, compounds that regulate cellular proliferation, or anti-inflammatory drugs, such as corticosteroids, to provide localized drug delivery. Such drugs are incorporated into the polymeric layer using techniques known in the art. Agents may also be incorporated into the base polymer that forms the body of the stent, as long as the incorporation does not have significant adverse effects on the desired physical characteristics of the stent such as during radial stent deployment and degradation time. For intravascular stents, it is preferred that the film have a thickness of from about 0.05 mm to 0.2 mm.

[0016] It is contemplated that the stent may be made by any method. In one preferred embodiment, the stent is a formed from a biodegradable polymeric band comprising a head having a slot and a tongue comprising a catch or locking mechanism proximate the longitudinal edge thereof. The cylindrical element, which has an inner and outer surface, is formed by inserting a portion of the tongue through the slot to provide a cylindrical element having a first reduced diameter configuration. Following deployment, the cylindrical element is in a second expanded diameter configuration wherein the distal catch mechanism engages the inner surface of the head and prevents radial collapse or recoil of the polymeric stent. In a second preferred embodiment, the stent is formed from a plurality of interconnected polymeric bands each of which comprises a head having a slot and a tongue comprising a catch mechanism proximate the longitudinal edge thereof.

[0017] In one embodiment, the stent is formed by laser cutting of a cylindrical tube. In another embodiment, the stent is formed by laser cutting a flat polymeric sheet in the form of the stent, and then rolling the pattern into the shape of the cylindrical stent and providing a longitudinal weld to form the stent. In another embodiment, the stents are created by chemically etching a flat polymeric sheet and then rolling and welding it to form the stent, or coiling a polymeric wire to form the stent.

[0018] In another preferred embodiment, the stent may also be formed by molding or injection molding of a thermoplastic or reaction injection molding of a thermoset polymeric material. The flat grid is then rolled and extremities are welded or glued to form a cylinder. Filaments of the compounded polymer may be extruded or melt spun. These filaments can then be cut, formed into ring elements, welded closed, corrugated to form crowns, and then the crowns welded together by heat or solvent to form the stent. Lastly, hoops or rings may be cut from tubing stock, the tube elements stamped to form crowns, and the crowns connected by welding or laser fusion to form the stent.

[0019] Generally, the struts are arranged in patterns that are designed to contact the lumen walls of a vessel and to maintain patency of the vessel thereby. A myriad of strut patterns are known in the art for achieving particular design goals.

[0020] It is contemplated that a crimped stent may incorporate slits or open spaces to allow for the temporary reduction in diameter of the cylindrical tube without substantially altering the wall thickness. Moreover, a stent embodying the present invention can include teeth and corresponding catching structure that operates to maintain an expanded form. Moreover, polymer based stents embodying structure defined by a wire or ribbon coil or helix or a knitted mesh configuration are possible examples of self-expanding stents. Other important design characteristics of stents include radial or hoop strength, expansion ratio or coverage area, and longitudinal flexibility. One strut pattern may be selected over another in an effort to optimize those parameters that are of importance for a particular application.

[0021] It is also contemplated that the biodegradable stent may have a programmed pattern of in vivo degradation. Stent polymeric structure allows for differential speed degradation. See, for example, U.S. Pat. No. 5,957,975, the entirety of which is incorporated by reference. In one embodiment, the stent comprises at least one substantially cylindrical element having two open ends and a plurality of regions circumferentially spaced around the cylindrical element and extending from one open end to the other open end of the cylindrical element. Each of the regions is configured or designed to have a desired in vivo lifetime. At least one of the regions is designed to have a shorter in vivo lifetime than the other region or regions. This means that the region having the shorter in vivo lifetime degrades sooner after deployment than the regions having a longer in vivo lifetime. Thus, when stents designed in accordance with the present invention are deployed within the lumen of a vessel of a patient, the cylindrical element acquires one or more fissures which extend from one open end of the cylindrical element to the other open end of the cylindrical element within a desired, predetermined period of time after the stent is deployed in the patient. It has been determined that such fragmentation within a predetermined period of time after deployment allows for enlargement of the lumen of the vessel via the process of arterial remodeling.

[0022] The regions of the stent with the different in vivo lifetimes can be made in a variety of ways. Preferably, such stents are made by producing regions having a first in vivo lifetime, i.e., a shorter in vivo lifetime, in a polymeric layer having the predetermined second, or longer, in vivo lifetime. The regions having the first in vivo lifetime are produced by heating the respective regions of the polymeric layer having a second in vivo lifetime for a time and at a temperature sufficient to cause local partial degradation of the polymeric

chains. Such treatment, which can be accomplished using a piloted hot needle, laser beam, or flow of hot air, renders the polymer in the heated region more sensitive to hydrolytic degradation. Alternatively, the regions having a first in vivo lifetime may be produced in a polymeric layer having a second in vivo lifetime by incorporating a sufficient number of acidic ions into the respective regions of the polymeric layer. Preferably, the acidic ions are provided by compounds that are not soluble in blood.

[0023] Regions having a first in vivo lifetime can also be produced in a polymeric film having a second in vivo lifetime by exposure of the respective regions to beta radiation or gamma radiation for a sufficient time to induce partial cleavage of the polymeric chains within the respective regions. Provided the polymeric layer has a thickness of less than 0.3 mm, regions having a first in vivo lifetime can also be produced in a polymeric film having a second in vivo lifetime by introducing areas of mechanical weakness into the polymer. One method of introducing mechanical weakness is by reducing the thickness of the polymer in the respective region or forming holes therein. Regions having a first in vivo lifetime can also be produced in a polymeric film having a second in vivo lifetime by applying mechanical stress to the respective region. However, this latter process is difficult to control and, thus, is less preferred. Differing lifetimes can also be created by providing one or more different coatings over different regions of the biodegradable stent.

[0024] Another method for producing a polymeric layer in which one region or a plurality of spaced apart regions have a first in vivo lifetime and other regions have a second in vivo lifetime is to incorporate strips or fibers of a faster degrading bioresorbable polymer into a film made from a slower degrading polymer. For example, a mesh or a parallel array of fibers or strips of PGA or any other faster degrading bioresorbable polymer can be embedded into the respective regions of a polymeric film of PLA that may be designed to be slower degrading. Embedding can be achieved by inserting the mesh or fibers between two melted sheets of the slower degrading polymer. Provided the relative solubilities are compatible, the fibers or mesh can be placed in an organic solution of the slower degrading polymer and the desired polymeric film formed by evaporation of the organic solvent. One example of a method for embedding a mesh made from one polymer into a polymeric layer made from a second polymer is described in U.S. Pat. No. 4,279,249 issued to Vert et al. on Jul. 21, 1981, which is specifically incorporated herein by reference. A stent having the desired shape and orientation of regions is then formed from the polymeric layer by standard techniques such as stamping, employing a laser beam, or any other technique used in the art to tool a polymeric film.

[0025] The initial polymeric cylindrical device that is formed by any of these processes can be configured to have the final predetermined shape, length, wall thickness and diameter, all of which are tailored to the application for which the stent is to be utilized. For example, for cardiovascular applications the initial polymeric device that is formed by these processes can have a final predetermined length ranging from 0.5 cm to approximately 3 cm. For certain applications, the initial polymeric cylindrical device can have a final, predetermined diameter ranging from 0.50 mm to 8.0 mm with a final, predetermined wall thickness ranging from 0.05 to 0.5 mm. Alternatively, the initial cylindrical device that is formed by any of these processes can have a smaller diameter than the final predetermined diameter.

[0026] In those instances where the initial polymeric cylindrical device has a smaller diameter than the final predetermined diameter, slits or openings are formed in the cylindrical device as described above, and then the cylindrical device is deformed or expanded to the final shape and diameter. This can be achieved by inserting an expandable device such as a balloon into the stent.

II. Educating and Crimping the Stent

[0027] While it is at the final predetermined shape, size, and diameter, the cylindrical device is educated by heating the device to a temperature above the Tg of the polymer from which the device is formed. The device is heated for a time sufficient to erase any former process-related memory and to impart a new memory of the final predetermined shape and diameter to the polymeric cylindrical device. It is believed that such conditions allow the polymer chains to relax and reorganize themselves from an entanglement typical of the former processing stages to an entanglement typical of the high temperature at which the cylindrical device is compatible with the final or deformed shape and size. When the polymeric cylindrical device has an initial diameter that is less than the final predetermined diameter, it is desired to heat to a temperature well above the Tg of the polymer. This heating step erases the anisotropic stresses promoted by the extrusion or molding process and the former processing-related memory of the polymer chains. Good results have been obtained by heating a laser-precut polymeric cylindrical device formed from PLA75 and deformed from a diameter of 1.0 mm to 4 mm at a temperature of 80° C. for 30 minutes. Temperatures of from about 45° C. to about 120° C. and times of 5 minutes or more should be suitable for educating stents made from PLAx with $0<X<100$, PLAxGAy with $0<x<25$ and $75<Y<100$, or any PLAxGAyGLz.

[0028] The polymeric cylindrical device is then crimped. "Crimping" as used herein refers to a process that involves radial pressing on a polymeric cylindrical device having slits, or openings in the wall thereof to allow a decrease in the diameter of the device without substantially affecting the thickness of the wall or struts of the cylindrical device. Such process may also result in an increase in length of the cylindrical device.

[0029] To crimp the educated cylindrical device, it is mounted onto a device with a smaller diameter. The diameter of the educated cylinder is reduced by heating the cylinder to a temperature below the Tg of the polymer while evenly applying pressure on the exterior surface of the wall of the cylindrical device.

[0030] In some embodiments, the polymeric stent is crimped onto an inflatable device such as an inflatable balloon catheter. In this instance, the stent assembly after crimping comprises an inflatable balloon catheter and an expandable, educated, polymeric stent snugly and stably disposed thereon. Slits or open spaces that allow for a reduction in diameter of the cylindrical device without substantially altering the wall thickness during crimping are incorporated into the cylindrical device prior to the time the cylindrical device is crimped on the inflatable balloon catheter. The temperature at which the cylindrical device is heated during crimping is high enough to allow reduction in diameter of the cylindrical device but low enough to not erase the memory of the final predetermined shape and diameter of the educated cylindrical device. Ideally, the temperature is less than the glass transition state of the polymer. More preferably, the temperature is

at about 50° C. Thus, the temperature at which the educated cylindrical device is heated during crimping is less than the temperature at which the cylindrical device is heated during education of the cylindrical device. Further, the time it takes to crimp the educated cylindrical device can vary, depending upon the temperature, size and composition of the stent

[0031] In accordance with the present method, expansion of the polymeric stent can be achieved by any means. In one embodiment, a balloon is used merely as a carrier for the stent through the body. In this preferred embodiment, the stent expansion occurs by the positive recoil properties of the stent; thus, the expansion is balloon inflation independent. In another preferred embodiment, a balloon is inflates and/or heated to initiates the stent expansion. It is contemplated that the positive recoil properties of the stent would expand the stent to its final predetermined diameter. The temperature used to initiate the stent expansion may be any temperature at or below the Tg of the polymer. In a less preferred embodiment, a balloon is inflated to expand the polymeric stent to its final predetermined shape.

[0032] In another aspect, the method of the present invention starts with a polymeric tube whose diameter initially is less than the final predetermined diameter. Such tube is first heated to a temperature close to or above the Tg of the polymer and expanded to provide a cylindrical device whose diameter is equal to the final desired diameter. Thereafter the cylindrical device is educated as described above to provide an educated cylindrical device having a memory of the final predetermined shape and diameter, and then crimped on a balloon catheter as described above to provide an assembly comprising the balloon catheter and an expandable, educated, polymeric stent snugly and stably disposed thereon.

[0033] The present invention also provides an assembly comprising an inflatable balloon catheter and a polymeric stent prepared in accordance with the present method.

[0034] Advantageously, the stent of the present invention exhibits little to no relaxation-related negative recoil when deployed in the blood vessel of a subject. Advantageously, the assembly of the present invention has a diameter that allows it to be easily inserted into a blood vessel of the subject and advanced to a target site. Advantageously, the stent of the present invention exhibits expansion (positive recoil) and adaptation to the geometry of the artery when the stent is not fully deployed up to its final diameter during deployment. Positive recoil over several days will create outward radial pressure for long periods of time. This outward radial pressure aids in positive vascular remodeling by assisting stabilizing the injured artery or vulnerable plaque, assist in cellular progress to repair injury of original acute expansion, assist in security of tissue flaps, and the like.

[0035] In addition, the stent of the present invention is stably disposed on the balloon, meaning that a mechanical restraint is not required to prevent the stent from rapidly expanding to its final diameter during storage at room temperature. Thus, although not required, the assembly of the present invention, optionally, also comprises a retractable sheath covering the exterior surface of the stent. Such sheath serves to prevent deformation of the stent and preclude or slow expansion during storage.

III. Procedures for Determining Times and Temperatures for Educating and Crimping the Stent of the Present Invention

[0036] FIG. **1** shows one embodiment of the invention. FIG. **1** is a method for determining the temperatures and times suitable for educating the cylindrical device. This method results in a stent that is resistant to negative recoil, and in fact has positive recoil. Once the stent is created (step **101**), the stent is crimped onto a balloon catheter according to one of the above methods (step **102**). The temperature of at which the stent is heated during crimping is high enough to allow reduction in diameter of the stent but low enough to not erase the memory of the final predetermined shape and diameter of the educated stent. Ideally, the temperature is less than the glass transition state of the polymer. More preferably, the temperature is at about 50° C. Thus, the temperature at which the educated stent is heated during crimping is less than the temperature at which the stent is heated during education of the stent. Further, the time it takes to crimp the educated stent can vary, depending upon the temperature, size and composition of the stent. The method of FIG. **1** may be used to assist in determining the ideal temperature to use to educate the stent.

[0037] The balloon is then partially or fully inflated to initiate stent expansion (step **103**). The balloon is removed (step **104**) and the stent is stored at a temperature appropriate for storage (step **105**). In one preferred embodiment, the storage temperature is 37° C. While in storage, the stent may increase in diameter because of the positive recoil properties of the stent. After a predetermined amount of time, the stent will be examined to find if it exhibits negative recoil (step **106**). In one preferred embodiment, the amount of time is 4 to 6 weeks. In another preferred embodiment, the amount of time is the estimated for an artery wall to recover from PTC angioplasty. If the stent exhibits little to no negative recoil when stored under these conditions, the times and temperatures employed for educating the stent are suitable (step **108**). In those cases where the polymeric stent exhibits a small amount of recoil, the cylindrical device can be educated at a diameter slightly larger than the final predetermined diameter to compensate for the small amount of negative recoil (step **109**). Temperatures and times suitable for educating the stent to a reduced diameter can be assessed by allowing the stent-mounted balloon catheter of the present assembly to stay at room temperature or at the storage temperature. If the crimped stent stays collapsed at the small diameter corresponding to the deflated balloon under these conditions, the times and temperatures employed during suitable are suitable (step **107**).

[0038] Optimization of the imparted stent mechanical properties such as positive recoil can be achieved by storing the finished product at a room temperature below 20° C. Preferably, the finish product is refrigerated at about 6° to 8° C.

IV. Deployment of the Stent

[0039] The polymer-based stent is first preheating for a period of 3 to 6 min at around 37° C. The preheating of the stent can occur by any means, including heating in saline, the blood stream, or hot air. After the preheating period, the polymer-based stent assembly of the present invention is introduced into a duct, tube, or vessel, e.g. a blood vessel of a mammalian subject, preferably in conjunction with a guiding catheter, and advanced to a target site, e.g. the site of stenotic lesion. After it is located at the target site the balloon is rapidly inflated to initiate expansion of the stent. Alternatively, the stent may be placed on a deployment device that is capable of localized heating of the stent when the stent is correctly

positioned. During this process the diameter of the stent increases, but the thickness of the walls of the stent remain substantially the same.

[0040] It is further contemplated that fracturing of the plaque and deployment of the stent may be done concurrently. If a balloon is used in such cases, the balloon is inflated to a pressure of about 8 to 12 atmospheres to crack the plaque and expand the stent. Alternatively, the vessel may be pre-dilated using angioplasty without the stent. Thereafter, the stent is introduced into the desired site on a separate catheter, preferably an expanding balloon catheter.

[0041] In addition to coronary arteries, the present stent may be used in other arteries such as for example, femeroiliac arteries, the carotid artery, vertebro-basilar arteries, as well as in the interior of other hollow passageways such as for example veins, ureters, urethrae, bronchi, biliary and pancreatic duct systems, the gut, eye ducts, and spermatic and fallopian tubes. Indeed, it is further contemplated that certain aspects of the present invention include devices that are used as substitutes for veins, arteries, and ductal or tubal structures in the body.

[0042] It is to be recognized that aspects of the present invention are applicable to other medical devices. For example, the disclosed formulations can be employed to create a passive marker on an interventional or surgical device, such as a biopsy needle or other hand-held devices. In addition, entire medical devices or portions thereof can embody the imagable material of the present invention.

[0043] While only the presently preferred embodiments have been described in detail, as will be apparent to those skilled in the art, alternatives, additions, modifications and improvements may be made to the device and method disclosed herein without departing from the scope of the invention. Accordingly, it is not intended that the invention be limited, except as by the appended claims.

1. A method of educating a device comprising a cylindrical structure, wherein said device comprises at least one polymer, said method comprising:

deforming said device to a diameter of 1.0 mm to 4.0 mm; and

heating said device to a temperature above the Tg of the polymer from which said device is formed.

2. The method of claim **1**, whereby said device is formed from PLA75 and said device is heated to 80° C. for 30 minutes.

3. A method of fabricating a device comprising a cylindrical structure, wherein said device comprises at least one polymer, said method comprising:

educating said device to a final predetermined diameter size;

crimping at least part of said device to a diameter that is smaller than the final predetermined diameter size, wherein the crimping comprises heating said device to a temperature that is below the glass transition state of said at least one polymer for a time sufficient for said device to temporarily maintain a diameter that is smaller than the final predetermined diameter size.

4. The method of claim **3**, whereby said device comprising a cylindrical structure is crimped onto an inflatable device.

5. The method of claim **3**, whereby the temperature that is below the glass transition state is high enough to allow reduction in diameter of said device comprising a cylindrical structure but not low enough to erase the memory of the final predetermined shape and diameter of said device comprising a cylindrical structure.

6. The method of claim **3**, whereby said temperature that is below the glass transition state is about 50° C.

7. A method of deploying a stent, said method comprising:

fabricating a device comprising a cylindrical structure wherein said device comprises at least one polymer;

educating said device to a final predetermined diameter size and crimping said device to a diameter that is smaller than the final predetermined diameter size, wherein the crimping comprises heating said device to a temperature that is below the glass transition state of said at least one polymer for a time sufficient for said device to temporarily maintain the diameter that is smaller than the final predetermined diameter size;

placing said crimped device within a lumen of the patient's body; and expanding said device.

8. The method of claim **7**, whereby said device expansion is independent of an inflatable device.

9. The method of claim **7**, wherein whereby said device comprising a cylindrical structure is crimped onto an inflatable device.

10. The method of claim **7**, whereby said inflatable device is inflated and/or heated to initiate the expansion of said device comprising a cylindrical structure.

11. The method of claim **7**, whereby the positive recoil properties of said device comprising a cylindrical structure expands said device to its final predetermined diameter.

12. The method of claim **7**, whereby said inflatable device is heated to a temperature below the Tg of the at least one polymer to initiate expansion of said device comprising a cylindrical structure.

13. The method of claim **7**, such that when said device comprising a cylindrical structure is expanded, said device exhibits positive recoil to create outward radial pressure.

14. The method of claim **1**, whereby said device comprising a cylindrical structure is a stent.

15. The method of claim **1**, whereby said at least one polymer has at least one in vivo lifetime.

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