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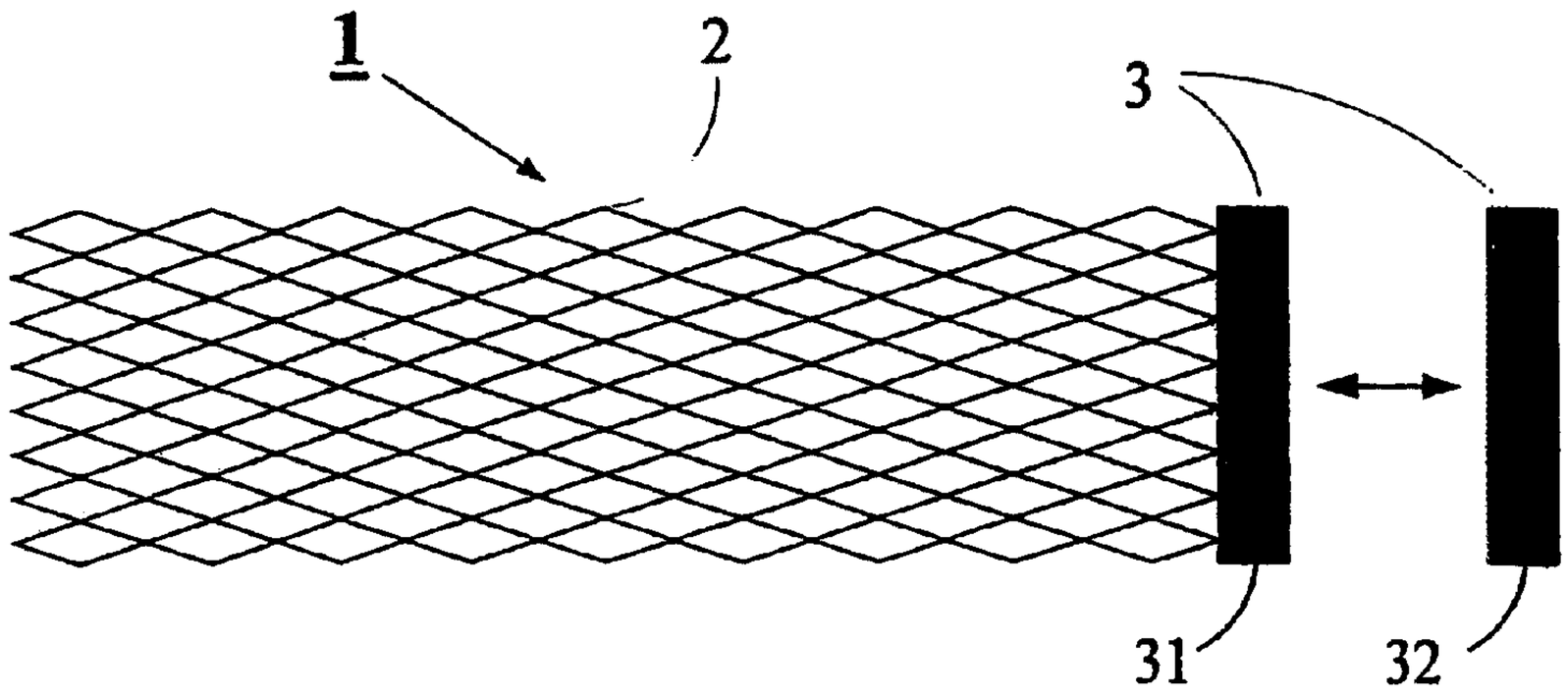
(54) **EXTENSEUR ET PROCEDE D'IMAGERIE PAR RESONANCE MAGNETIQUE POUR REPRESENTER ET DETERMINER LA POSITION D'UN EXTENSEUR**

(54) **STENT AND MR IMAGING METHOD FOR REPRESENTING AND DETERMINING THE POSITION OF A STENT**



(57) L'invention concerne un procédé de résonance magnétique, servant à représenter et à déterminer la position d'un extenseur introduit dans un objet à examiner, ainsi qu'un extenseur. Selon l'invention, l'extenseur (1) comporte au moins un circuit oscillant (4) passif ayant une inductance (2) et une capacitance (3) et dont la fréquence de résonance correspond sensiblement à celle du rayonnement haute fréquence injecté provenant du système à résonance magnétique. On obtient ainsi dans une zone limitée localement, qui est située à l'intérieur ou autour de l'extenseur, une réponse de signal modifiée qui est représentée à résolution spatiale.

(57) The invention relates to a magnetic resonance imaging process for representing and determining the position of a stent inserted in an examination object, and to a stent. In accordance with the invention, the stent (1) has at least one passive oscillating circuit (4) with an inductor (2) and a capacitor (3). The resonance frequency of this circuit substantially corresponds to the resonance frequency of the injected high-frequency radiation from the MR system. In this way, in a locally limited area situated inside or around the stent, a modified signal answer is generated which is represented with spatial resolution.

Abstract

The invention relates to a magnetic resonance imaging process for representing and determining the position of a stent inserted in an examination object, and to a stent. In accordance with the invention, the stent has at least one passive oscillating circuit with an inductor and a capacitor. The resonance frequency of this circuit substantially corresponds to the resonance frequency of the injected high-frequency radiation from the MR system. In this way, in a locally limited area situated inside or around the stent, a modified signal answer is generated which is represented with spatial resolution.

Stent and MR Imaging Method for Representing and Determining the Position of a Stent

Description

The invention relates to an MR (magnet resonance) imaging method representing and determining the position of a stent according to the generic part of Claim 1 and a stent according to the generic part of Claim 9.

Background of the Invention

MR imaging methods have been known for some time. They are based on the resonance alternating effect between a high-frequency electromagnetic alternating field and specific atomic nuclei of an object to be examined, in particular a human or an animal body that is arranged in a strong external magnetic field. The atomic nuclei precess in the magnetic field (B_o) by the so-called Lamor frequency that is proportional to the strength of the magnetic field. When applying an electromagnetic alternating field whose magnetic alternating component (B_1) is vertical to the direction of the strong magnetic field (B_o), the spins of the atomic nuclei flip and associated relaxation times may thus be measured.

In the description of a scientific model the magnetization of the individual spins is described by total magnetization. This total magnetization in its equilibrium condition is parallel to the external magnetic field and called equilibrium magnetization. By means of an HF-impulse

applied with the Lamor frequency (resonance frequency), the magnetization may be deflected by an angle α with regard to the direction of the magnetic field. The angle α is proportional to the time period of the HF-impulse applied and the strength of the magnetic field (B_1) of the HF-impulse. Subsequent to an excitation by the angle α, the total magnetization precesses around the direction of the magnetic field. The precessing magnetization may be recorded by a coil that is oriented vertically to the direction of the magnetic field in form of a voltage signal. The strength of the voltage signal is proportional to $\sin(\alpha)$, proportional to the density of the spins in the signal emitting volume and inversely proportional to the temperature.

The maximal signal response of a given volume is thus attained after 90° excitation. The recorded signal amplitude decreases exponentially with the relaxation time T_2^*, since the individual spins fall out of phase due to the fluctuating magnetic fields. Simultaneously, the total magnetization increases exponentially again in the direction of the magnetic field towards the equilibrium magnetization with relaxation time T_1. By means of magnetic gradient fields switched at the correct point in time, it is possible to image differentiated combinations from the spin density and the two relaxation times in a gray scale encoded image with spatial resolution.

It is further known to locally induce an amplification of the excitation of the nuclear spins by means of a resonance circuit. For this, so called "fiducial markers" are known that have compartments filled with special signal-intensive liquids surrounded by a resonance circuit. (Burl et al.: "Tuned Fiducial Markers To Identify Body Locations with Minimal Perturbation of Tissue Magnetization", in: Journal of Magnetic Resonance in Medicine 1996, p. 461 - 493.) The resonance circuit has the resonance frequency of the MR system.

If such a fiducial marker is brought into the imaging volume of a nuclear magnetic resonance tomograph, the resonance circuit is excited when electromagnetic radiation is applied at resonance frequency. This results in amplification of the magnetic alternating field within the inductance of the resonance circuit. The increased magnetic component of the magnetic field increases the deflection angle α of the protons within the inductance. With a small angle of excitation (α <90°) of the protons by the nuclear spin system, the protons experience an increased excitation angle within the inductance. In the ideal case, protons are excited with a small angle of 1° to 10° in the imaging volume, whereas the protons within the inductance are excited with 90°. Even with identical relaxation times and with an identical spin density, the signal from the compartment surrounding the resonance circuit is clearly more intensive than the signal of the other parts of the image. Since this signal amplification is localized, it may be used to determine positions.

According to the law of reciprocity, it is also true that the MR response signals of the protons within the compartment surrounding the resonance circuit (fiducial markers) are amplified. Due to the inductance, the magnetic field lines originating from the spins within the coil are bundled such that more signal is emitted from the volume within the inductance and applied to a associated receptor coil. This amplification of emitted and then received signals is considered independent of an increased excitation. Both effects result in a changed signal response of the fiducial marker.

Disadvantageously, fiducial markers make use of separate signal emitting volumina, which for visibility in the MR image must be at least a few cubic millimeters in size and must be

placed specifically in the examination object or must be integrated into the systems that are placed in the examination object. Often this is not possible.

With the introduction of open magnets and new techniques with closed MR systems, it has become possible to carry out interventional and minimally invasive techniques such as punction, catherization and surgical processes under MR tomographic control. However, ferromagnetic or paramagnetic metals or impurities in other materials result in artefacts in the images.

Problems result from the tools used for interventional and minimally invasive techniques since they usually consist of ferromagnetic or paramagnetic material and/or that they are so small that they are about the size of one pixel (ca. 1 mm) in MR images. In particular, stents made of metal or plastic are barely visible in the MR image due to their fine skeletons and can best be located by means of artefacts. When materials that are not visible in the MR image are used, they can be seen only as "shadows". These disadvantages result in the fact that MR monitoring is frequently unsatisfactory and that an x-ray method with all its known disadvantages is used instead for imaging.

From DE 195 10 194 A1 an active-invasive magnet resonance system for the production of selective MR angiograms is known, whereby an invasive apparatus is provided with an HF coil by which the nuclear spin magnetization of the blood flowing in the vessel is changed locally. By means of special MR image impulse sequences, only the blood that has a changed nuclear spin magnetization is selectively detected and imaged.

US patent 5,445,151 describes a method for flow measurements in flowing fluids, in particular in blood, whereby the invasive apparatus is provided with at least two HF coils, whereby a local change in nuclear spin magnetization produced by one HF coil is sensed at the other HF coil and the delay interval is used for the computation of flow velocity.

The two publications cited do not refer to the imaging of medical apparatuses introduced into a body. Furthermore, they have the disadvantage that they are active systems whereby the apparatuses introduced are permanently connected via cable connections to extracorporeal components.

Patent publication DE 195 07 617 A1 describes an MR method whereby a surgical instrument, such as a catheter, is introduced into an examination object whereby the catheter is provided with a micro-coil at its point. The position of the micro-coil is determined by specific sequential techniques.

EP-A-0 768 539 discloses an MR method for determining the position of an object which has been introduced into the body of a patient. A coil arrangement without connection to extracorporeal components is attached on the object to be introduced into the body, for instance, a catheter or a surgical instrument, and a signal change which occurs due to the coil is used to determine the location of the object.

Objects of the Invention

The object of the present invention is to provide an MR imaging method for representing and determining the position of a stent introduced into an examination object and to provide a stent which allows for clear, signal-intensive imaging of a stent in MR images as well as for improved

fluid flow measurements.

Summary of the Invention

To accomplish this object, the invention is characterized by an MR imaging method with the characteristics of Claim 1 and a stent with the characteristics of Claim 9. Advantageous and preferred embodiments of the invention are reported in the dependent claims.

To accomplish this object, provision is made in the invention to integrate a resonance circuit into the skeleton of the stent to be introduced into the examination object or to form a resonance circuit from the skeleton which induces a changed response signal in a locally defined area in or around the stent that is imaged by spatial resolution. The resonance frequency of the resonance circuit is essentially equal to the resonance frequency of the applied high-frequency radiation of the MR imaging system. Since that area is immediately adjacent to the stent from inside or outside, the position of the stent is clearly recognizable in the correspondingly enhanced area in the MR image. Because a changed signal response of the examined object is induced by itself, only those artefacts can appear that are produced by the material of the stent itself.

Due to the clear imaging of the stent in the MR image a precise position determination is possible. Furthermore, based on the changed signal conditions, improved flow measurement of the medium flowing through the stent or along the stent is now possible. Use is made of the fact

that different excitation is present inside and outside the stent.

The object of the present invention is accomplished as it is based on the surprising discovery that suitable resonance circuits can be provided on a stent itself. The invention provides that the inductance and capacitance providing the resonance circuit are formed by the material of the stent, thereby resulting in a synergistic effect. Alternatively, inductance and capacitance are integrated into the stent as separate components, whereby the inductance is formed by a separate coil, which is interwoven into the unfoldable skeleton of the stent such that it unfolds along with the skeleton at the time of the unfolding of the stent.

According to the invention, the signal response of the spins within the inductance is changed. Two processes contribute to this. On the one hand, the resonance circuit tuned to the resonance frequency is excited by the application of high-frequency radiation and the nuclear spins detected by the field of the resonance circuit experience amplified excitation through local amplification of the alternating magnetic field in or near the inductance. In other words, protons detected by the field lines of the induced magnetic field are deflected at a larger angle than the protons on the outside of this induced magnetic field. An increased flip of the nuclear spins results. Accordingly, the signal response sensed by a receptor coil and evaluated for imaging can be amplified. It is furthermore possible that only the spins within the inductance experience saturation and that the signal is diminished with regard to the environment. In both cases, a change in signal response is apparent.

On the other hand - independent of amplified excitation - the MR response signals of the protons within the inductance are amplified. The inductance thus bundles the magnetic field lines originating from the spins within the inductance, which results in an amplified signal emission and an application to an associated receptor coil that receives the amplified signals and transmits them for MR imaging. This effect is described in the publication by J. Tanttu: "Floating Surface Coils", in: XIV ICMBE and VII ICMP, Espoo, Finland 1985.

According to the present invention, both of these effects may be used in the process of changing the signal response. However, the second effect, i.e., an amplification of the MR response signal, may also be used alone.

Accordingly, a first embodiment of the present invention is characterized in that the application of high-frequency radiation excites the resonance circuit, thus resulting in an amplified excitation of the nuclear spins in the locally defined area. Preferably, the locally defined area in which such an amplified excitation of the nuclear spins take places is located within the stent. This is obviously the case if the skeleton of the stent forms the inductance.

A second embodiment of the invention, however, provides, that with the application of the high-frequency radiation the resonance circuit becomes detuned or that the capacitance is short circuited such that no enhanced excitation of the nuclear spins takes place in the locally defined area. However, during measurements of the signal response of the locally defined area, the detuning of the resonance circuit or the short circuiting of the capacitance is canceled again, thus causing the resonance circuit to provide an amplification of the radiated MR response signals of the protons. It was in particular found that this variant makes possible the imaging of the area in and around the stent with high quality, i.e., that it provides local imaging beyond the pure position determination. In addition to the position of the stent, the MR image provides improved information regarding the structure, etc. of the inside and/or the environment of the stent.

An amplification of the excitation of the nuclear spins is, for example, suppressed, in that the condenser of the resonance circuit is short circuited during excitation by means of crossed diodes. The amplification of the emitted signals is thus not influenced, since the small induced voltage from the spins within the inductance is below the conducting-state voltage during emission.

General reference is made to the fact that the change of the signal response according to the invention will usually be an amplification of the signal response. However, this depends on numerous factors, in particular on the excitation sequences used. For instance, with quick consecutive sequences it is possible that a saturation of the excitation of the spins within the inductance is present, thus no signal is produced there. There is, however, no saturation present in the area outside of the inductance, where a smaller excitation of the nuclear spins takes place, thus a signal is produced here. Correspondingly, in this example, a decrease in the signal response occurs in the area detected by the field of the inductance.

A preferred embodiment of the invention provides that the resonance circuit is adjusted to the resonance frequency after insertion of the stent into the examination object by unfolding the stent.

Advantageously, inductance and/or capacitance are hereby adjustable for resonant tuning of the resonance circuit. This makes sense, in particular, if, after application or unfolding of the stent, the product of inductance and capacitance, and thus the resonance frequency of the resonance circuit, change.

In an advantageous embodiment of the present invention, at least two resonance circuits are formed or arranged on the stent, whereby the coils of the respective inductance are in particular arranged vertically aligned relative to each other or arranged behind each other. Vertical coils aligned to each other ensure that in every arrangement of the stents in the outer magnetic field,

one component of the inductance runs vertical to the field direction of the outer magnetic field, such that a changed signal response is guaranteed. In addition, using suitable sequence techniques, coils arranged behind each other are particularly suited to carry out a flow measurement (i.e., determination of velocity) of the medium flowing through or along the stent.

Any type of conventional systems may be used for the MR imaging system.

In the stent according to the present invention, in a first alternative the inductance is formed by the material or the skeleton of the stent. Additional parts are thus avoided and the inductance is formed simply and automatically at the time of the unfolding of the stent during application.

The stent thus preferably consists of a material that has at least one layer of high conductivity forming the inductance and one other layer with low conductivity forming the skeleton for the actual stent function. The layer with high conductivity is cut at suitable locations to thus form various areas of the skeleton that are insulated from each other, thus forming an inductance.

Alternatively, the inductance of the resonance circuit is formed by a separate coil that is integrated into the stent skeleton. For instance, the coil is woven, knitted, welded, soldered or

glued into the skeleton of the stent. The coil is connected with the skeleton such that it unfolds with the unfolding of the stent along with the skeleton either by elastic self-expansion or by plastic or thermally induced remote expansion.

The skeleton of the stent is provided preferably in form of a single, double or multiple helix, as a metal frame or as knitted, cut, or etched sheet metal or tube.

Preferably, the capacitance of the stent is at least partially also formed from the stent material, in particular by parallel loops of the inductance, whereby a dielectric may be arranged between the loops. Corresponding surfaces may be formed during the manufacture of the stent. The capacitance may generally be formed as a dielectric by means of suitable arrangements of the conducting layers and a material such as the stent skeleton. With corresponding geometry of the elements forming the condenser, the implantation tissue may also serve as a dielectric.

Alternatively, provision is made to form the capacitance of the stent by means of a separately provided condenser that is connected to the body of the stent.

Advantageously, the apparatus according to the present invention is formed in such a way that with a change in the geometry of the apparatus during application, for instance, by a widening of the stent, the product of inductance and capacitance on the resonance circuit remains essentially

constant, in particular an increase in inductance with a decrease in capacitance occurs or vice versa. This guarantees that the resonance frequency remains essentially unchanged.

Description of several exemplary embodiments

Exemplary embodiments of the present invention are explained in the following in detail with reference to the drawing. They depict:

Fig. 1 - schematically, a stent according to the invention that forms a resonance circuit with an inductance and a capacitance;

Fig. 2a-2g - various electrical diagrams of the apparatus according to the invention;

Fig. 3 - a more precise depiction of the skeleton of the stent of Fig. 1;

Fig. 4a-4b - two examples for the construction of the stent material;

Fig. 5 - a section through the stent material of Fig. 4a;

Fig. 6 - schematically, a stent according to the invention with an integrated coil;

Fig. 7 - a stent according to the invention with a second inductance vertical to the first inductance, and

Fig. 8 - a stent according to the invention with two resonance circuits arranged behind each other.

Figure 1 shows schematically a stent 1 according to the invention, which may, for instance, be made of metals such as platinum, titanium or titanium alloys and compounds, or of plastics or carbon fibers. The area of application of stents is particularly the bridging of narrow areas due to tumors or otherwise narrow areas (e.g., gastro-intestinal and bronchial tract) in internal organs, arterial and venous vessels, vascular strictures, peripheral and central vascular stenoses, in particular in coronary heart disorders. Further applications are the creation of new vascular

passages (shunts) in organs, e.g., in the liver.

The stent effects a mechanical sealing and a permanent widening of the corresponding region and leaves a smooth surface with improved blood flow, enlarging the vascular volume and decreasing the recurrence of occlusions that frequently occur after conventional balloon dilation.

However, the success rate of stents is limited, since in the area of the skeleton, reocclusions that acutely close the lumen may occur e.g., by invasive growth of tumor tissue or by blood clots, so-called thromboses. Due to luxuriant tissue touching the skeleton, new arteriosclerotic deposits may occur and narrow the lumen of the stent again or close it. Since clinical symptoms occur only with more extensive narrowings, which, depending on the degree of occlusion, have an increasingly higher risk of an acute occlusion through thromboses, follow-up examination of the stent function is of utmost importance. However, to date, this is possible only with invasive catherization of the vessel affected, the use of allergenic kidney stressing contrast medium and x-ray techniques. Based on the pronounced susceptibility of artefacts caused by state of the art stents, an MR check-up is virtually impossible. Ultrasound examination are also very limited due to the strong formation of sound echos from the stent skeleton. Until now, the inside of the stent has eluded any form of medical diagnostic imaging.

Stents usually consist of metal skeletons, e.g., continuous metal wires or a type of mesh tube or they are produced from metal tubes by means of laser or spark erosion techniques. Within the framework of this application we always use the term "skeleton" (Gerüst) for all these stent embodiments. For insertion, a stent is, e.g., mounted on a balloon catheter and placed at the

implantation location by means of the catheter and then unfolded, whereby the diameter of the balloon increases the diameter of the stent, pushing it against the wall of the vessel. In addition to balloon expanding plastically deformable stents, self-expanding elastic or thermally expanding stents are known.

Due to their metallic structure, known stents can not be imaged in an MR image, but frequently form partially distinctive artefacts such that precise placement and monitoring of the placement during the process as well as functional monitoring after placement are not guaranteed when using magnetic resonance tomography as an imaging method.

For improved imaging and functional monitoring of the stent in the MR image, the stent 1 according to the invention and according to Fig. 1 is provided with an inductance 2 and a capacitance 3. The inductance of the stent 1 forms the skeleton 2 of the stent 1. Inductance of the stent 1 is formed by the skeleton 2 of the of the stent 1. Provision is made that the individual components of the skeleton 2 are insulated relative to each other as disclosed in detail by way of example in Fig. 3. Insulation of the individual components of the skeleton may already take place during the manufacturing process whereby an insulating layer is applied to the skeleton already formed between the individual phases of the manufacturing process of the stent made from a metal tube.

The inductance 2 is electrically connected to a capacitance 3, whereby inductance 2 and capacitance 3 form a resonance circuit. In Fig. 1, the condenser 3 is provided as a plate condenser with two plates 31, 32. However, any other desired condenser may be used. It is within the framework of this invention that the condenser 3 does not represent an individual

component, but that it consists simply of the inductance 2 from the material of the stent 1, for example, it is formed by parallel wires of the wire skeleton. It is noted, that for reasons of clearer depiction, the electrical connection between the condenser plate 32 and the inductance is not shown in Fig. 1.

Fig. 2a discloses the electrical diagram of the resonance circuit 4 provided in the stent 1, consisting of inductance 2 and capacitance 3. According to Fig. 2b, an optional additional switch 10 is provided, which can be activated or deactivated electrically or magnetically, for instance, mechanically by means of a catheter used in the application.

The resonance circuit 4 can be designed in a great variety of embodiments. According to Fig. 2c, it may have several parallel switched inductances 2a to 2n and according to Fig. 2d it may have several parallel switched capacitances 3a to 3n. Furthermore, several inductances and/or capacitances may be serially switched. Several resonance circuits may also be provided on one stent which may each have a switch and may have serially and/or parallel switched inductances and/or capacitances. Especially with several parallel or serially switched inductances, flow measurements may be refined by means of suitable sequences.

The resonance circuit 4 has a resonance frequency that corresponds to the high-frequency radiation applied to the MR imaging system in which the human body into which the stent is inserted, is placed.

In the stent 1 according to the invention, the resonance circuit 4 is excited by the applied high-frequency pulses of the MR system, since its resonance frequency corresponds to the frequency of the applied HF-pulse. This results in amplification of the magnetic field in the inductance of the resonance circuit or near the inductance which again may result in an amplified excitation of the protons in the corresponding area. In an excitation of the nuclei outside of the inductance by an angle that is smaller than 90°, nuclei within the inductance may experience an excitation of 90° and thus respond at a maximum amplitude. The protons or nuclei arranged in the area of the inductance thus experience a stronger excitation than the protons arranged outside the inductance.

The increase in the deflection angle within the inductance may be up to a factor of 45 in comparison with the protons outside the inductance. It is therefore possible to deflect the protons inside the inductance by an angle of 90° (max. signal response), whereas the protons outside the inductance or outside the magnetic field produced by the resonance circuit, experience no more than a small angle excitation of 2° to 10°. This results in the fact that the inside area of the stent can be imaged substantially brighter in an MR image than the rest of the area. Therefore, the location of the stent in the human body can be precisely determined.

Various designs of the resonance circuit 4 are possible for the tuning of the resonance frequency of the resonance circuit 4 to the frequency of the applied HF pulse.

In one variant, provision is made that the quality of the resonance circuit is kept relatively low in order to realize a resonance circuit with the broadest possible bandwidth and thus to cover the largest possible range of resonance frequencies.

A second variant discloses providing an apparatus with the capability to keep the product of inductance and capacitance constant even after a change of the geometry as was observed in the example referring to the unfolding of the stent. This may take place either in that the stent is given a geometry that changes its properties as little as possible during unfolding of the stent, i.e., in particular, it has a constant inductance and a constant capacitance. A widening of the stent at the implantation location thus substantially causes no change in the resonance frequency of the resonance circuit.

Constancy of the product of inductance and capacitance may be realized, among other things, by a compensation of the changing inductance by a correspondingly changing capacitance. For instance, provision is made that a condenser surface is enlarged or decreased for compensation of a changing inductance by a correspondingly changing capacitance, such that the capacitance increases or decreases according to the corresponding distance between the condenser surfaces. The movability of the condenser plate 32 with regard to the condenser plate 31 and the adjustability of the capacitance are depicted schematically in Fig. 1 by a double arrow.

A third variant provides that an adjustment of the resonance circuit in the magnetic field of the nuclear spin tomograph is induced by a change or adjustment of the inductance and/or the capacitance of the resonance circuit after their placement. For example, a change of the condenser surface is provided by means of the application instrument located in the body, such as a catheter. A decrease in the inductance and thus an adjustment of the resonance circuit to the resonance frequency in the nuclear spin tomograph may take place, for instance, by a laser induced mechanical or electrolytic insulation of coil segments. A change in the capacitance may

also take place by a laser induced mechanical or electrolytic insulation of the capacitance.

Fig. 3 schematically discloses a possible embodiment of a stent according to Fig. 1. According to Fig. 4a and 4b, the stent material consists of two or more layers 81, 82. The first layer 81 is the material for the actual stent function. It has poor conductivity and a high level of stability and elasticity. Suitable materials are mainly nickel-titanium, plastic or carbon fibers. The layer(s) 82 are the material for the formation of the inductance. Thus, the layer 82 has very high conductivity. Suitable materials are mainly gold, silver or platinum which, in addition to their high level of conductivity, are also characterized by biocompatibility. When using less biocompatible electric conductors such as copper, a suitable plastic or ceramic coating may achieve electrical insulation and biocompatibility.

The manufacture of the stent material according to Figs. 4a, 4b takes place, for example, in that a tube made of titanium or titanium alloys or compounds coated with the material for the formation of the inductance are consequently cut by known laser or spark erosion or waterjet cutting techniques.

According to Fig. 3, a coil with the material of Fig. 4a is formed as follows. The stent 1 consist of a two layered material that forms a honey-comb structure 101 and may, e.g., be cut from a tube by means of laser cutting techniques. Fig. 3 shows the tube folded apart. The left and the right side are identical. The conductive layer of the honey-comb structure is interrupted along the

lines 9. For this purpose, the conductive layer is cut during manufacture of the stent after the formation of the structure at the corresponding locations 91 by means of a chemical, physical or mechanical process. Such a location 91 where the conductive layer 82 arranged on the actual stent material is interrupted is schematically depicted in Fig. 5.

By the separation locations 91, a current path through the conductive material 82 is defined as indicated in Fig. 3 by arrows 11. A coil arrangement 2 is created that forms the inductance of the stent 1. Conductive material for the coil function is selected such that the resistance through the conductor formed by the conductive material from one end to the other of the stent is lower than the default resistance of the stent material.

The inductance 2 is formed automatically at the time of the unfolding of the stent material during the application of the stent.

When using a three layered material according to Fig. 4b, the formation of an inductance takes place in a corresponding manner, whereby the layers of the conductive material are provided with separation locations for the formation of a current path. The use of two conductive layers has the advantage that the cross-section of the conductive tracks (lands) is effectively doubled.

In a further development of the exemplary embodiment of Fig. 3 to 5, the conductive layer 82 is in each case additionally coated with an insulating plastic such as a pyrolene in order to safely prevent current flow through the adjacent blood that would decrease the inductance of the coil. Pyrolenes are well suited since they are biocompatible and bond quite well with metal alloys.

To coat the stent with pyrolenes after the manufacturing process, the stent is held in a bath with pyrolenes or vaporized with pyrolenes.

An estimate of the required capacitances and inductances follows for the further disclosure of details of the invention. In the exemplary embodiment, a plate condenser is used and the coil is assumed to be a helix with a fixed number of turns. The resonance frequency of a nuclear spin system is usually in the range between 2 MHz to 90 MHz. The resonance frequency of the nuclear spin system is equal to the product of the magnetic field strength and the gyromagnetic relationship g. At a medium field strength of 1 tesla a resonance frequency of ca. 42 MHz results. The resonance frequency of the resonance circuit is determined by Thomson's resonance equation. It is inversely proportional to the root of the product of the inductance and the capacitance.

The product of conductance and capacitance thus is equal 1.4×10^{-19} S^2. Depending on the number of turns and the stent having an assumed diameter of 8 mm and a length of 40 mm, an inductance of approx. 4×10^{-6} Vs/A results. The resultant surface of a plate condenser with a relative dielectric constant of 2 and a distance of 0.1 mm between the individual plates is approx. 0.2 mm^2. Such a small surface of a plate condenser is easily realized in a stent. With stronger magnetic fields or frequencies, the resultant surface of a plate condenser can be further reduced to 0.014 mm^2.

Fig. 6 depicts an alternative exemplary embodiment of a stent 1′, that forms an inductance 2′ and a capacitance 3′. The inductance 2′ here is provided in form of a helix shaped coil 5 that is not

formed by the skeleton of the stent itself, but rather is an additional wire woven into the stent skeleton 101. In this exemplary embodiment the stent function and the coil function are separated.

The coil 5 is again connected to a condenser 3′ for the formation of a resonance circuit that is either also a separate component or, alternatively, realized by adjacent coil turns or integrated surfaces of the stent.

At the time of the application of the stent, the coil 5, together with the stent material 101 having a smaller radius, is wound onto an application instrument such as a catheter and expands to the desired diameter at the site of the application together with the stent material 101. For this, the wire or the coil 5 is, preferably, provided with a shape memory or the wire or the coil 5 is preloaded on the application instrument.

The intersection surface or the distance of the two condenser plates of the condenser 3′ is in turn designed movable for the adjustment of the resonance frequency of the resonance circuit. However, it is definitely within the framework of the invention that an adjustment to the resonance frequency can take place in a different form or manner than described above.

In the exemplary embodiment of Fig. 7, the inductance 2″ of the stent is depicted schematically. It can be formed either from the stent material (Fig. 3) or as an additional wire (Fig. 6). No individual condenser is provided in this exemplified embodiment. Instead, two loops 21, 22 of the inductance 2″ actually form the capacitance, whereby a dielectric 6 with as high a dielectric

the capacitance is short circuited during excitation, whereas the [sic] resonance circuit is formed without diodes. This results in the fact that during application of high-frequency MR excitation impulses to a subsection of the device, which subsection is surrounded by the resonance circuit without diodes, amplified excitation takes place. However, in the other subsection that is surrounded by the resonance circuit with diodes, a changed signal response now exists compared to the surrounding tissue, as was disclosed with reference to Fig. 2e. With the application of suitable sequence techniques, such an arrangement is particularly effective for the determination of flow and thus for the functional control of the device.

In a further development of the invention (not depicted), provision is made that the inductance of the device itself is used as a receptor coil for the acquisition of MR response signals, whereby the inductance is connected via cable connection to extracorporeal function components. It thus becomes possible to use the inductance of the resonance circuit increasingly actively for the imaging. Due to the necessity of a cable connection to extracorporeal function components this will, however, in general only be possible during a surgical procedure.

The invention is not limited in its embodiment to the previously disclosed exemplary embodiments. Rather, a number of variants which make use of the invention even with fundamentally different types of embodiments, is conceivable.

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Thus, the condenser 3‴ is not short circuited during the emission of MR response signals of the atomic nuclei and a resonance circuit 4‴ is formed that effects an amplification of the emitted MR response signals of the protons and thus changes the measured signal response.

The diodes 12 may be realized in a large variety of ways in the stent skeleton. In particular separate components may be used or the diodes may be formed by or in cooperation with the stent material, such as structures mounted on the stent skeleton.

With structures that are in principle the same as those disclosed in Fig. 2e, the condenser 3‴ in Fig. 2f is not short circuited, but rather the resonance circuit 4‴ is only detuned in the excitation phase by connecting an additional condenser 13, such that an amplified excitation of the nuclear spins takes place to a limited extent only. During the emission of MR response signals, the diodes 12 lock such that the resonance circuit 4‴ is not detuned now and an amplification of the emitted MR response signals takes place, which results in a changed signal response that is imaged in the MR image.

In Fig. 2g the resonance circuit 4‴ is not detuned by connecting a condenser but by connecting a coil 14.

In a further development of the invention, it is possible to also determine the flow velocity of blood flowing through the apparatus. For this, known sequence techniques are used. For instance, saturation impulses are applied to the area of blood supplying tissue, whereby a variation of either the location of the saturation impulses or the time difference between saturation pulse

and the small angle excitation permit computation of flow velocity and, with it, functional information regarding the status of the vessel. Any known methods of flow determination may be used in connection with the stent according to the invention. New sequence techniques may specifically use the characteristics of the stent, i.e., an amplified excitation and an amplified reception or just an amplified reception of the area surrounded by the stent.

Fig. 8 shows a stent 1"" that is preferably used in flow measurements. The stent has two resonance circuits 4a, 4b arranged following each other which are depicted schematically. The resonance circuits 4a, 4b may be formed from the stent material or from additional components as described with reference to the above-described exemplified embodiments. The first resonance circuit 4a is provided with two crossed diodes in accordance with Fig. 2e such that the capacitance is short circuited during excitation. The other resonance circuit 4b is formed without diodes.

This results in the fact that during application of high-frequency MR excitation impulses to a subsection of the stent, i.e., the subsection that is surrounded by the resonance circuit 4b without diodes, amplified excitation takes place. However, with regard to the surrounding tissue, a changed signal response now exists in the other subsection that is surrounded by resonance circuit 4a as was disclosed in Fig. 2e. With the application of suitable sequence techniques, such an arrangement is particularly effective for the determination of flow and thus for the functional control of the stent.

In a further development of the invention (not depicted), a catheter or balloon is equipped with a receptor coil apparatus. Instead of, or in addition to, an external receptor coil of the MR system, the catheter or the balloon receives the signal amplified by the stent and transmits it .

1. MR imaging method for the imaging and determination of the position of a stent with an unfoldable skeleton introduced into an examination object, whereby

a) the examination object is arranged in an external magnetic field,

b) by means of application of high-frequency radiation of a specific resonance frequency, transitions between spin energy levels of the atomic nuclei of the examination object are excited and

c) MR signals thus produced are detected as signal responses, evaluated, and imaged in spatial resolution,

characterized

in that, in a locally defined area in or around the stent, a changed signal response is produced whereby the skeleton of the stent forms or has integrated therein at least one passive resonance circuit with an inductance and a capacitance whose resonance frequency is essentially equal to the resonance frequency of the applied high-frequency radiation and the area is imaged with the changed signal response in spatial resolution.

2. Method according to Claim 1, **characterized in** that the application of the high-frequency radiation excites the resonance circuit and thus an amplified excitation of the nuclear spins of the examination object results in the locally defined area.

3. Method according to Claim 2, **characterized in** that the locally defined area where an amplification of the excitation of the nuclear spins takes place is located within the stent (1).

4. Method according to Claim 1, **characterized in** that with the application of the high-frequency radiation the resonance circuit becomes detuned or the capacitance is short circuited to the extent that no amplified excitation of the nuclear spins takes place in the locally defined area, whereas by measuring of the signal response of the locally defined area the detuning of the resonance circuit or the short circuiting of the capacitance is canceled, thus resulting in a change in the signal response.

5. Method according to at least one of the preceding claims, **characterized in** that the resonance circuit at the stent is adjusted to the resonance frequency by unfolding of the stent after insertion of the stent into the examination object.

6. Method according to at least one of the preceding claims, **characterized in** that the inductance and/or the capacitance may be adjusted for the resonant tuning of the resonance circuit.

7. Method according to at least one of the preceding claims, **characterized in** that at least two resonance circuits formed or arranged on the stent are used, whereby the coils of the respective inductances are arranged differently, in particular aligned vertically to each other or behind each other.

8. Stent with a skeleton that can be unfolded,

characterized

by at least one passive resonance circuit (4, 4‴) with an inductance (2, 2′, 2″, 2‴) and a capacitance (3, 3′, 3‴) whose resonance frequency is essentially equal to the resonance frequency of the applied high-frequency radiation of an MR imaging system, whereby the unfoldable skeleton of the stent is designed as the inductance (2, 2″, 2‴) or the inductance (2′, 2″, 2‴) is integrated into the skeleton such that it unfolds at the time of the unfolding of the stent along with the skeleton.

9. Stent according to Claim 8, **characterized in** that the skeleton consists of a material that has at least one layer (82) that is highly conductive that forms the inductance.

10. Stent according to Claim 9, **characterized in** that the stent material has at least two layers (81, 82), at least one layer (82) with high conductivity and one layer (81) with low conductivity which form the material for the actual stent function.

11. Stent according to Claim 9 or 10, **characterized in** that the layer (82) with high conductivity is separated at suitable locations (91) such that various mutually insulated areas of the skeleton are present, such that an inductance is formed.

12. Stent according to Claim 11, **characterized in** that the skeleton has a honey-comb structure (101), whose conductive layer is separated regularly above and beneath the crossing points of the honey-comb structure (101) to form a coil arrangement.

13. Stent according to at least one of Claims 8 to 12, **characterized in** that the skeleton (2, 2′, 2″) of the stent has the shape of a single, double or multiple helix.

14. Stent according to Claim 8, **characterized in** that the inductance (2′) of the resonance circuit is formed by a separate coil (5) that unfolds along with the skeleton during the unfolding of the stent.

15. Stent according to at least one of Claims 8 to 14, **characterized in** that the capacitance of the stent is at least partially formed by the stent material, in particular by two loops (21, 22) of the inductance, whereby a dielectric (6) is arranged between the loops.

16. Stent according to at least one of Claims 8 to 14, **characterized in** that the capacitance of the stent is formed by a separately provided condenser, in particular a plate condenser or a cylinder condenser.

17. Stent according to at least one of Claims 8 to 16, **characterized in** that the stent has means (13) for detuning the resonance circuit when applying the high-frequency radiation.

18. Stent according to Claim 17, **characterized in** that the means for detuning the at least one resonance circuit are designed such that they switch a condenser (13) parallel to the capacitance (3‴) of the resonance circuit with the application of high-frequency radiation.

19. Stent according to Claim 17, **characterized in** that the means for the detuning the at least one resonance circuit are designed such that they switch a coil (14) parallel to the inductance (2‴) of the resonance circuit with the application of high-frequency radiation.

20. Stent according to at least one of Claims 8 to 16, **characterized in** that the stent is provided with means (12) for the short circuiting of the capacitance (3‴) with the application of high-frequency radiation.

21. Stent according to Claim 20, **characterized in** that the means for the short circuiting of the capacitance have two diodes (12) which are switched parallel to the capacitance (3‴).

22. Stent according to at least one of Claims 8 to 21, **characterized in** that a switch (10) is provided, by which the at least one resonance circuit can be activated or deactivated.

23. Stent according to at least one of Claims 8 to 22, **characterized in** that the inductance (2) and/or the capacitance (3) of the resonance circuit are adjustable for tuning to the resonance frequency of the MR system.

24. Stent according to at least one of Claims 8 to 23, **characterized in** that the stent is formed such that when a change in the geometry of the stent occurs during its application that the product of the inductance and the capacitance of the resonance circuit substantially remains constant, in particular in that an amplification of the inductance accompanies a decrease in capacitance.

25. Stent according to at least one of Claims 8 to 24, **characterized in** that the quality of the resonance circuit (4) is relatively low.

26. Stent according to at least one of Claims 8 to 25, **characterized in** that the resonance circuit (4) has several parallel or serially switched inductances (2a, 2n) and/or capacitances (3a, 3n).

27. Stent according to at least one of Claims 8 to 26, **characterized in** that the stent has several resonance circuits (2″, 7; 4, 4b) with several inductances which are arranged differently, in particular aligned vertical to each other or arranged behind each other.

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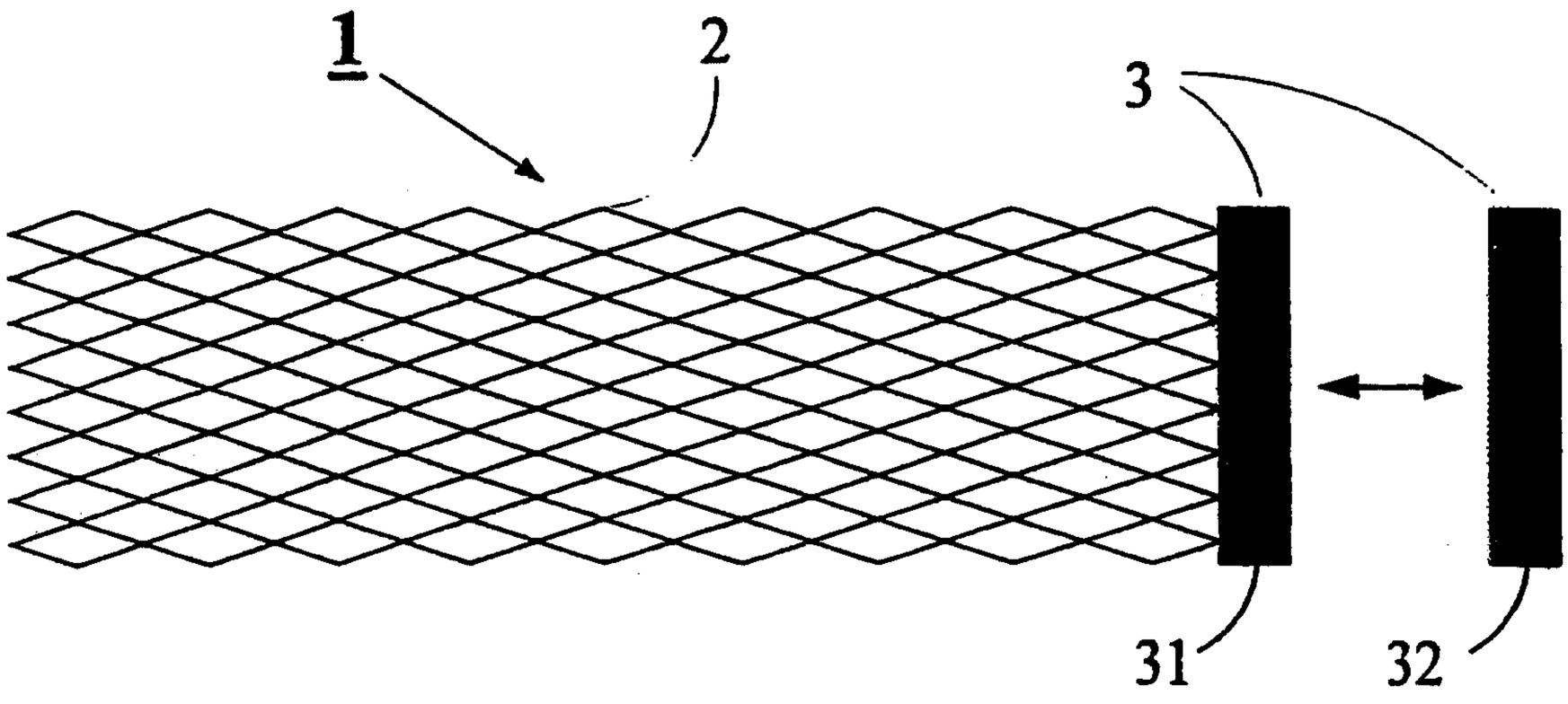

Fig. 1

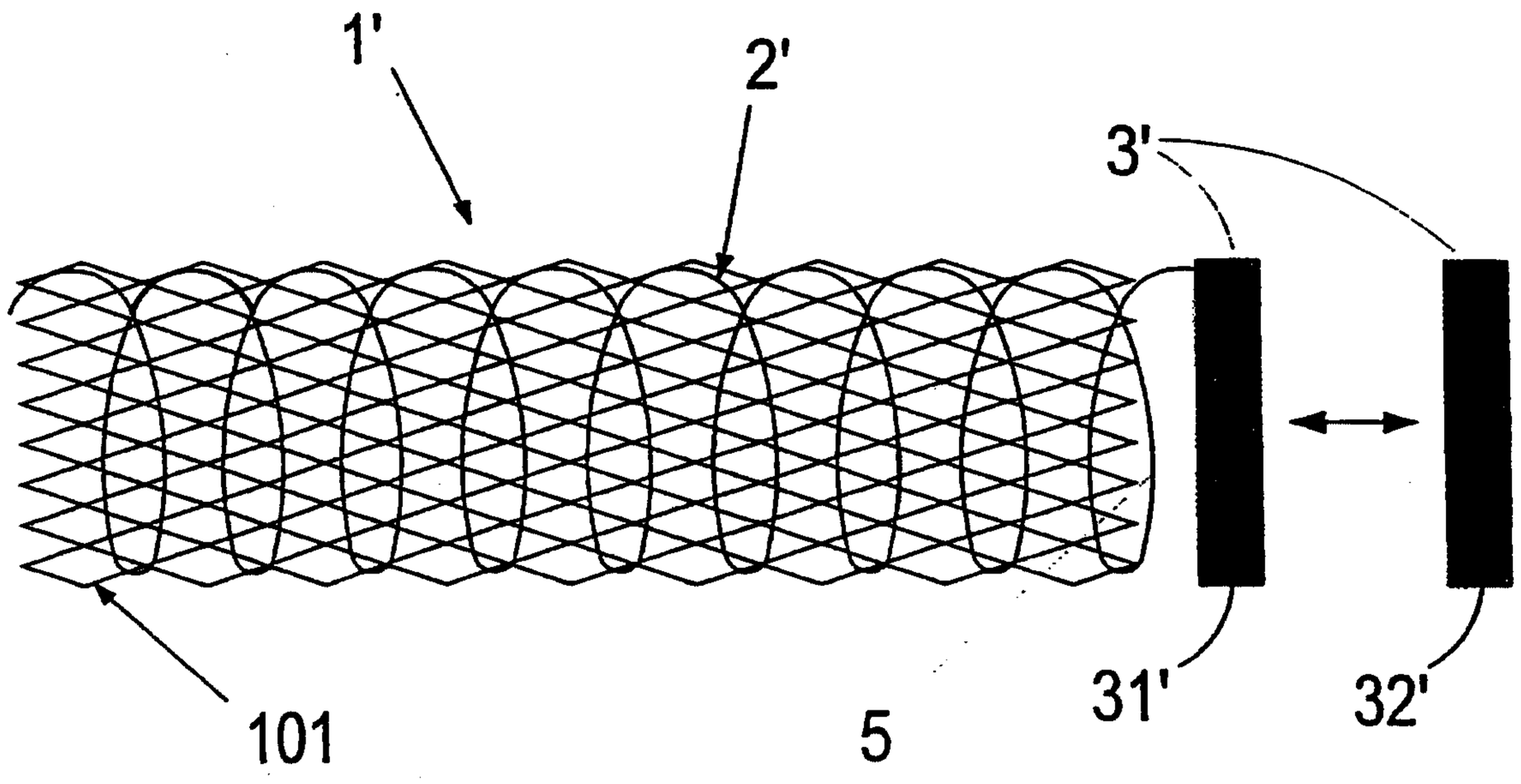

Fig. 6

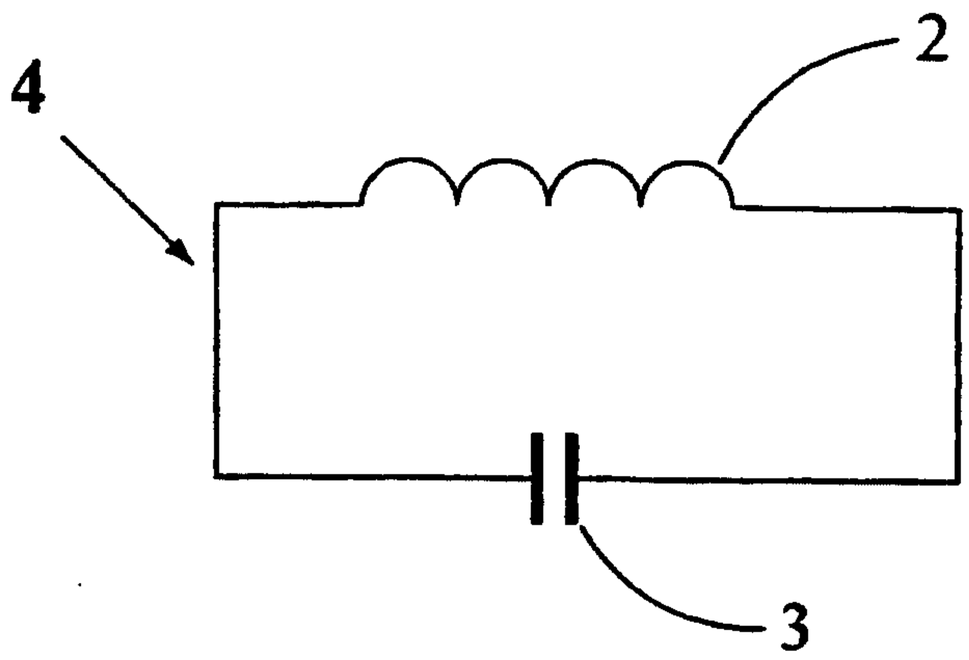

Fig. 2a

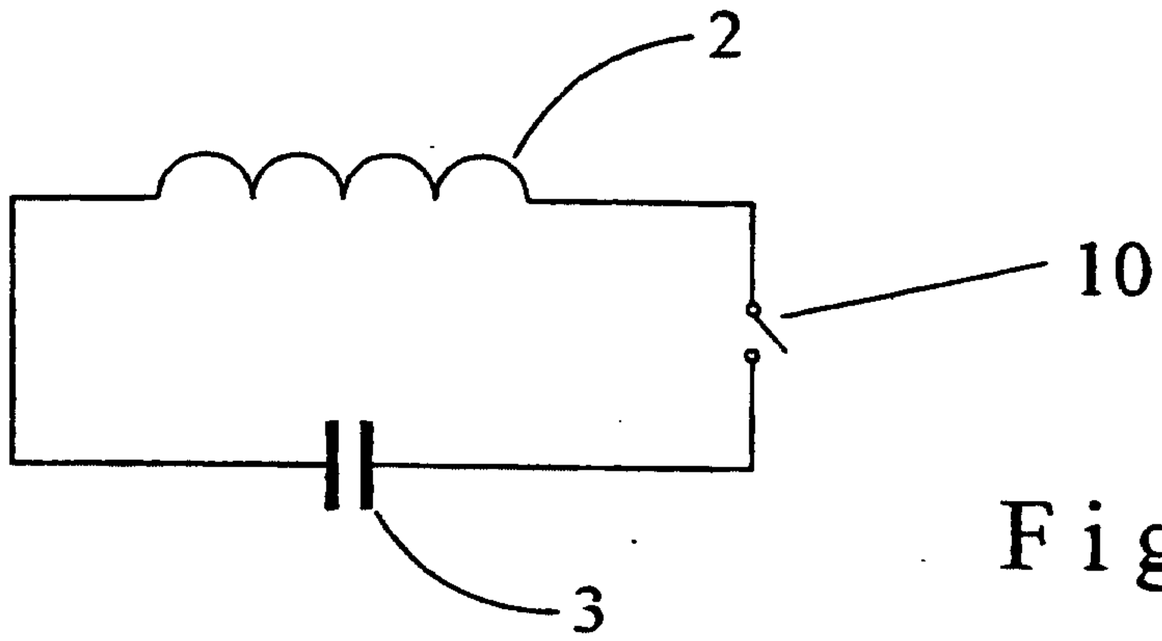

Fig. 2b

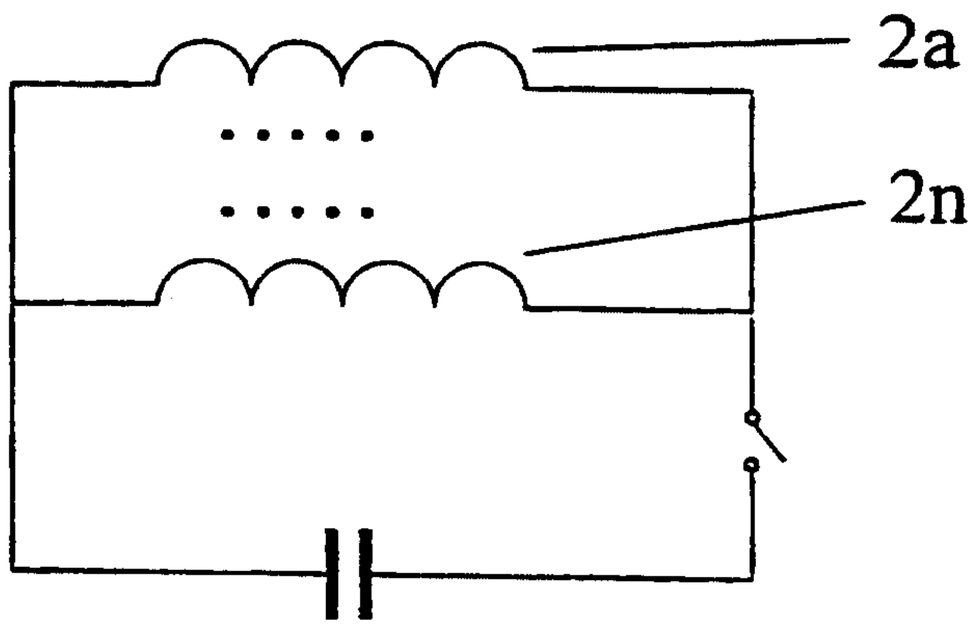

Fig. 2c

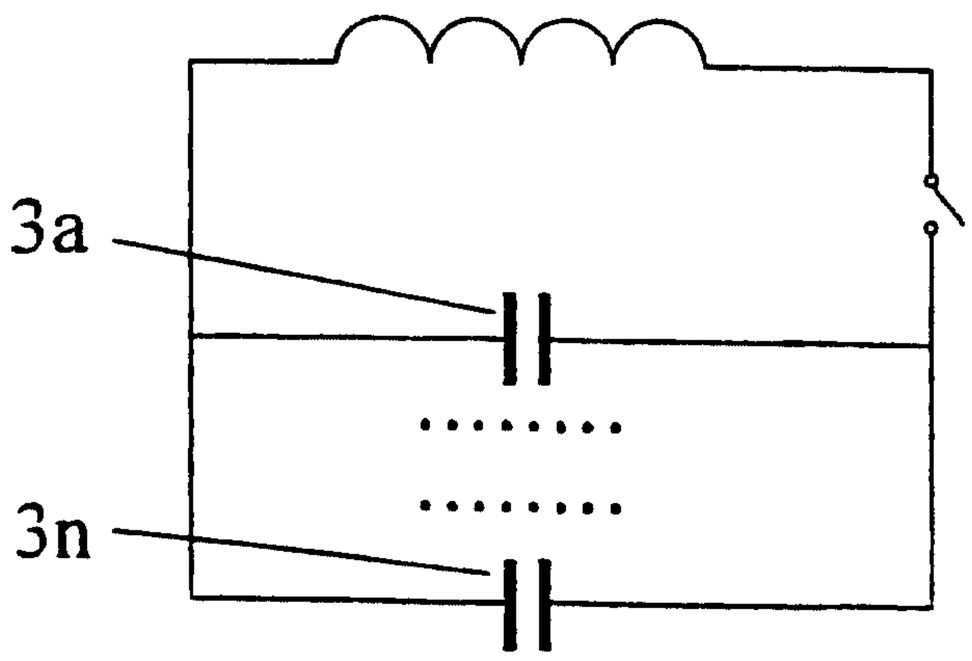

Fig. 2d

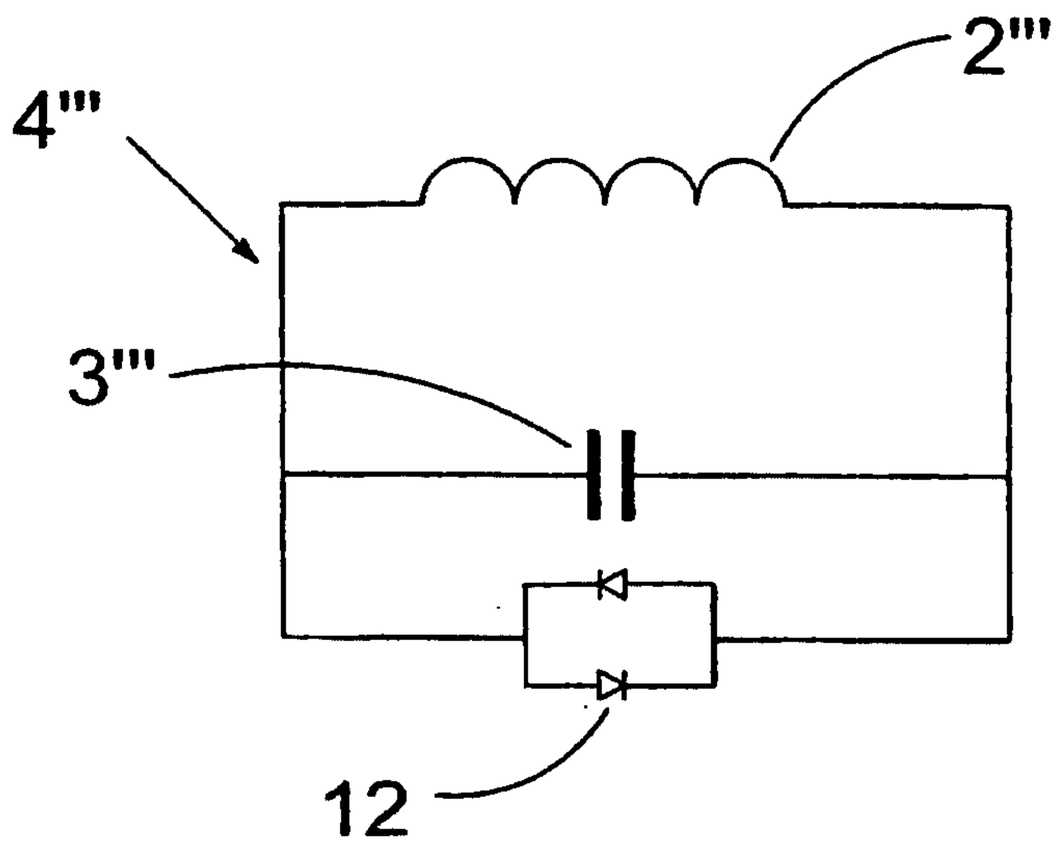

Fig. 2e

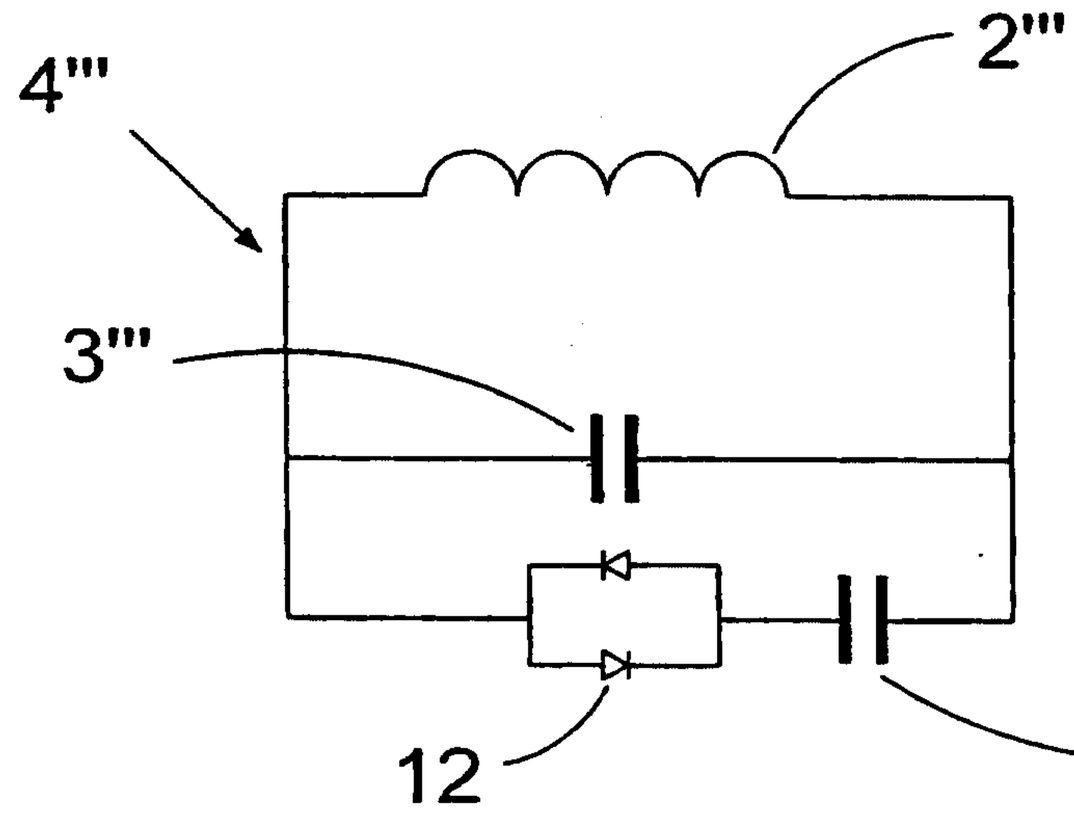

Fig. 2f

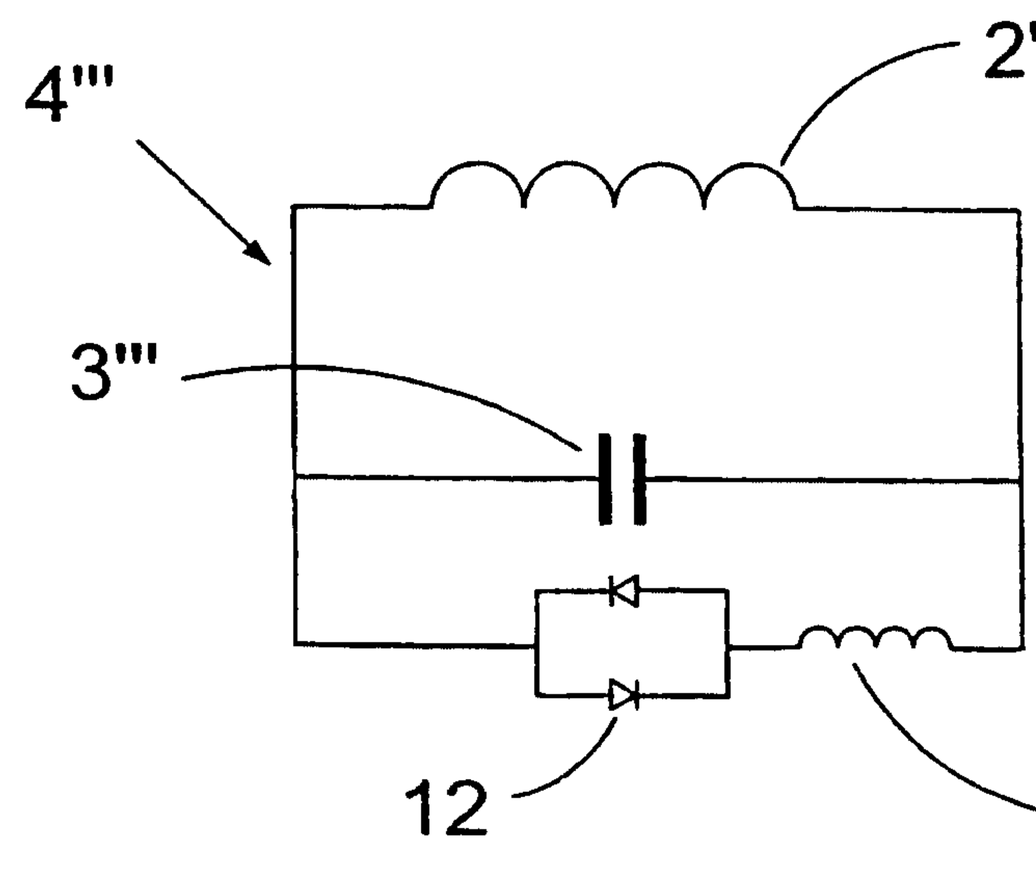

Fig. 2g

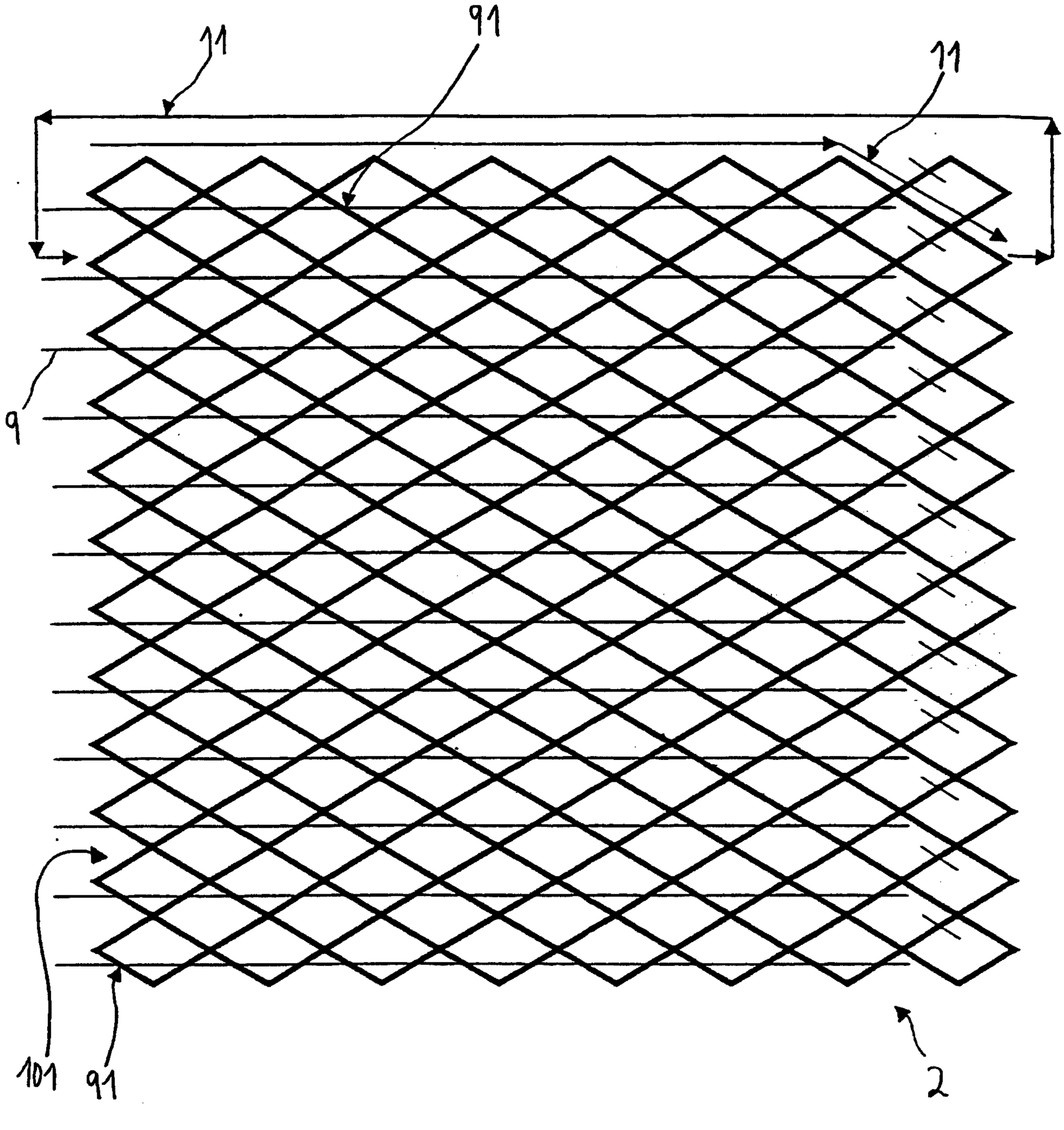

Fig. 3

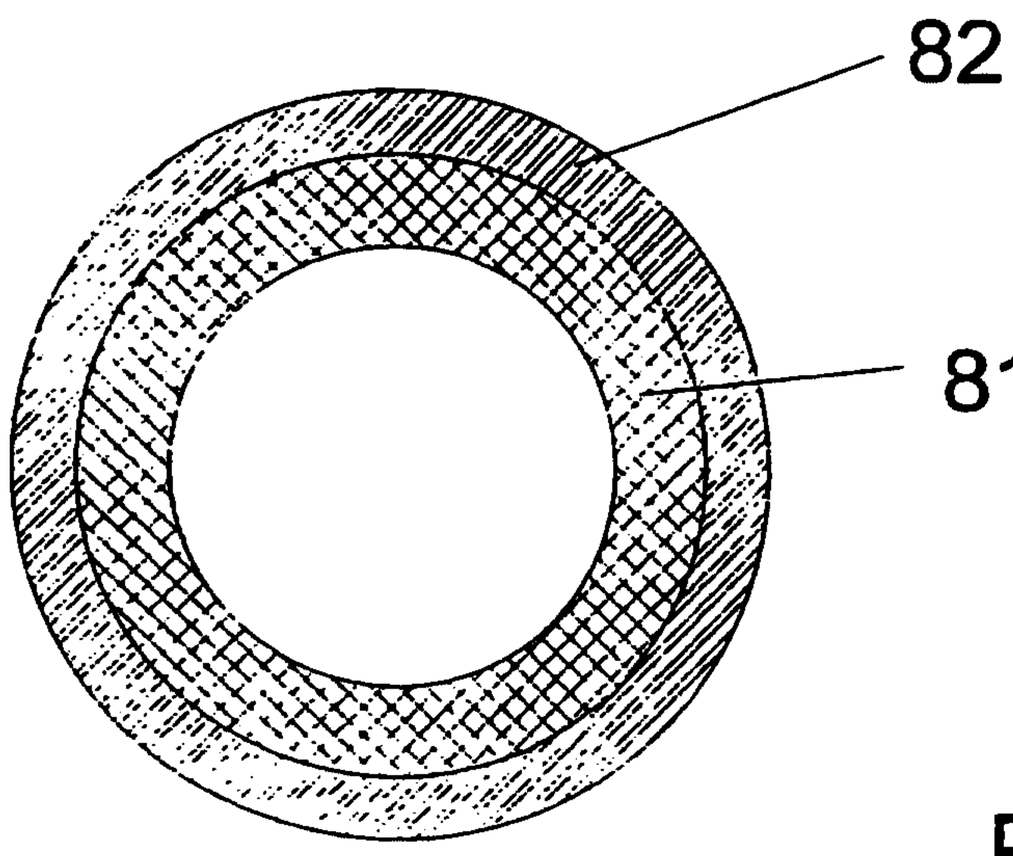

Fig.4a

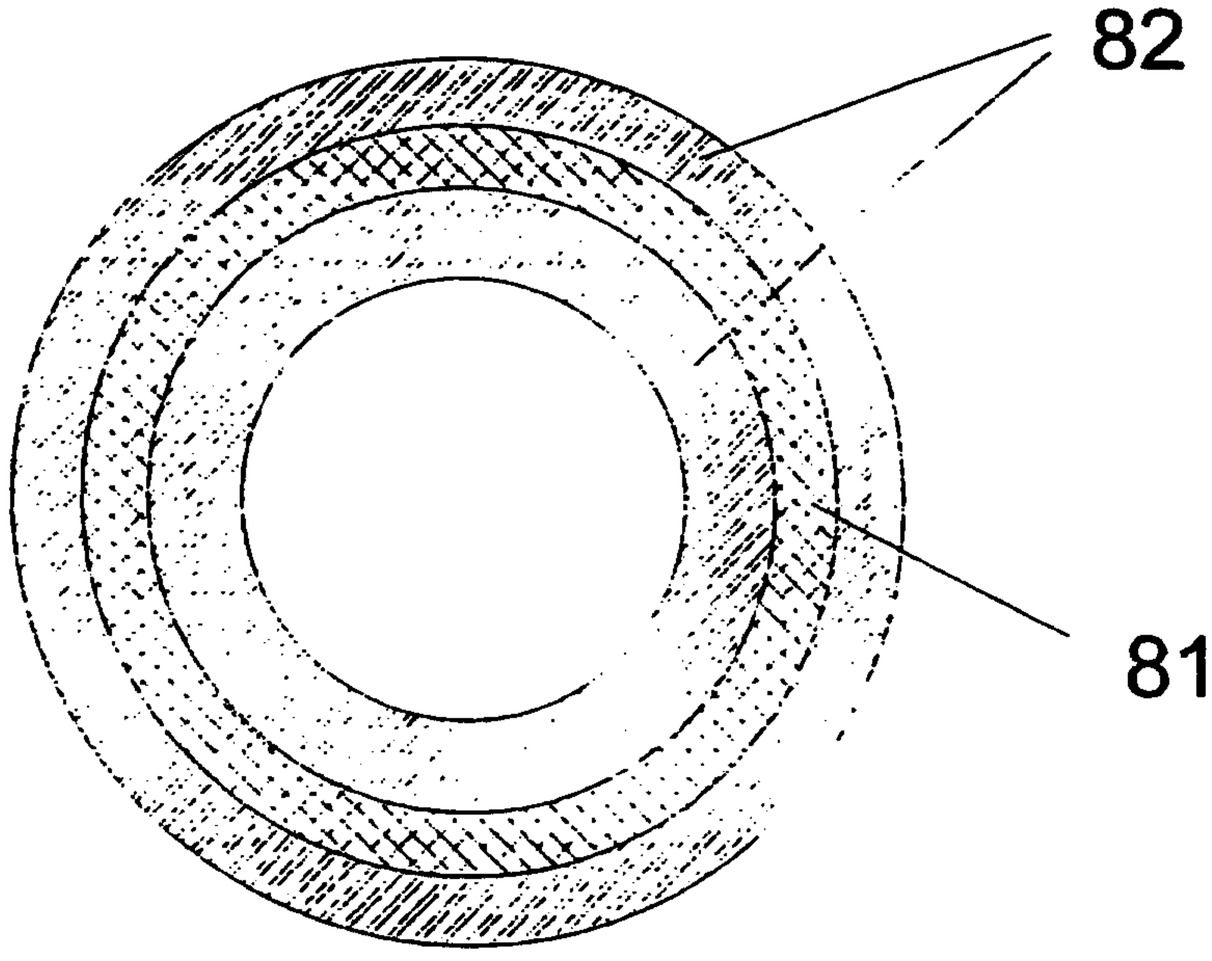

Fig.4b

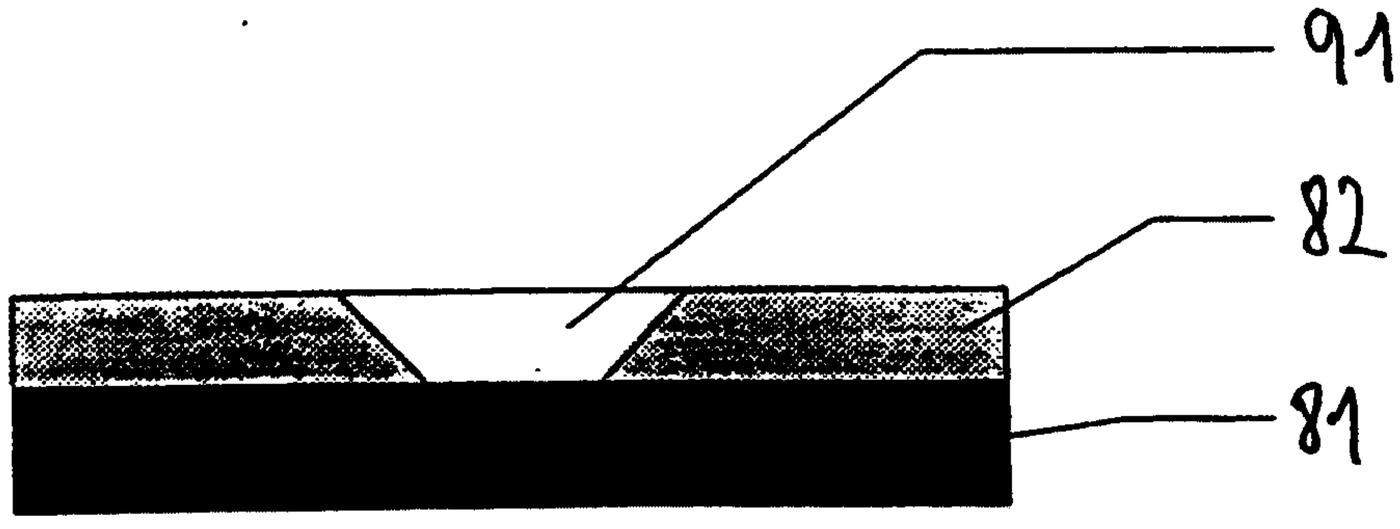

Fig. 5

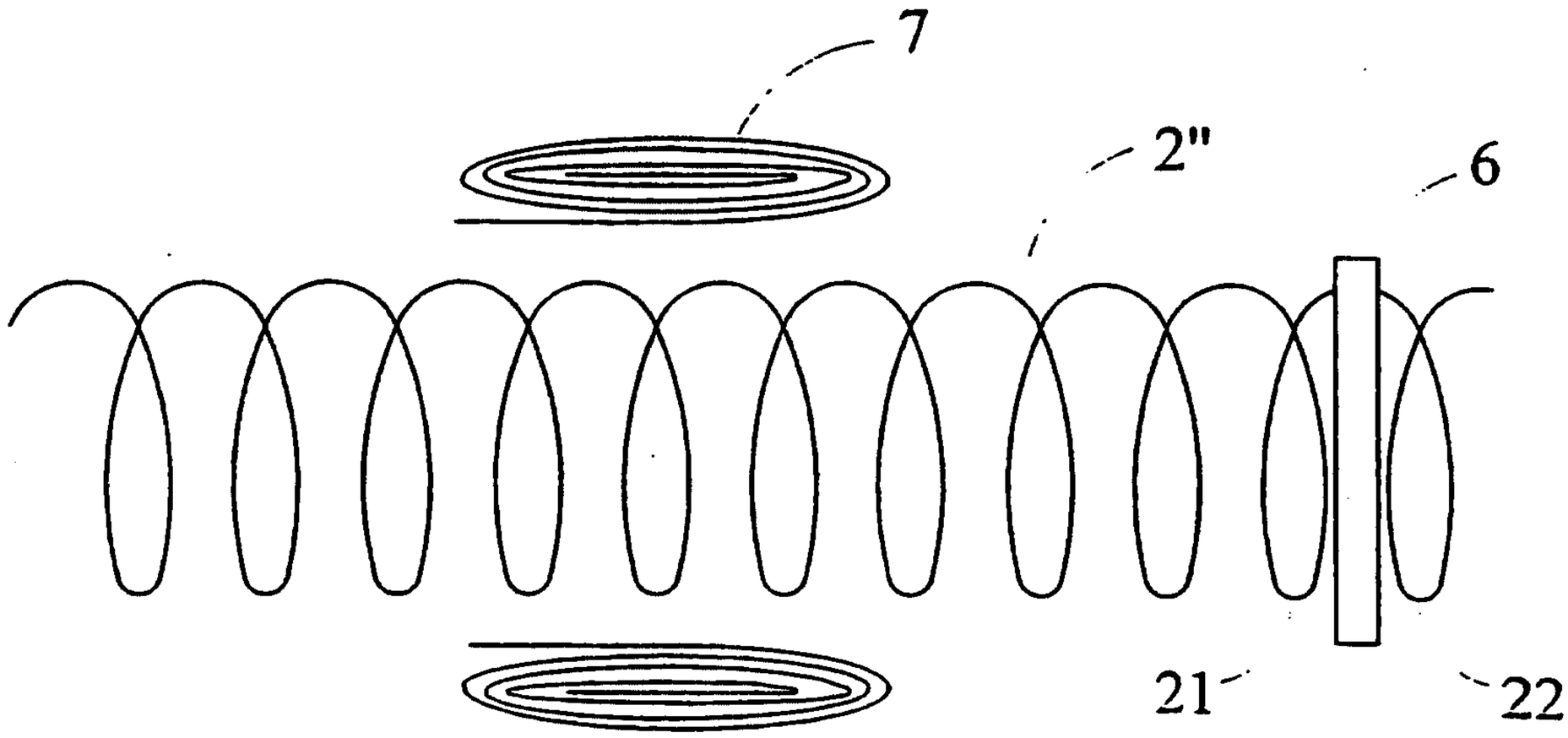

Fig.7

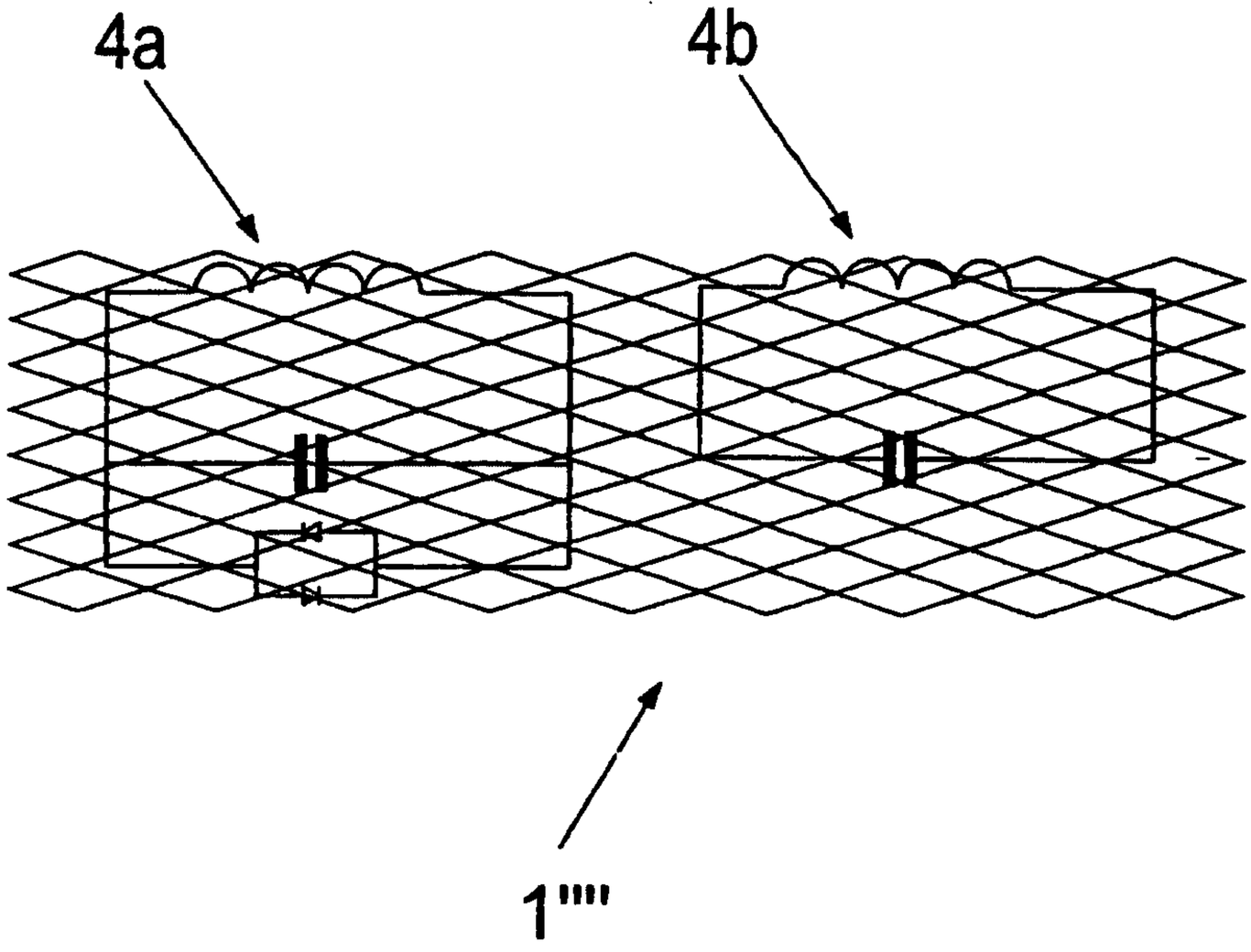

Fig.8