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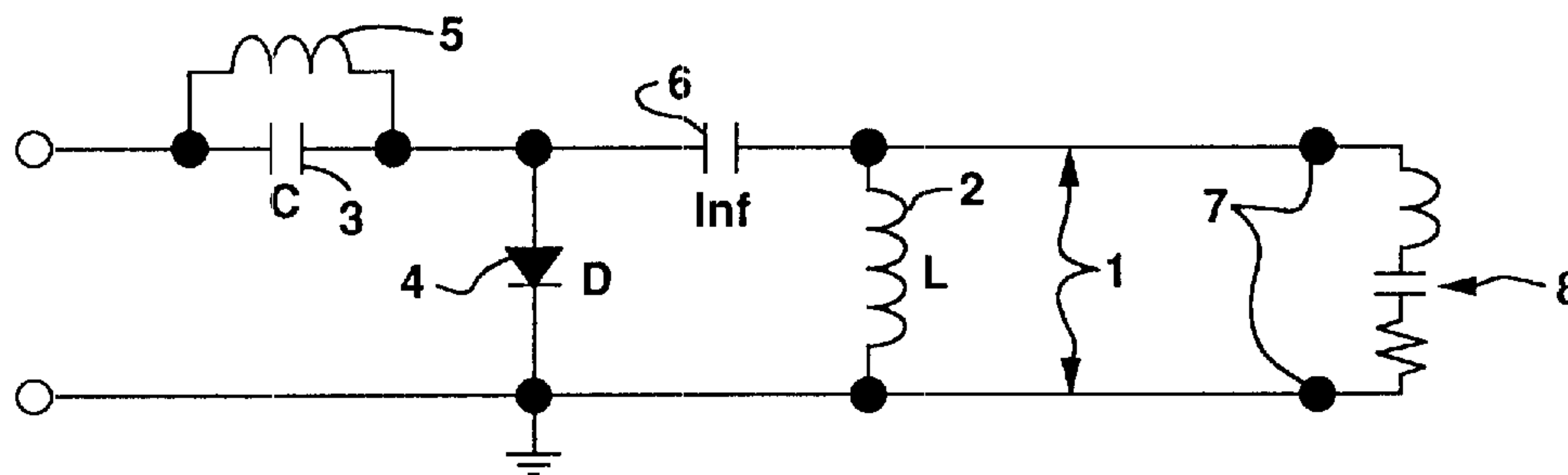
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(54) **BOBINE D'IMAGERIE A RESONANCE MAGNETIQUE**

(54) **MAGNETIC RESONANCE IMAGING COIL**



(57) A receive loop for use in magnetic resonance imaging, comprising: a collapsible imaging coil adapted to be inserted into a patient's body and expanded therein for receiving radio frequency signals emitted by the body, the coil being characterized by an inductive impedance; a quarter-wavelength transmission line connected to the coil, the imaging coil and transmission line in combination being characterized by a capacitive impedance; and apparatus connected to the transmission line remote from the imaging coil for tuning the combination coil and transmission line to resonate in response to the body emitting the radio frequency signals, whereby the radio frequency signals are received.



ABSTRACT

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A receive loop for use in magnetic resonance imaging, comprising: a collapsible imaging coil adapted
5 to be inserted into a patient's body and expanded therein for receiving radio frequency signals emitted by the body, the coil being characterized by an inductive impedance; a quarter-wavelength transmission line
10 connected to the coil, the imaging coil and transmission line in combination being characterized by a capacitive impedance; and apparatus connected to the transmission line remote from the imaging coil for tuning the combination coil and transmission line to resonate in
15 response to the body emitting the radio frequency signals, whereby the radio frequency signals are received.

MAGNETIC RESONANCE IMAGING COILField of the Invention

The present invention relates in general to magnetic resonance imaging, and more particularly to a receive
5 coil designed to collapse into a small diameter catheter.

Background of the Invention

High resolution imaging of internal organ systems using conventional magnetic resonance imaging (MRI) is limited by the size and location of the receive coil used
10 to pick up weak radio frequency signals emitted by a patient's body. Increased image quality can be achieved through the use of "surface coils" which are small receive coils placed on the patient's body surface as close as possible to the organs of interest. Small
15 "surface coils" result in good image quality. Unfortunately, as the dimensions of the coil are reduced, so also is the detection sensitive volume of the coil, thereby limiting the use of small surface coils to organs which are relatively close to outside body surfaces.

20 Consequently, obtaining very high resolution magnetic resonance (MR) images of internal organs that are not close to an external body surface has been found to be problematic. In an effort to overcome the disadvantages discussed above, the suggestion has been
25 made to use internal MRI receive coils which are designed to be inserted into the patient and placed much closer to the organ of interest than would be possible with an external surface coil.

The first description of internal magnetic resonance
30 receive coils appears in an article by H.L. Kantor, R.W. Briggs, and R.S. Balaban in Circ. Res. 55:261-266, 1984. These authors used internal magnetic resonance receive coils for NMR spectroscopy in laboratory animals. The use of these coils for magnetic resonance imaging was not
35 suggested.

Various investigators have subsequently disclosed designs and applications of internal coils for magnetic resonance imaging. Reference may be made to the following published articles in this area:

Harman RR, Young IR, Bydder GM, Butson PC, Spencer
D. Proc SMRM 5:56-57, 1986;

Patrick JL, Mehdizadeh M, Hurst GC, Proc SMRM 5:47-
48, 1986;

5 Schnall MD, Kressel HY, Pollack HM, Lenkinski RE,
Proc SMRM 5:197, 1986;

Wesbey GE, Higgins CB, Hale JD, Valk PE, Card.
Interv, Rad 8:342, 1986;

10 Grist TM, Kneeland JB, Stafl A, Hyde JS, Slane JMK,
Jesmanowicz A., Proc SMRM 6:221, 1987;

Harman RR, Butson PC, Hall AS, Magn. Res. Med. 6:49,
1988; and

Misic. GJ, Rhinehart EJ, Claiborne TC, Proc SMRM
8:179, 1989;

15 The prior art designs can be distinguished on the
basis of whether or not they fit inside a catheter
(typically 1-3 mm diameter), and also whether or not they
expand to a larger size upon placement inside the body.
The following table illustrates the earliest publications
20 in each of these categories.

TABLE 1

	Catheter-based	Non-catheter-based
Expandable	Kantor 1984	Schnall 1986
Non-expandable	Patrick 1986	Implanted Coils

25

The class of coils which are most widely applicable
to intraluminal imaging of internal organs (e.g. bladder,
prostate, vasculature, etc.) are those which are both
catheter-based and expandable. Only the earlier Kantor
30 disclosure describes coils of this type. However, as
discussed above, Kantor et al worked primarily in the NMR
spectroscopy area, and did not apply their expandable
catheter-based design to the problem of MR imaging.

Non-catheter-based receive coils are generally of
35 large dimension (e.g. 5-15 mm diameter) and of low
flexibility, and are applicable to only a few organs

(prostate, cervix). Non-expandable catheter-based designs are useful only for imaging structures which are extremely close to the catheter, such as the walls of blood vessels, and do not permit imaging of most organs.

5 The expandable catheter-based design of Kantor et al requires electrical components to be situated at the tip of the receive coil, which means the coil is not collapsible to small dimensions as would be the case if there were no components at the coil tip.

10 Summary of the Invention

According to the present invention, a receive loop is provided in which substantially all electrical components are placed remote from the imaging coil, such that the collapsed coil diameter is minimized. A
15 quarter-wavelength transmission line is provided between the coil and the electrical components for tuning and matching of frequencies. In addition, according to the preferred embodiment, a hinged coil tip is provided for an enhanced circular expanded coil shape, and a locking
20 mechanism may be used to lock the coil into an expanded state. Moreover, the expandable catheter-based receive coil of the present invention has been adapted specifically for MR imaging, not NMR spectroscopy, as in Kantor et al.

25 In general, according to an aspect of the present invention there is provided a receive loop for use in magnetic resonance imaging, comprising:

a) a collapsible imaging coil adapted to be inserted into a patient's body and expanded therein for
30 receiving radio frequency signals emitted by said body, said coil being characterized by an inductive impedance;

b) a quarter-wavelength transmission line connected to said coil, said imaging coil and transmission line in combination being characterized by a
35 capacitive impedance; and

c) means connected to said transmission line remote from said imaging coil for tuning said combination

coil and transmission line to resonate in response to said body emitting said radio frequency signals, whereby said radio frequency signals are received.

Brief Description of the Drawings

5 A preferred embodiment of the invention will be described in greater detail below with reference to the following drawings, in which:

Figure 1 is a schematic diagram of the receive loop according to the present invention;

10 Figure 2 is a schematic representation of the coil design according to a successful prototype,

Figure 3 illustrates a preferred embodiment of the coil design having a hinged tip;

15 Figure 4 is a schematic illustration of the coil and associated electrical components; and

Figures 5a to 5d illustrate three alternative embodiments of the coil.

Detailed Description of the Invention

20 As discussed above, high resolution imaging of internal organ systems using a conventional magnetic resonance imaging (MRI) is partially limited by the size of external body and surface coils and the proximity with which these coils can be placed with respect to the organ of interest. Increased spacial resolution can
25 potentially be achieved by the use of internally placed imaging coils.

Practical design of internal coils must address two important criteria: 1) Robustness and reusability; and
30 2) Expandability from relatively small collapsed dimensions to relatively large expanded dimensions, rendering these coils useful for a variety of clinical applications.

According to the preferred embodiment, all electronic components have been placed external to the
35 coil to achieve a minimum collapsed diameter, and the individual coil conductor strips have been insulated to

allow for a simple, robust mechanical loop design which can be easily sterilized between uses.

Turning to Figure 1, a schematic representation is shown for a preferred embodiment of the invention, comprising a quarter-wavelength coaxial cable 1 for transforming the coil impedance from inductive to capacitive. Parallel inductor 2 and series capacitor 3 are used to match and tune the loop impedance to 50 ohms at the resonant frequency of the MRI (e.g. 63.8 MHZ), and a PIN diode 4 is used in conjunction with the quarter wavelength transformer to decouple the inductor 2 from the imaging coil 8 when the MRI is in transmitting mode. An additional inductor 5 and capacitor 6 are used as a DC bypass and block, respectively.

According to the preferred embodiment, the capacitor 3 comprises a high-Q porcelain chip capacitor (e.g. ATC - 100B or DLI - C17). The inductor 2 preferably comprises an 18Awg enamelled copper magnet wire, and the cable 1 preferably comprises a low-loss, micro-coaxial cable, 0.055" O.D. (e.g. Gore-Tex™, CXN-2056) characterized by a quarter wavelength of the MR signal at frequency of 63.8 MHZ.

The well known MRI scanning sequence includes a transmission stage in which the patient to be imaged is exposed to a radio frequency (RF) excitation pulse, whose envelope power level can reach 20kW. During this stage of the scanning sequence, the currents induced in the imaging coil 8, due to magnetic flux linkage, must be minimized for two reasons. Firstly, in order to exclude the possibility of core heating through resistive losses, such heating possibly leading to patient burns, and secondly, to maintain magnetic field homogeneity.

Forward biasing the PIN diode 4 with a DC voltage pulse synchronized with the RF excitation pulse, both being provided by the MRI scanner (not shown), results in a short circuit across the diode. This appears at the imaging coil terminals 7 as an open circuit due to the

transformation properties of the quarter wavelength cable 1, and hence current flow in the coil 8 due to inductive coupling is minimized.

During the receive stage of the scanning sequence, the echoes emitted by the subject being imaged are inductively coupled to the coil 8, as the diode 4 is reversed biased, and sent via the transmission line to the MRI receiver front-end. Hence, the diode 4 and quarter wavelength cable 1 work in combination to ensure patient safety and maximized image quality.

The shape, material properties, width, thickness and overall size of the coil 8 are critical for determining the field of view of resultant images and the ability of the coil to open inside organs or body cavities.

In one prototype of the coil 8 illustrated in Figure 2, a 42 mm x 90 mm coil was formed from a flat strip of beryllium copper, which is a highly conductive non-magnetic alloy of high modulus of elasticity. The coil material was 0.010" thick x 0.040" wide beryllium copper (e.g. Brush Wellman 190 HM Temper). With this prototype coil, image quality in human bladder specimens and live pig bladders were found to be excellent. However, the magnetic field depth of penetration into the bladder wall, at the tip, was limited by the sharp point 9.

The preferred embodiment of coil 8 is illustrated in Figure 3, employing a hinge 10 in the form of a single turn of spring at the coil tip. The hinged tip design increases the useful image depth at the tip by permitting the entire coil to be brought closer to the organ wall at the tip than would be possible with the embodiment of Figure 2, without sacrificing the collapsibility properties of the coil. For the embodiment of Figure 3, the coil material was comprised of 0.012" thick x 0.0625" wide, beryllium copper.

Turning to Figure 4, the coil 8 is shown attached to a low-loss non-magnetic microcoaxial cable nested within a 7 French catheter 11, forming an inner assembly. An

outer 8 French catheter 12 fits over the assembly in a coaxial fashion, allowing for compression of the coil to 3 mm x 105 mm. The outer catheter 12 is provided with a valve and side port extension 13, of a well known design.

5 All discrete components are housed according to the preferred embodiment in an interface box 14 at a distance of one quarter wavelength from the coil 8, and an additional quarter wave coaxial cable (not shown) provides connection between the interface box 14 and the
10 MRI receiver (not shown) by means of a BNC female adapter 15. The electronics assembly within box 14 is therefore reusable, the coil 8 is sterilizable and the outer sheath 12 is disposable.

Excessive image brightness which may occur due to
15 high signal intensity near the surfaces of the coil can be compensated by a number of methods, such as post-processing or loading of the coil insulation with paramagnetic or superparamagnetic material.

The images obtained from the emission prototype
20 probes according to the present invention have been found to be superior to those obtained from external body or surface coils and demonstrates the clinical utility of such devices. Potential applications of the coil include gastrurinary, gastrointestinal tract, biliary tree and
25 vascular studies.

In summary, the novelty of mounting all electronic components remote from the coil in a component box, rather than on the coil itself, provides greater flexibility in mechanical design of the receive coil
30 loop. This flexibility permits, for example, small collapsed dimensions and the ability to hinge the tip of the coil, as discussed above with reference to Figure 2. Neither expandable catheter-based coil designs or "component-free" designs are believed to be known in the
35 art.

Other embodiments or variations of the design are possible. For example, the electrical components can be

placed at opposite ends of the coil 8 for increased image quality, as shown in the alternative embodiment of Figure 5a. However, this design will increase the minimum collapsed coil diameter. In addition, the coil 8 can be
5 fixed in an expanded position using a locking mechanism such as a locking wire 16, as shown in the embodiment of Figure 5b. Furthermore, the coil 8 can be designed to be in the form of two or more loops rather than a single loop (Figure 5c and 5d) forming a "wire basket", for more
10 uniform image intensity around the coil.

All such embodiments and variations are believed to be within the sphere and scope of the invention as defined in claims appended hereto.

THE EMBODIMENTS OF THE INVENTION IN WHICH AN EXCLUSIVE
PROPERTY OF PRIVILEGE IS CLAIMED ARE DEFINED AS FOLLOWS:

1. A receive loop for use in magnetic resonance
5 imaging, comprising:
 - a) a collapsible imaging coil adapted to be
inserted into a patient's body and expanded therein for
receiving radio frequency signals emitted by said body,
said coil being characterized by an inductive impedance;
 - 10 b) a quarter-wavelength transmission line
connected to said coil, said imaging coil and
transmission line in combination being characterized by a
capacitive impedance; and
 - c) means connected to said transmission line
15 remote from said imaging coil for tuning said combination
coil and transmission line to resonate in response to
said body emitting said radio frequency signals, whereby
said radio frequency signals are received.
- 20 2. The receive loop as defined in claim 1, further
comprising means for selectively decoupling said imaging
coil from said transmission line.
3. The receive loop as defined in claim 1, wherein
25 said imaging coil further includes a hinged tip.
4. The receive loop as defined in claim 1, wherein
said imaging coil and transmission line are encased
within a catheter for facilitating compression of the
30 coil prior to insertion of the coil into the patient's
body and expansion of the coil thereafter.
5. The receive loop as defined in claim 1, further
comprising means for locking said imaging coil expanded
35 in said patient's body.

6. The receive loop as defined in claim 1, wherein said means connected to said transmission line comprises an inductor connected in parallel with said transmission line and a capacitor connected in series with said transmission line, for tuning said combination imaging coil and transmission line at the frequency of said radio signals.

7. The receive loop as defined in claim 6, wherein said means for decoupling comprises a diode connected in parallel with said transmission line, for presenting a short circuit to said transmission line when forward biased.

8. The receive loop as defined in claim 6, wherein said capacitor is a high-Q porcelain capacitor and said inductor is constructed from Awg enamelled copper wire.

9. The receive loop as defined in claim 1, wherein said transmission line comprises a low-loss, micro-coaxial cable.

10. The receive loop as defined in claim 1, wherein said imaging coil is configured in a loop terminating in a sharp point and having dimensions of 42 mm x 90 mm, said coil being fabricated from a flat strip of beryllium copper 0.010" thick x 0.040" wide.

11. The receive loop as defined in claim 1, wherein said imaging coil is configured in a loop terminating in a hinged tip and having dimensions of 42 mm x 90 mm, said coil being fabricated from a flat strip of beryllium copper 0.012" thick x 0.0625" wide.

12. The receive loop as defined in claim 1, further comprising capacitors disposed at opposite ends of said imaging coil.

13. The receive loop as defined in claim 5, wherein said means for locking comprises a wire lock connected between opposite ends of said imaging coil.

5 14. The receive loop as defined in claim 1, further comprising one or more additional imaging coils oriented to form a wire basket.

FIG. 1

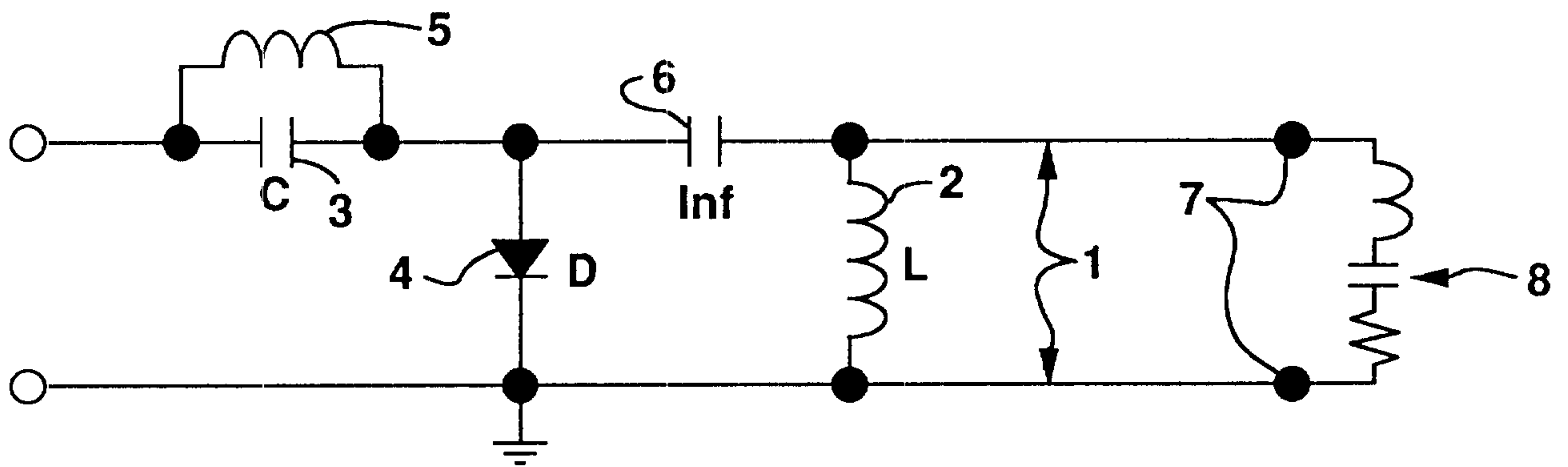


FIG. 2

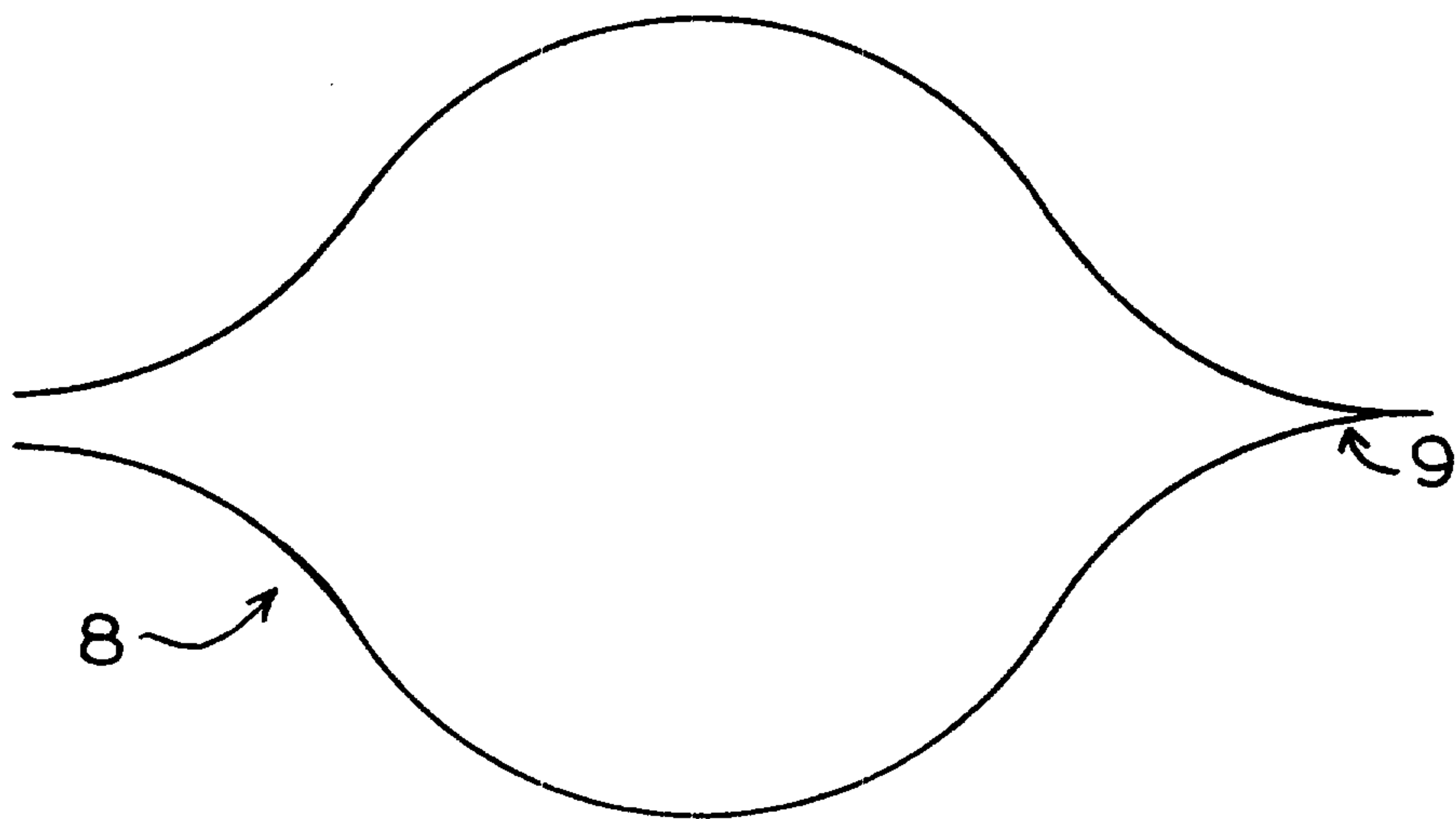
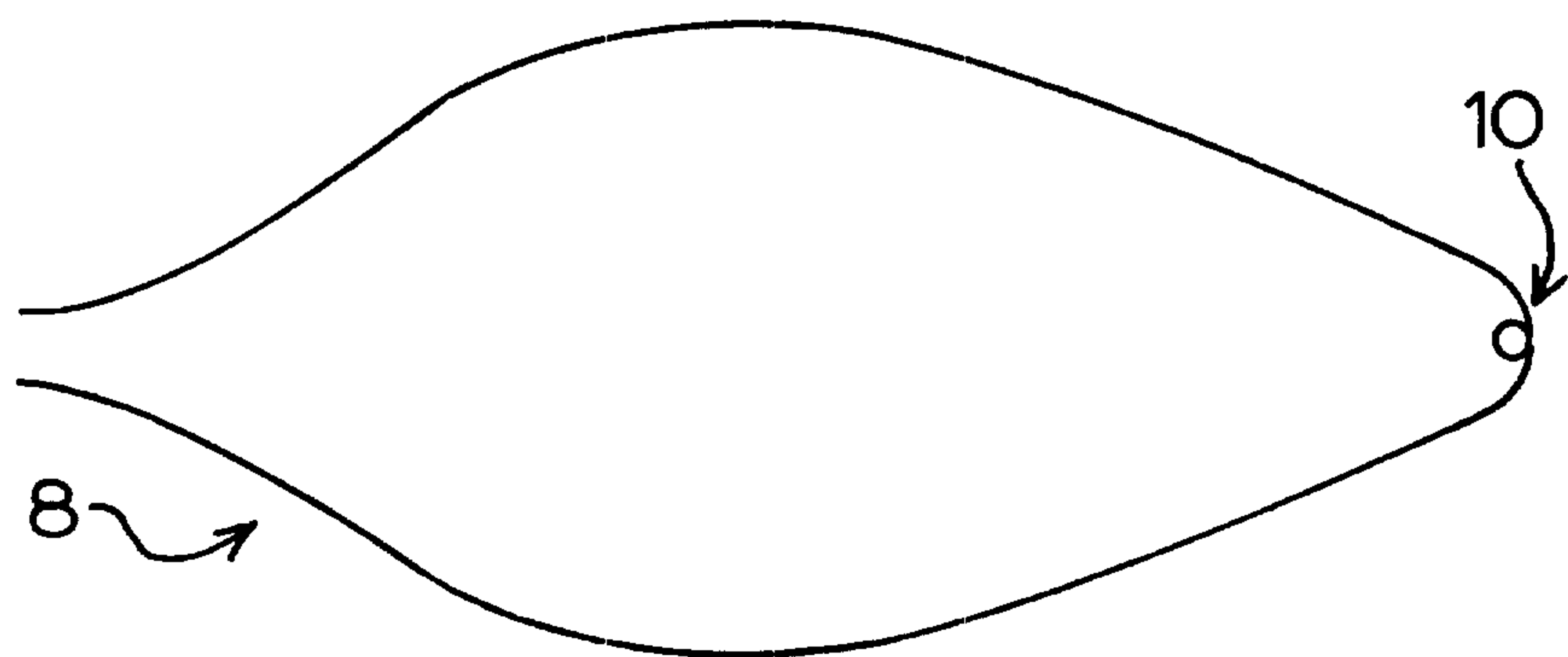


FIG. 3



Sine: M. Lunn

FIG. 4

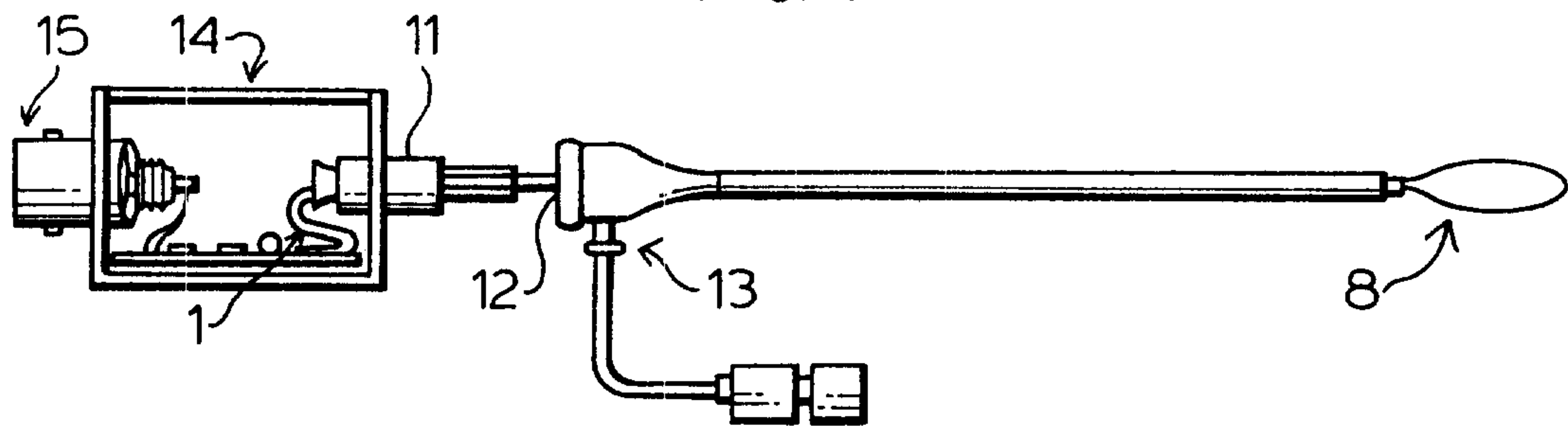


FIG. 5A

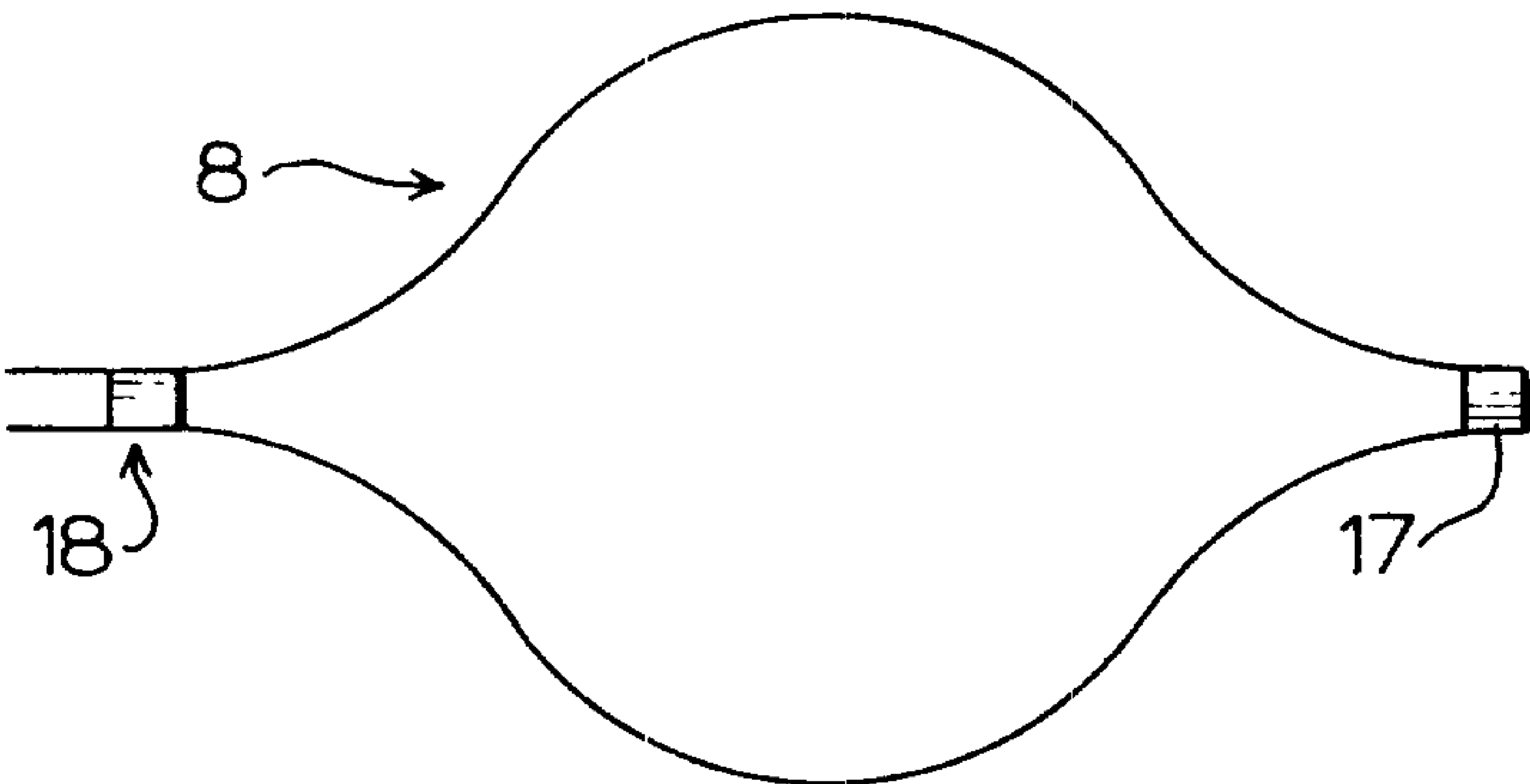


FIG. 5B

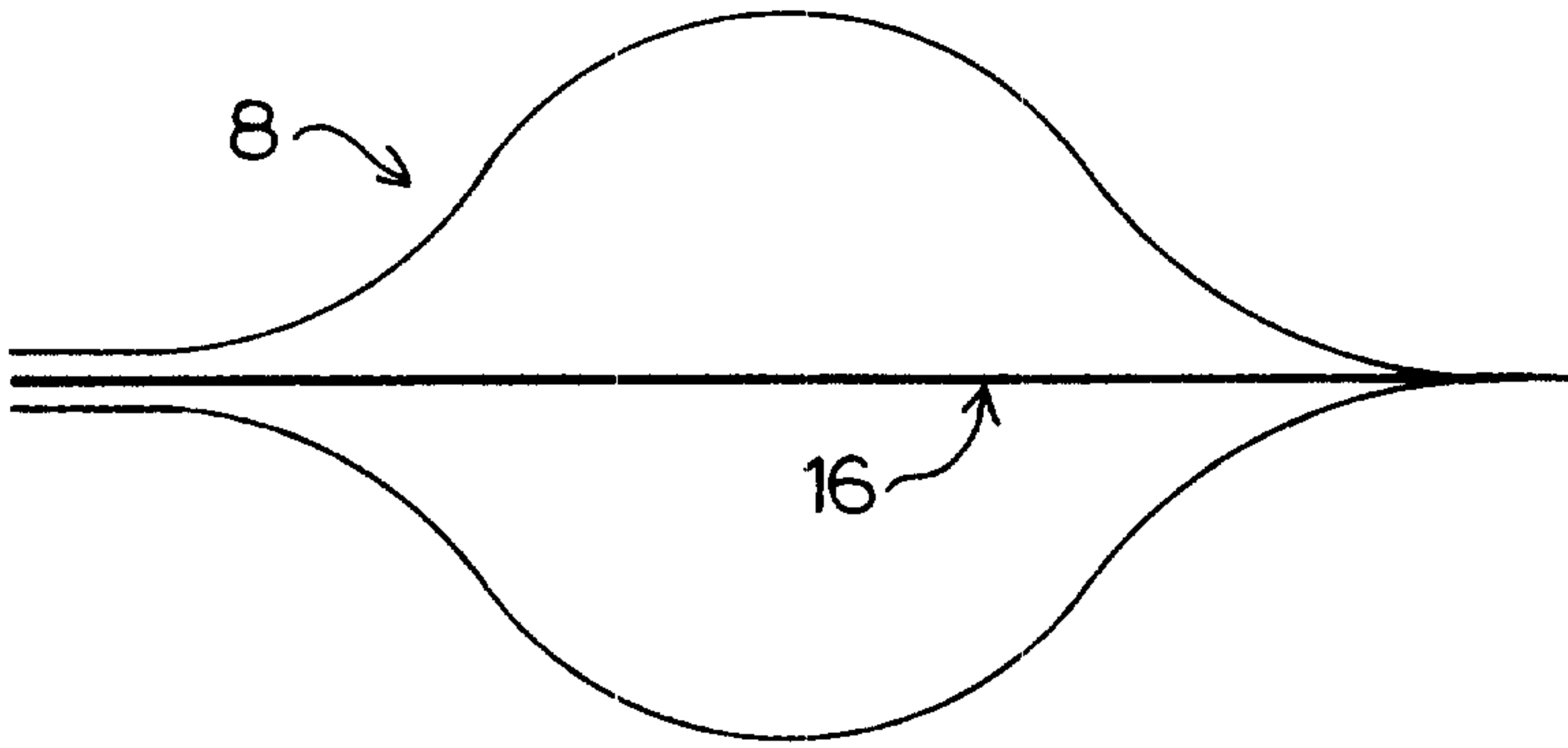


FIG. 5C

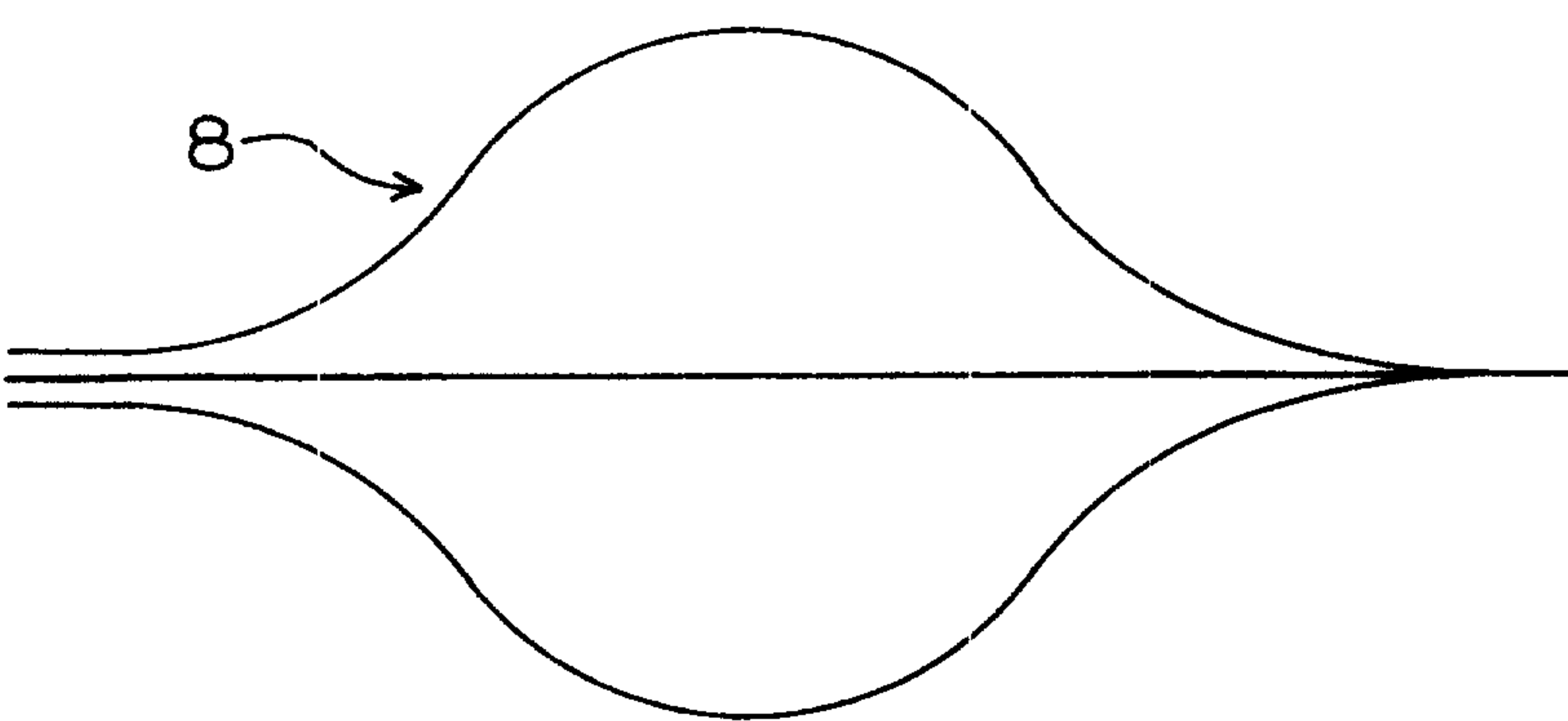
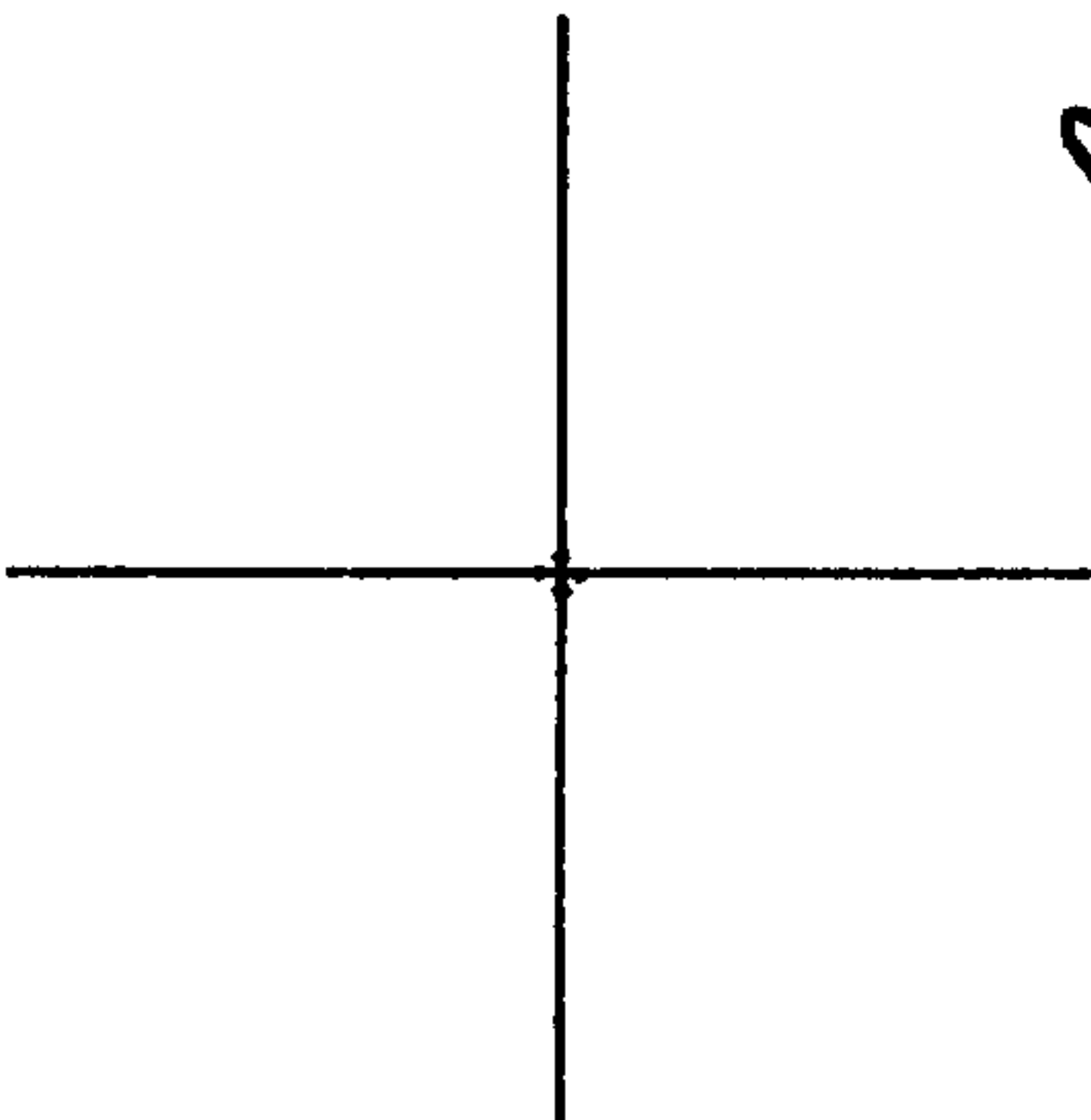


FIG. 5D



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