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(19) (CA) **CANADIAN PATENT** (12)

(54) Prosthetic Vascular Graft

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Abstract of the Disclosure

A prosthetic vascular graft is provided. It has walls less than 0.8 millimeters in thickness and is formed of highly expanded polytetrafluoroethylene with a microscopic superstructure of nodes and interconnecting fibrils. The nodes are substantially uniformly separated by an average distance greater than a typical red cell. The superstructure of the graft controls transmural cellular ingrowth and assures the establishment and maintenance of a thin, viable neointima.

The present invention relates to prosthetic vascular structures and, more particularly, to vascular prostheses fabricated from highly expanded polytetrafluoroethylene.

Frequently in cardiovascular surgery, it is necessary to bypass or replace 5 blood vessels, whether veins or arteries, to assure an adequate and balanced blood flow to particular organs, extremities or areas of the body.

Unsuccessful attempts were made during the early years of this century to implant prosthetic or artificial vessels fabricated from glass and metal. With the availability of inert synthetic materials such as, for example, nylon, Orlon (the Trade 10 Mark of du Pont for an acrylic fibre), Dacron (the Trade Mark of du Pont for a polyester fibre made from polyethylene terephthalate) and Teflon (the Trade Mark of du Pont for polytetrafluoroethylene or PTEE) during the late 1940's and early 1950's, large arterial replacements were achieved with increasing degrees of success.

Cardiovascular surgeons presently have available knitted and woven vessels 15 of Dacron and Teflon which may be used as replacements for arteries having relatively large inside diameters (approximately 7 millimeters). However, no clinically acceptable small arterial prosthesis has heretofore been available. Surgeons have found it necessary to scavenge marginally important or superficial vessels, such as, for example, the saphenous vein, to serve as replacements for defective small bore arteries.

20 It is an object of a principal aspect of the present invention to provide a prosthetic vascular structure capable of replacing or bypassing natural blood vessels having relatively small inside diameters as well as those vessels of intermediate and large bore.

The transplantation of saphenous veins from the patient's legs to more critical portions of the cardiovascular system entails numerous disadvantages: The entire surgical procedure is unduly protracted by having first to excise the venous replacement from one part of the patient's body, then to prepare the replacement for implantation, and finally to 5 implant the substitute vessel at another point in the patient's cardiovascular system. Prolonged exposure to anesthesia and multiple incisions combine to increase the probability of both infection and post-operative discomfort.

Cardiovascular surgery frequently requires grafts of various lengths and diameters in achieving, for example: the femoral artery to popliteal artery bypass; the 10 coronary artery bypass; the renal artery bypass, etc. Occasionally, however, especially in older patients, the saphenous veins themselves are inadequate for use as replacements; and in some instances only unacceptably short segments of the saphenous veins are available for transplantation.

It is an object of another aspect of the present invention to provide an artificial 15 vascular structure which may be prefabricated in various lengths and diameters, thereby eliminating unnecessary incisions, minimizing exposure to anesthesia, conserving already limited surgical resources, and assuring an ample supply of small bore vascular replacements.

Operations such as, for example, the femoral/popliteal bypass require 20 especially long grafts which ideally taper in cross-sectional area from their proximal to their distal

ends. Heretofore, transplanted saphenous veins have been used to accomplish this bypass procedure. Because blood flow through the saphenous vein is unidirectional in character, it is necessary to reverse the vein when it is being used as an arterial substitute. The inside diameter of a saphenous vein naturally tapers between its proximal and distal ends. Reversing 5 the vein for implantation between the femoral and popliteal arteries results in a corresponding reversal of this taper so that the smaller diameter end must be grafted to the relatively large femoral artery while the larger diameter end is grafted to the relatively small popliteal artery. The reversed taper of an implanted saphenous vein causes deceleration of the blood flow while the turbulence inducing discontinuities at the bypass junctures contribute to 10 stasis and associated thrombosis.

It is an object of the present invention to provide a prosthetic structure comprising expanded polytetrafluoroethylene having a microscopic structure of spaced nodes interconnected by fibrils, and an average distance between said nodes small enough to prevent trans-mural blood flow but no less than the average 15 dimension of an average red blood cell.

It is an object of a further aspect of the present invention to provide a small bore prosthetic vascular structure which may be fabricated in relatively long segments, which segments decrease in inside diameter between proximal and distal ends so as to facilitate their implantation as peripheral artery replacements and to assure a close hemodynamic 20 simulation of the corresponding natural vessel.

The inner surface of natural blood vessels is characterized by a thin delicate layer of endothelial cells known as the intima. The primary function of this layer is to provide a smooth interface between the blood stream and the vessel wall. For example, a ruptured artery may, after healing, include rough or irregular protrusions from the wall into the blood 25 stream. As the natural intima re-establishes itself over the wound area, it serves to lessen

the severity in irregular wall transitions and thereby to assure laminar blood flow.

While the outer surface of a vessel prosthesis is encapsulated by fibrous growth as a result of normal rejection processes, the inner surface typically becomes isolated from the blood stream by a layer which has been referred to, with varying 5 degrees of accuracy, as the neointima or pseudointima. To be classified as a true neointima, it would be necessary for the inner surface of the artificial vessel to be covered with an extremely thin lining of viable endothelial cells. Although such a lining would vary in thickness from point to point, it would be typically less than ten cells thick. There has heretofore existed no known vessel prosthesis of any size or configuration which, 10 when implanted, would support the growth and maintenance of a true neointima layer. Accordingly, the vessel/blood interface associated with state of the art prostheses is characterized by a pseudointima consisting at best of a few irregularly islands of endothelial growth but made up largely of compacted fibrin which has been flow sculptured by the blood stream. Occasionally, portions of this pseudointima will fracture or particu- 15 late and introduce emboli into the patient's blood stream.

The formation and maintenance of a true neointima requires a continuous extravascular source of nourishment to supplement whatever nourishment might be supplied by diffusion from the adjacent blood stream. In extremely short grafts (less than 2 to 3 centimeters), cellular ingrowth has been observed along the inside surfaces 20 from the suture lines at the ends of the graft. In such cases, the tissue growth, augmented by capillary ingrowth, can provide a continuous nutrition route capable of supporting a viable neointima. However, the thickness of this inner layer of tissue is so great as virtually to occlude all but large bore grafts. Not only is ingrowth of this type unpredictable but it can be expected or tolerated at all only in very short grafts of relatively large inside 25 diameter.

Extensive efforts have been made toward the fabrication of a porous vascular structure which would permit uniform transmural tissue ingrowth sufficient to assure the formation and continuous nutrition of a true neointima layer. The culmination of this prior art effort is represented by grafts which are machine woven from threads consisting of tightly twisted Dacron or Teflon fibers. The threads used in the manufacture of these woven grafts, while extremely small by garment industry standards, are enormous when viewed in the context of a hemodynamic system under pressure. As is true of any knitted or woven fabric, the minimum size of the interstices between threads is determined by the diameter of the threads themselves. Because these interstices or voids are quite large in woven grafts, it is necessary to preclot the graft by dipping it in blood in order to prevent excessive transmural leakage after implantation.

The large size of the threads used in prior art woven grafts and the tightness with which the constituent fibers are twisted renders each individual thread virtually impenetrable to cellular ingrowth and virtually beyond cellular circummigration. From the viewpoint of a single fibroblast, a knitted vessel prosthesis looks like two or three mammoth structures separated by equally mammoth voids which have been filled by the clotting process with masses of coagulated fibrin and proteinaceous matter. Viewed on a similar microscopic basis, the inner wall of the knitted graft appears as a series of large, rough cylinders of inert material separated by cavities which are equal or larger in breadth and depth to the diameter of the cylinders. As the blood flows through a prosthetic vessel of this type, the cavities are filled with slow moving blood while the irregularity and protrusion of the threads promotes turbulent blood flow. Thrombosis throughout the wall of the graft and at its inner surface, combined with the large size and separation of the knitted threads, serves to initiate buildup of an irregular pseudointima while at the same time blocking and inhibiting transmural cellular ingrowth of the type necessary for the support of a uniform, viable neointima.

Accordingly, it is an objective of a major aspect of the present invention to provide a homogeneously porous vascular prosthesis characterized by small nodes interconnected by extremely fine fibrils to form an open superstructure which will allow uniform, controlled transmural cellular ingrowth and thereby assure the establishment 5 and maintenance of a thin, viable neointima as well as firm structural integration of the graft into the body.

It is an object of another aspect of the present invention to provide a porous vascular prosthesis characterized by a superstructure which is substantially impermeable to the flow of relatively high viscosity liquids such as, for example, blood at normal 10 pressures.

Briefly stated, the invention in one of its aspects constitutes a prosthetic vascular device formed from a small bore tube of polytetrafluoroethylene which has been heated, expanded and sintered so as to have a microscopic superstructure

of uniformly distributed nodes interconnected by fibrils and characterized by: (a) an average internodular distance which is (i) large enough to allow transmural migration of typical red cells and fibroblast, and (ii) small enough to inhibit both transmural blood flow at normal pressures and excessive tissue ingrowth; and (b) an average wall thickness which is (i) small enough to provide proper mechanical conformity to adjacent cardiovascular structures, and (ii) large enough, when taken in conjunction with the associated internodular distance, to prevent leakage and excess tissue ingrowth, to allow free and uniform transmural nutrient flow, and to assure mechanical strength and ease of implantation.

By an aspect of this invention, a prosthetic vascular structure of highly expanded polytetrafluoroethylene is provided, having: (i) a macroscopically tubular configuration with proximal and distal ends and (ii) a macroscopic superstructure of irregularly spaced nodes of various sizes and shapes interconnected by fibrils; said vascular structure including: a. an average wall thickness in the range between 0.2 and 0.8 millimeters; b. a substantially uniform distribution of nodes throughout said tubular configuration; c. an average density in the range between 0.2 and 0.5 grams per millimeter; and d. an average distance between said nodes small enough to prevent transmural blood flow and thrombosis but less than the maximum dimension of an average red cell; whereby means are provided for smoothly conveying the flow of blood between at least two points in a living organism while assuring and controlling cellular ingrowth through the wall of the tubular configuration to promote and nourish a thin, viable neointima over the inner surface thereof, and firmly to attach said prosthetic vascular structure to adjacent tissue of said living organism.

By another aspect of this invention, a prosthetic vascular structure of highly expanded polytetrafluoroethylene is provided having: (i) a macroscopically tubular configuration with proximal and distal ends, and (ii) a microscopic superstructure of irregularly spaced nodes of various sizes and shapes interconnected by fibrils; said vascular structure including: a. an average inside diameter in the range between 2 and 6 millimeters; b. an average wall thickness in the range between 0.2 and 0.8 millimeters; c. a substantially uniform distribution of generally ellipsoidal nodes having major axes disposed in a generally

radial orientation with respect to the tubular configuration and an average minor axis dimension less than 18 microns; d. an average density in the range between 0.2 and 0.5 grams per milliliter; and e. a tensile strength in the range of between 2500 and 6500 pounds per square inch; whereby means are provided for smoothly conveying the flow of blood 5 between at least two points in a living organism while assuring and controlling cellular ingrowth through the wall of the tubular configuration to promote and nourish a thin, viable neointima over the inner surface thereof and firmly to attach said prosthetic vascular structure to adjacent tissue of said living organism.

In the accompanying drawings:

10 Figure 1 is a photomicrograph, taken at 250X, of a section through the wall of a vascular graft embodying an aspect of the present invention after having been implanted as a femoral artery segment for a period of eight months;

15 Figure 2 is a photomicrograph, taken by means of a scanning electron microscope at 1000X showing the node and fibril superstructure characterizing highly expanded polytetrafluoroethylene vessel prostheses embodying an aspect of the present invention; and

Figure 3 is a macroscopic stylization of a graft segment showing the sectional plane viewed in Figure 1.

Referring to Figure 1, it is seen that the graft employing the prosthetic vascular structure of an aspect of the present invention consists of a wall 1 having an outer surface 2 20 and an inner surface 3. Corresponding numerical denominations have been applied to the macroscopic view of the graft shown in Figure 3 where the cutaway section shows the plane of wall 1 depicted in the photomicrograph of Figure 1.

The graft as shown generally in Figure 3 is further characterized by a proximal end 4 and a distal end 5. By convention, blood flow through the graft is in the direction 25 indicated by arrow 6, i.e., from the proximal end 4 to the distal end 5.

As can be seen in Figure 1, the inner surface 3 of graft wall 1 is covered by a uniform layer of endothelial cells forming the neointima 7. The outer surface 2 is covered by a uniform, firmly attached, encapsulation 8 of collagenous matter which includes substantial capillary ingrowth (not shown).

5 The graft wall 1 is a superstructure made up of polytetrafluoroethylene nodes 9 which appear in Figure 1 as irregularly shaped islands of inert white material. The polytetrafluoroethylene nodes 9 are interconnected by fine fibrils which, because of their extremely small diameters

and their disaffinity for photographic stain, are not visible in Figure 1.

Figure 2 depicts the node/fibril superstructure of Figure 1 prior to implantation. The polytetrafluoroethylene nodes 9 and the many interconnecting fibrils 10 may be clearly observed in the absence of cellular ingrowth and at a magnification of approximately 4 times greater than Figure 1. On the other hand, the implanted graft of Figure 1 has been thoroughly and uniformly permeated by fibroblasts having cellular nuclei which appear in the photomicrograph as black dots 11. The nuclei of the endothelial cells forming the neointima 7 appear as black dots 11a in Figure 1. As may be observed, the neointima is approximately eight cells in thickness.

Methods and techniques for expanding polytetrafluoroethylene have been known for many years. One of the earliest disclosures containing a discussion of such procedures and resultant products is found in Japanese Patent No. 13,560/67 which was filed November 1, 1963 and officially published on August 1, 1967. Partially expanded polytetrafluoroethylene has been used to provide electrical insulation on cables and to form low friction structural members such as, for example, bearings and piston rings, to provide sealing in liquid systems, and, without clinically meaningful or reproducible success, for artificial vessel replacements. See for example, Japanese Patent No. 13,560/67; Soyer, et al, "A New Venous Prosthesis", Surgery, Vol. 72, page 864 (December 1972); and Matsumoto, et al, "A New Vascular Prosthesis for Small Caliber Artery", Surgery, Vol. 74, page 519 (October 1973).

The basic process for expanding polytetrafluoroethylene is quite simple: The constituent resin is first subjected to shear by, for example, extrusion into the desired geometrical configuration. The extrudate is then heated at a temperature below the sintering temperature of 327°C and physically stretched or expanded along at least one axis. The expanded member is then physically restrained against contraction and is sintered by brief exposure to temperature in excess of 327°C, thereby crystallizing the

expanded structure and providing moderate tensile strength of up to approximately 6500 psi. As the raw extrudate is stretched, the non-porous polytetrafluoroethylene separates into solid nodes 9 of polytetrafluoroethylene which remain structurally interconnected by polytetrafluoroethylene fibrils 10 which are drawn from the nodes 5 during expansion. (See Figure 2.) Node size and distribution in the final product is adversely affected by: very rapid expansion, uneven expansion, insufficient heating, non-uniform heating, and irregularly distributed expansion forces. The distance between nodes 9 is directly proportional to the extent to which the extrudate has been expanded. When polytetrafluoroethylene is properly expanded along one axis, virtually no 10 dimensional changes are observed in the orthogonal direction. The directional vector 12 in Figures 1, 2 and 3, indicates the axis along which grafts embodying the prosthetic vascular structure of aspects of the present invention are typically expanded.

As can be seen in Figures 1 and 2, the nodes 9 are crudely ellipsoidal in configuration with their major axis disposed at approximately right angles to the axis of 15 expansion 12. The nodes 9 are of random but generally uniform size and are distributed in a homogenous pattern throughout the wall 1. Furthermore, the nodes are extremely small, typically less than a few times the size of a normal fibroblast or red cell.

Because the minor axes of the ellipsoidal nodes 9 are transverse to the generally radial direction typifying cellular ingrowth, the invading fibroblasts never encounter 20 any massive polytetrafluoroethylene barrier. It is believed that the size and orientation of the nodes 9, together with their tapered configuration, facilitates cellular migration and ingrowth. To avoid substantial impediment to ingrowth, the average minor axis dimension of the nodes 9 is less than three times the major dimension of a typical red cell, i.e., less than 18 microns.

25 It has been found that the average internodular distance, as measured along the axis of expansion 12, must fall within a relatively narrow range of values, viz.,

between approximately 6 and 80 microns. As will be appreciated by those skilled in the art, the term "average" when used in conjunction with internodular distance and node size cannot be used or interpreted with statistical precision; rather, the term is intended to connote a nominal or typical dimension derived from a broad sample. By way of 5 example, where the average internodular distance is said to be 30 microns, it would be expected that some of the nodes would be separated by only a few microns while others might be separated by 90 or 100 microns. In the ideal graft, each node 9 would have a perfect elongated football shape and would be separated from its neighbours by uniformly distributed fibrils 10 of equal lengths. Unfortunately, such perfection is rarely, 10 if ever, achieved in a microscopic environment.

When the average internodular distance is less than the major dimension of a typical red cell, or approximately 6 microns, inadequate cellular ingrowth has been observed. In such cases, the node/fibril superstructure is so tightly packed as to preclude either the establishment or continued nutrition of a viable neointima.

15 Associated with very large internodular distances is a loss of tensile strength and overall structural integrity. The graft becomes progressively more pliable and progressively more difficult to handle during surgery. Excessively expanded grafts will be subject to deformation and leakage at the suture line. Furthermore, excessive cellular ingrowth has been observed in grafts having an average internodal distance in excess of 20 approximately 80 microns. Where the inside diameter of the graft is critically small, excessive cellular penetration of this type can lead to the formation of a pseudointima or an unacceptable thickening of the neointima with an accompanying occlusion of the lumen.

As the average internodular distance is extended beyond, for example, 150 to 25 200 microns, the graft superstructure becomes progressively more permeable to blood flow and is characterized by substantial interstitial clotting and progressively decreasing

and non-uniform cellular ingrowth. Ultimately, were it possible to reproduce internodular distances comparable in size to the interstitial voids of the size characterizing woven grafts, then virtually all transmural growth would be inhibited, and the support of a true neointima would be impossible.

5 Just as the nodes 9 must be of such size, configuration and orientation to avoid substantial impediment to cellular ingrowth, so too, must they be substantially uniformly distributed throughout the length and cross-section of the graft. It has been found that 10 clumps or groupings of closely packed nodes can serve as barriers to ingrowth. It has also been found particularly critical to the creation and support of a viable neointima, that the nodes immediately adjacent to and defining the inner surface 3 of the graft be uniformly distributed, i.e., neither clumped so as to block the flow of nutrition to the neointima nor so widely spaced as to define deep thrombosis inducing irregularities in the inner wall 3.

15 A phenomenon referred to as "skin effect" is attributed to non-uniform force distribution occurring either during the extrusion or the expansion of tubular structures of the type embodying the prosthetic vascular structure of aspects of the present invention. Skin effect involves a relatively high concentrate of nodes in a given circumferential plane, usually either the inner surface 3 or the outer surface 2. A slight skin effect is observed on the outer surface 2 of the graft segment shown in Figure 1, while the inner surface 3 is characterized by an open regular node pattern. The limited skin effect 20 observed at outer surface 2 is obviously acceptable in degree since, as Figure 1 clearly shows, it has not adversely affected cellular ingrowth.

25 In some instances, skin effect is a highly localized phenomenon, yet in other instances it may be observed over the entire inner or outer surface of the graft. A skin effect which might be acceptable at the outer surface of the graft might well be unacceptable at the inner surface because of the greater criticality for establishing a regular super-



structure which will allow ample and uniform nutrient flow to, and mechanical support for, the neointima 7.

5 Wall thickness is another factor affecting the establishment and maintenance of a viable neointima in grafts embodying the prosthetic vascular structure of aspects of the present invention. For any particular internodular distance within the acceptable range, the thickness of the graft wall can be made so great as to preclude complete transmural cellular migration and ingrowth.

10 Nutrition of the fibroblasts in wall 1 as well as the endothelial cells forming the neointima 7 depends primarily upon capillary ingrowth, which normally becomes well developed within the third and fourth weeks after implantation of the graft. Diffusion of nutrients from the blood stream itself, especially during the period immediately following implantation, and transmural diffusion of nutrition also contribute to tissue growth.

15 There is a limit to which cellular ingrowth will penetrate from the outer surface 2 into a graft wall. The depth or extent of nutritional diffusion is also limited. Where the wall of a particular graft is too thick (considering the internodular distance), an unnourished calcifying layer will form within the wall of the graft in any area which is so far removed from the outer surface 2 as to be beyond the range of normal cellular ingrowth and which is too far from the inner surface 3 to receive nutrients by diffusion 20 from the blood stream. This calcifying layer, itself a result of inadequate nutritional flow, acts as a barrier, further inhibiting the growth of new cells. Within the acceptable range of average internodular distances, it has been found that wall thicknesses greater than approximately 0.8 millimeters are excessive.

25 While excessive wall thickness may result in an unacceptably stiff graft or one which inhibits transmural ingrowth, insufficient wall thickness results in a limp, unman-

ageable graft or one which allows excessive cellular ingrowth. Thus, wall thickness and internodular distance are the two prime determinants of graft pliability and strength and are also the two major factors which combine to control the extent and uniformity of transmural ingrowth.

5 Grafts embodying the prosthetic vascular structure of an aspect of the present invention, having wall thicknesses in the range between 0.2 and 0.8 millimeters and characterized by internodular distances in the range between 6 and 80 microns have exhibited excellent mechanical properties and have resulted in controlled ingrowth sufficient to assure the support of a thin, viable neointima which does not unduly restrict the 10 lumen of the graft. Grafts falling outside these ranges have been found to be marginal or clinically unacceptable. Grafts falling into these preferred ranges are characterized by moderately high tensile strength in the range between 2500 and 6500 pounds per square inch and average density between 0.2 and 0.5 grams per milliliter.

15 In the introductory portion hereof it was pointed out that reversed saphenous vein grafts have been frequently used to bypass diseased segments of peripheral arteries. This requires anastomosis of the graft to arteries having substantially different diameters. The proximal artery always has a larger diameter than the distal artery, as is true, for example, in the case of the common femoral artery to-popliteal artery bypass performed for the purpose of routing blood flow around obstruction in the superficial femoral artery. 20 The femoral artery normally has a diameter of 6 to 8 millimeters whereas the popliteal artery is typically between 3 and 4 millimeters in diameter. If a constant diameter prosthesis or a reversed saphenous vein is used in accomplishing the bypass, a sudden cross-sectional area change is present at either or both ends of the graft. From a hemodynamic point of view, these sudden area changes are highly disadvantageous in that they 25 tend to produce turbulence with pockets of stasis, causing deposition of platelets, fibrin and cellular materials and thus the formation of thrombi which can ultimately propagate



to occlusion of the vessel. These problems are greatly minimized and even substantially eliminated in accordance with one embodiment of the present invention wherein the inside diameter of the prosthesis tapers gradually over its entire length from the diameter of the larger proximal vessel down to the diameter of the smaller distal vessel. Tapering of the graft promotes streamlining of the blood and assures laminar rather than turbulent flow. Tapering also results in accelerated blood flow with an associated increase in the velocity of the blood elements; as opposed to deceleration, which characterizes flow through the reversed saphenous vein. Thus, tapered grafts of prosthetic vascular structures of aspects of the present invention not only provide for the growth and maintenance of a viable neointima but also promote accelerated laminar blood flow with the resultant elimination of clotting due to stasis and turbulence.

In addition to the significant biomedical, hemodynamic and mechanical advantages which accompany the use of vessel prostheses of the type described herein, numerous surgical, procedural and post-operative benefits are also realized. Cardio-vascular surgery calling for a small bore vessel substitute has heretofore required initial harvesting of the saphenous vein for use as a graft. This involves the following surgical steps which are rendered unnecessary by the use of the prosthetic vascular structure of aspects of the present invention:

1. groin incision and proximal dissection of the patient's saphenous vein;
2. lower thigh incision and distal dissection of the patient's saphenous vein;
3. multiple medial thigh incisions for completion of the saphenous vein dissection;
4. removal of the saphenous vein and careful ligation of all branches;

5. reversal of the saphenous vein prior to implantation so that venous valves do not obstruct arterial flow, thereby necessitating the awkward anastomosis of the small (approximately 3 millimeters) distal end of the vein to a substantially larger proximal artery and, in the case of a peripheral artery replacement, anastomosis of the 5 relatively large proximal end of the vein to a very small bore distal artery;

6. careful inspection for possible leaks from the various branches in the graft or from those remaining in the patient's thigh; and

7. closing of the multiple incisions necessitated by removal of the saphenous vein.

10 Use of expanded polytetrafluoroethylene grafts of the type described and claimed herein not only eliminates these unnecessary surgical procedures, but also shortens operative and anesthetic time by between one and two hours, does away with the post-operative discomfort and possible infection resulting from numerous leg wounds and greatly conserves already limited hospital and surgical resources.

15 The manufacture of prosthetic vascular structures embodying aspects of the present invention is extremely simple and can be readily performed with the most rudimentary laboratory equipment, realizing of course that more sophisticated equipment is required for volume production and quality control. The basic physical, chemical and procedural parameters for expanding polytetrafluoroethylene are presented and discussed 20 in the previously cited Japanese Patent No. 13,560/67; however, an example will be given illustrating the fundamental technique involved in making small bore polytetrafluoroethylene grafts having the claimed structure.

25 Polytetrafluoroethylene is extruded to form tubing having an average inside diameter of approximately 4 millimeters and an average wall thickness of approximately 0.5 millimeters. Unsintered tubing of this type, identified by the manufacturer's



No. S16882-7, may be obtained from W. S. Shamban Company (71 Mitchell Road, Newberry Park, California 91320). The unsintered extrudate, which is quite fragile, is carefully cut with a razor blade into lengths of, for example, 7.3 centimeters. Small aluminum plugs of virtually any configuration are inserted into the end of the tubing and 5 secured thereto by tightly wrapped stainless steel wire. A relatively short end segment is thus confined between the inserted plug and the stainless steel wire. These plugs provide points for handling and attachment during the subsequent heating, elongation and sintering steps.

The tubing and plug assembly is placed in a uniformly heated oven for 10 approximately one hour at 275°C. Thereafter, the assembly is removed from the oven, the plugs are grasped and stretched apart manually to obtain a tubular length of 23 centimeters. The time required for removal and elongation should be made as short as possible to reduce the effects of cooling. Elongation should be carried out at a moderate, uniform rate and the plugs should be moved apart along a common axis of expansion to 15 assure uniform force distribution. Typically, this manual operation has required less than ten seconds and has yielded good results.

The elongated assembly is then secured against contraction by restraining the plugs at the desired separation. This may be achieved in any number of obvious ways, as for example, by using plugs with enlarged ends which are placed in a fixture having 20 U-shaped slots separated by the desired distance of 23 centimeters.

While still restrained, the elongated assembly is returned to the oven for approximately forty-five seconds at 400°C. during which time the node/fibril superstructure is sintered and becomes fixed. The elongated grafts are then cut to the desired lengths and after sterilization are ready for implantation.

In large commercial applications, expansion is achieved mechanically in the oven itself at closely controlled rates and is immediately followed by sintering. However, excellent grafts, such as, for example, the one shown in Figure 1, have been obtained by the simple laboratory techniques outlined above.

5 Fabrication of tapered grafts such as, for example, those used for peripheral artery replacement involves the additional step of reshaping a sintered tube of desired length and diameter over a tapered stainless steel mandrel which has been heated to approximately 300°C. After the entire assembly is allowed to cool, the slightly re-expanded graft retains the shape of the mandrel and may be removed for use without
10 further heat treatment.

It will be apparent to those skilled in the art that the disclosed prosthetic vascular structure may be modified in numerous ways and may assume many embodiments and configurations other than those specifically set forth and described above. For example, the basic prosthetic structure may be made various lengths and in diameters up
15 to approximately 40 millimeters without affecting the structural integrity or operativeness of the graft. Various secondary configurations such as bifurcated grafts may also be produced. Because grafts embodying the prosthetic vascular structure of aspects of the present invention are substantially impermeable to transmural blood flow, it will be obvious to those of ordinary skill in the art that the patients need not be heparinized to
20 avoid leakage through the graft.

X

CLAIMS:

1. A prosthetic structure comprising expanded polytetrafluoroethylene including a structure comprising spaced nodes interconnected by fibrils, said structure including an average distance between nodes greater than or equal to approximately 6 microns but no greater than approximately 200 microns.
2. The prosthetic structure as defined in Claim 1 wherein the structure is a tube with a proximal end and a distal end.
3. The prosthetic structure of Claim 1 wherein said nodes are generally ellipsoidal in shape and have an average dimension along their minor axes less than three times the maximum dimension of an average red cell.
4. The prosthetic structure of Claim 2 wherein said nodes are generally ellipsoidal in shape and have an average dimension along their minor axes less than three times the maximum dimension of an average red cell.
5. The prosthetic structure of Claim 4 wherein the major axes of said nodes are in a generally perpendicular orientation with respect to the axis of the tube.
6. The prosthetic structure of claim 2, 4, or 5 wherein the tube has an average wall thickness of between 0.2 and 0.8 millimeters.
7. The prosthetic structure of any one of claims 1 to 6 wherein the structure has an average density of between 0.2 and 0.5 grams per cubic millimeter.

8. The prosthetic structure of any one of claims 1 to 7 wherein a substantially uniform distribution of nodes exists throughout the structure.

9. The prosthetic structure of any one of claims 1 to 7 wherein the structure is made up of irregularly spaced nodes.

10. The prosthetic structure of claim 2, 4, 5 or 6 wherein the tubular configuration thereof has an average inside diameter of less than 40 millimeters.

11. The prosthetic structure of claim 10 wherein the tubular configuration has an average inside diameter of less than approximately 8 millimeters.

12. The prosthetic structure of claim 11 wherein the tubular configuration thereof has an average inside diameter of between 2 and 6 millimeters.

13. The prosthetic structure of any one of claims 1 to 12 further having a tensile strength in the range between 2500 and 6500 psi.

14. The prosthetic structure of claim 2, 4, 5, 6, 10 or 11 wherein the tubular configuration thereof tapers from a first inside diameter at the proximal end to a second inside diameter at the distal end.

15. The prosthetic structure of claim 14 wherein said first inside diameter is in the range between 5 and 8 millimeters and the second inside diameter is in the range of between 2 and 6 millimeters.

16. The prosthetic structure of any one of claims 1 to 15 wherein the average distance between the nodes is between approximately 6 and 80 microns.

17. A prosthetic device comprising expanded, porous polytetrafluoroethylene comprising a microstructure consisting of nodes interconnected by fibrils, wherein substantially all said fibrils are between 6 and 200 microns in length thereby permitting tissue ingrowth.

18. An artificial vascular prosthesis comprising a tube of expanded, porous polytetrafluoroethylene possessing a matrix tensile strength in at least one direction exceeding 7300 psi and possessing a microstructure consisting of nodes interconnected by fibrils, wherein said fibrils are from 6 to 200 microns in length thereby permitting tissue ingrowth.

19. The prosthesis in accordance with claim 17 or 18, wherein said fibrils are about 20 to 100 microns in length.

20. The prosthesis in accordance with claim 17 or 18, wherein the average distance between the nodes is between approximately 6 to 80 microns.

21. A prosthetic vascular structure of expanded polytetrafluoroethylene having:

- (i) proximal and distal ends, and
- (ii) a microscopic structure of irregularly spaced nodes of various sizes and shapes interconnected by fibrils; said vascular structure including:
 - (a) an average wall thickness in the range between 0.2 and 0.8 millimeters;
 - (b) a substantially uniform distribution of nodes throughout said vascular structure;

(c) an average density in the range between 0.2 and 0.5 grams per cubic millimeter; and

(d) an average distance between said nodes of between approximately 6 and 80 microns;

whereby means are provided for smoothly conveying the flow of blood between at least two points in a living organism while assuring and controlling cellular ingrowth through the wall of the vascular structure to promote and nourish a thin, viable neointima over the inner surface thereof, and firmly to attach said prosthetic vascular structure to adjacent tissue of said living organism.

22. The prosthesis in any of claims 17 to 21 wherein the inside diameter of said tube being less than 8 millimeters.

23. The prosthetic structure of claim 22 having an average inside diameter of between 3 and 6 millimeters.

24. The prosthetic structure of any one of claims 17 to 23 further having a tensile strength in the range between 2500 and 6500 psi.

25. The Prosthetic structure defined in any one of claims 1, 3, 8, 13, 16 or 17 wherein the structure includes a wall having a thickness of between 0.2 and 0.8 millimeters.

26. The prosthetic structure of claim 18, 19, or 20 wherein the tube has an average wall thickness of between 0.2 and 0.8 millimeters.

27. The prosthetic structure of any one of claims 1-17 or 25 wherein the structure is a vascular structure.

13 4 15 19

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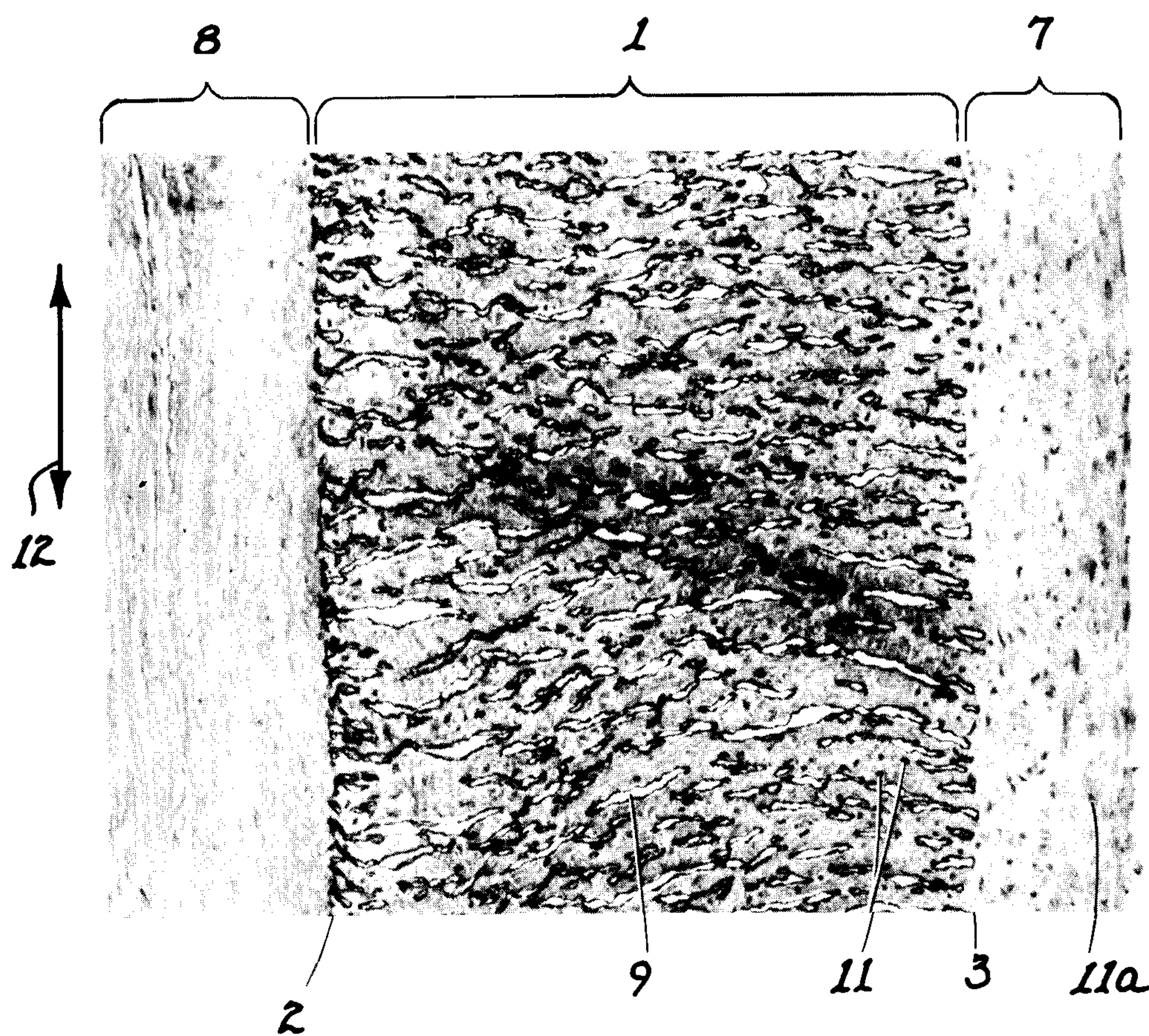


fig. 1

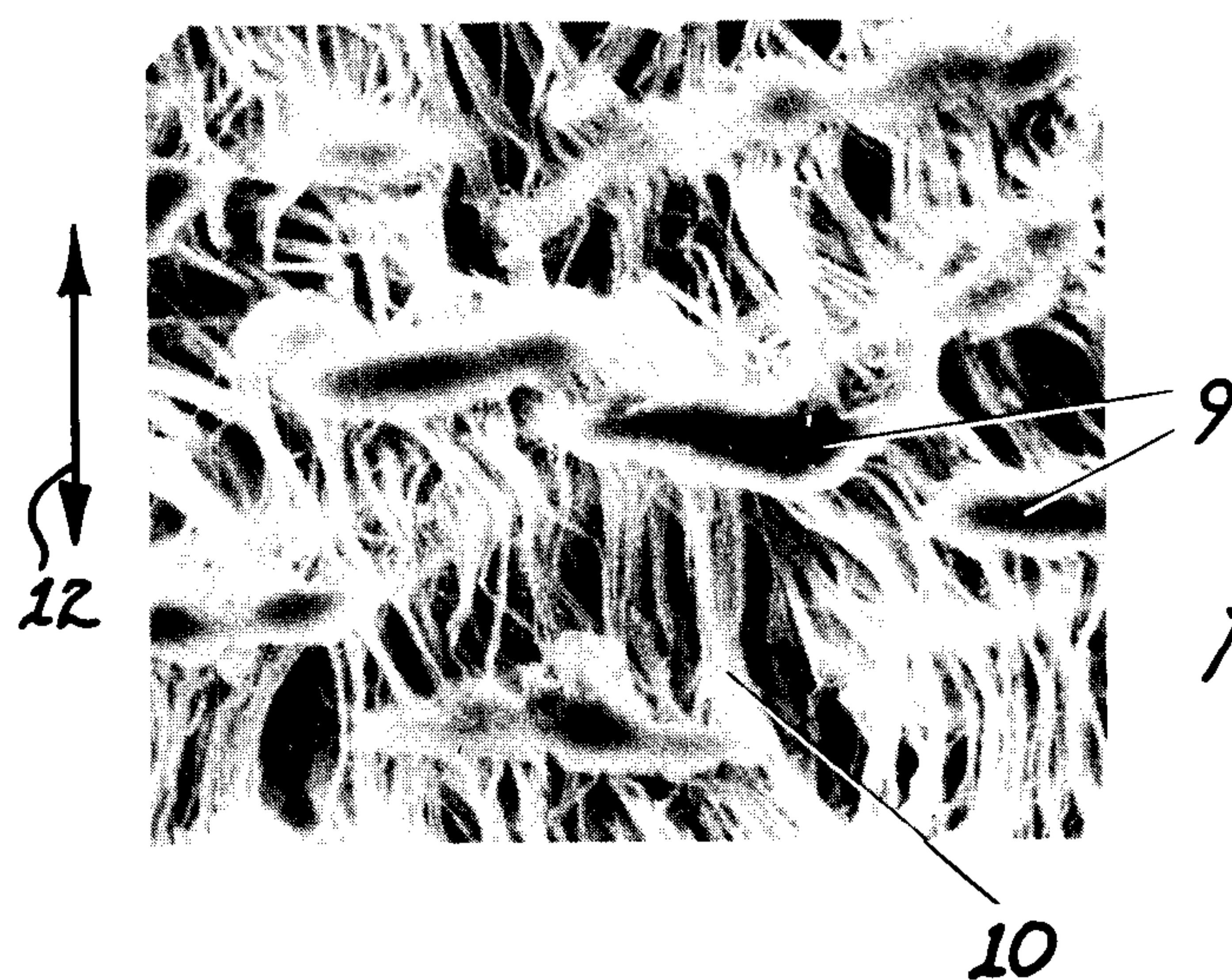


fig. 2

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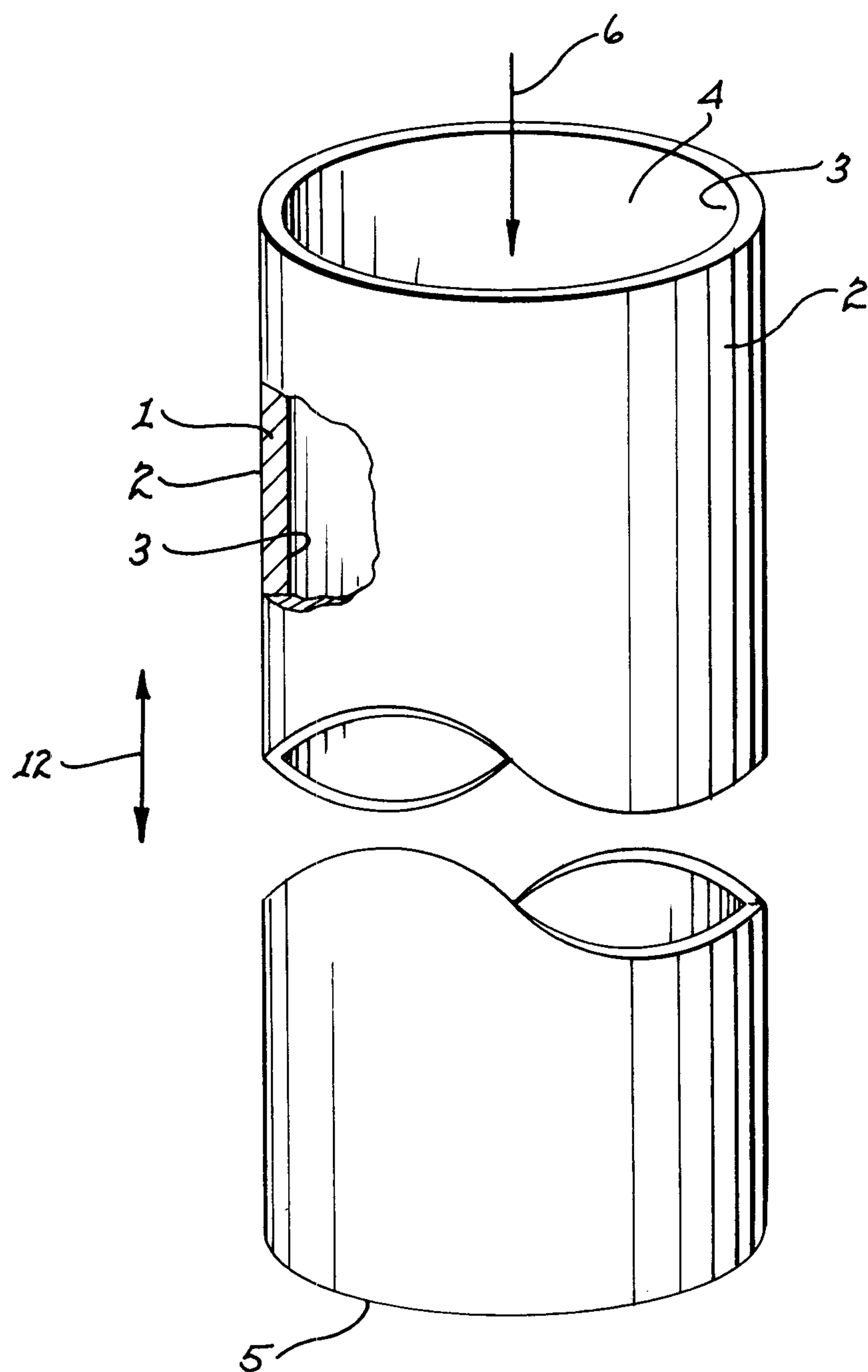


Fig. 3

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