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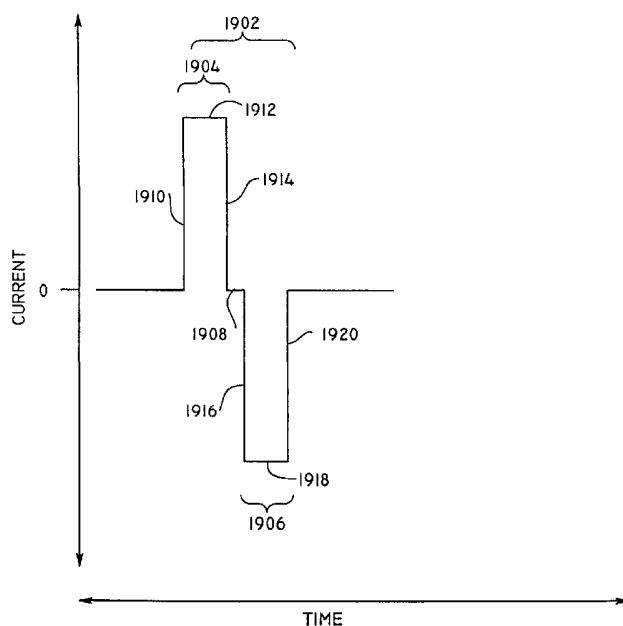
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(54) Title: CURRENT WAVEFORMS FOR ANTI-BRADYCARDIA PACING FOR A SUBCUTANEOUS IMPLANTABLE CARDIOVERTER-DEFIBRILLATOR



(57) Abstract: A power supply for an implantable cardioverter-defibrillator for subcutaneous positioning between the third rib and the twelfth rib and using a lead system that does not directly contact a patient's heart or reside in the intrathoracic blood vessels and for providing anti-bradycardia pacing energy to the heart, comprising a capacitor subsystem for storing the anti-bradycardia pacing energy for delivery to the patient's heart; and a battery subsystem electrically coupled to the capacitor subsystem for providing the anti-bradycardia pacing energy to the capacitor subsystem.



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**CURRENT WAVEFORMS FOR ANTI-BRADYCARDIA PACING FOR A
SUBCUTANEOUS IMPLANTABLE CARDIOVERTER-DEFIBRILLATOR**

5 CROSS-REFERENCE TO RELATED APPLICATIONS

 The present application is a continuation-in-part of U.S. patent application
entitled "SUBCUTANEOUS ONLY IMPLANTABLE CARDIOVERTER-
DEFIBRILLATOR AND OPTIONAL PACER," having Serial No. 09/663,607,
filed September 18, 2000, pending, and U.S. patent application entitled
10 "UNITARY SUBCUTANEOUS ONLY IMPLANTABLE CARDIOVERTER-
DEFIBRILLATOR AND OPTIONAL PACER," having Serial No. 09/663,606,
filed September 18, 2000, pending, of which both applications are assigned to the
assignee of the present application, and the disclosures of both applications are
hereby incorporated by reference.

15 In addition, the present application is filed concurrently herewith U.S.
patent application entitled "MONOPHASIC WAVEFORM FOR ANTI-
BRADYCARDIA PACING FOR A SUBCUTANEOUS IMPLANTABLE
CARDIOVERTER-DEFIBRILLATOR," U.S. patent application entitled
"MONOPHASIC WAVEFORM FOR ANTI-TACHYCARDIA PACING FOR A
20 SUBCUTANEOUS IMPLANTABLE CARDIOVERTER-DEFIBRILLATOR"
and U.S. patent application entitled "CURRENT WAVEFORMS FOR ANTI-
TACHYCARDIA PACING FOR A SUBCUTANEOUS IMPLANTABLE
CARDIOVERTER DEFIBRILLATOR," the disclosures of which applications are
hereby incorporated by reference.

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FIELD OF THE INVENTION

 The present invention relates to an apparatus and method for performing
electrical cardioversion/defibrillation and optional pacing of the heart via a totally
subcutaneous non-transvenous system.

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BACKGROUND OF THE INVENTION

 Defibrillation/cardioversion is a technique employed to counter arrhythmic
heart conditions including some tachycardias in the atria and/or ventricles.

Typically, electrodes are employed to stimulate the heart with electrical impulses or shocks, of a magnitude substantially greater than pulses used in cardiac pacing. Shocks used for defibrillation therapy can comprise a biphasic truncated exponential waveform. As for pacing, a constant current density is desired to
5 reduce or eliminate variability due to the electrode/tissue interface

Defibrillation/cardioversion systems include body implantable electrodes that are connected to a hermetically sealed container housing the electronics, battery supply and capacitors. The entire system is referred to as implantable cardioverter/defibrillators (ICDs). The electrodes used in ICDs can be in the form
10 of patches applied directly to epicardial tissue, or, more commonly, are on the distal regions of small cylindrical insulated catheters that typically enter the subclavian venous system, pass through the superior vena cava and, into one or more endocardial areas of the heart. Such electrode systems are called intravascular or transvenous electrodes. U.S. Pat. Nos. 4,603,705, 4,693,253,
15 4,944,300, 5,105,810, the disclosures of which are all incorporated herein by reference, disclose intravascular or transvenous electrodes, employed either alone, in combination with other intravascular or transvenous electrodes, or in combination with an epicardial patch or subcutaneous electrodes. Compliant epicardial defibrillator electrodes are disclosed in U.S. Pat. Nos. 4,567,900 and
20 5,618,287, the disclosures of which are incorporated herein by reference. A sensing epicardial electrode configuration is disclosed in U.S. Pat No. 5,476,503, the disclosure of which is incorporated herein by reference.

In addition to epicardial and transvenous electrodes, subcutaneous electrode systems have also been developed. For example, U.S. Patent Nos. 5,342,407 and
25 5,603,732, the disclosures of which are incorporated herein by reference, teach the use of a pulse monitor/generator surgically implanted into the abdomen and subcutaneous electrodes implanted in the thorax. This system is far more complicated to use than current ICD systems using transvenous lead systems together with an active can electrode and therefore it has no practical use. It has in
30 fact never been used because of the surgical difficulty of applying such a device (3 incisions), the impractical abdominal location of the generator and the electrically poor sensing and defibrillation aspects of such a system.

Recent efforts to improve the efficiency of ICDs have led manufacturers to produce ICDs which are small enough to be implanted in the pectoral region. In addition, advances in circuit design have enabled the housing of the ICD to form a subcutaneous electrode. Some examples of ICDs in which the housing of the ICD serves as an optional additional electrode are described in U.S. Pat. Nos. 5,133,353, 5,261,400, 5,620,477, and 5,658,321 the disclosures of which are incorporated herein by reference.

ICDs are now an established therapy for the management of life threatening cardiac rhythm disorders, primarily ventricular fibrillation (V-Fib). ICDs are very effective at treating V-Fib, but are therapies that still require significant surgery.

As ICD therapy becomes more prophylactic in nature and used in progressively less ill individuals, especially children at risk of cardiac arrest, the requirement of ICD therapy to use intravenous catheters and transvenous leads is an impediment to very long term management as most individuals will begin to develop complications related to lead system malfunction sometime in the 5-10 year time frame, often earlier. In addition, chronic transvenous lead systems, their reimplantation and removals, can damage major cardiovascular venous systems and the tricuspid valve, as well as result in life threatening perforations of the great vessels and heart. Consequently, use of transvenous lead systems, despite their many advantages, are not without their chronic patient management limitations in those with life expectancies of >5 years. The problem of lead complications is even greater in children where body growth can substantially alter transvenous lead function and lead to additional cardiovascular problems and revisions. Moreover, transvenous ICD systems also increase cost and require specialized interventional rooms and equipment as well as special skill for insertion. These systems are typically implanted by cardiac electrophysiologists who have had a great deal of extra training.

In addition to the background related to ICD therapy, the present invention requires a brief understanding of a related therapy, the automatic external defibrillator (AED). AEDs employ the use of cutaneous patch electrodes, rather than implantable lead systems, to effect defibrillation under the direction of a bystander user who treats the patient suffering from V-Fib with a portable device containing the necessary electronics and power supply that allows defibrillation.

AEDs can be nearly as effective as an ICD for defibrillation if applied to the victim of ventricular fibrillation promptly, i.e., within 2 to 3 minutes of the onset of the ventricular fibrillation.

5 AED therapy has great appeal as a tool for diminishing the risk of death in public venues such as in air flight. However, an AED must be used by another individual, not the person suffering from the potential fatal rhythm. It is more of a public health tool than a patient-specific tool like an ICD. Because >75% of cardiac arrests occur in the home, and over half occur in the bedroom, patients at risk of cardiac arrest are often alone or asleep and can not be helped in time with
10 an AED. Moreover, its success depends to a reasonable degree on an acceptable level of skill and calm by the bystander user.

What is needed therefore, especially for children and for prophylactic long term use for those at risk of cardiac arrest, is a combination of the two forms of therapy which would provide prompt and near-certain defibrillation, like an ICD,
15 but without the long-term adverse sequelae of a transvenous lead system while simultaneously using most of the simpler and lower cost technology of an AED. What is also needed is a cardioverter/defibrillator that is of simple design and can be comfortably implanted in a patient for many years.

20 SUMMARY OF THE INVENTION

A power supply for an implantable cardioverter-defibrillator for subcutaneous positioning between the third rib and the twelfth rib and using a lead system that does not directly contact a patient's heart or reside in the intrathoracic blood vessels and for providing anti-bradycardia pacing energy to the heart,
25 comprising a capacitor subsystem for storing the anti-bradycardia pacing energy for delivery to the patient's heart; and a battery subsystem electrically coupled to the capacitor subsystem for providing the anti-bradycardia pacing energy to the capacitor subsystem.

30 BRIEF DESCRIPTION OF THE DRAWINGS

For a better understanding of the invention, reference is now made to the drawings where like numerals represent similar objects throughout the figures where:

FIG. 1 is a schematic view of a Subcutaneous ICD (S-ICD) of the present invention;

FIG. 2 is a schematic view of an alternate embodiment of a subcutaneous electrode of the present invention;

5 FIG. 3 is a schematic view of an alternate embodiment of a subcutaneous electrode of the present invention;

FIG. 4 is a schematic view of the S-ICD and lead of FIG. 1 subcutaneously implanted in the thorax of a patient;

10 FIG. 5 is a schematic view of the S-ICD and lead of FIG. 2 subcutaneously implanted in an alternate location within the thorax of a patient;

FIG. 6 is a schematic view of the S-ICD and lead of FIG. 3 subcutaneously implanted in the thorax of a patient;

15 FIG. 7 is a schematic view of the method of making a subcutaneous path from the preferred incision and housing implantation point to a termination point for locating a subcutaneous electrode of the present invention;

FIG. 8 is a schematic view of an introducer set for performing the method of lead insertion of any of the described embodiments;

20 FIG. 9 is a schematic view of an alternative S-ICD of the present invention illustrating a lead subcutaneously and serpigiously implanted in the thorax of a patient for use particularly in children;

FIG. 10 is a schematic view of an alternate embodiment of an S-ICD of the present invention;

FIG. 11 is a schematic view of the S-ICD of FIG. 10 subcutaneously implanted in the thorax of a patient;

25 FIG. 12 is a schematic view of yet a further embodiment where the canister of the S-ICD of the present invention is shaped to be particularly useful in placing subcutaneously adjacent and parallel to a rib of a patient;

30 FIG. 13 is a schematic of a different embodiment where the canister of the S-ICD of the present invention is shaped to be particularly useful in placing subcutaneously adjacent and parallel to a rib of a patient;

FIG. 14 is a schematic view of a Unitary Subcutaneous ICD (US-ICD) of the present invention;

FIG. 15 is a schematic view of the US-ICD subcutaneously implanted in the thorax of a patient;

FIG. 16 is a schematic view of the method of making a subcutaneous path from the preferred incision for implanting the US-ICD;

5 FIG. 17 is a schematic view of an introducer for performing the method of US-ICD implantation; and

FIG. 18 is an exploded schematic view of an alternate embodiment of the present invention with a plug-in portion that contains operational circuitry and means for generating cardioversion/defibrillation shock waves;

10 Fig. 19 is a graph that shows an example of a biphasic waveform for use in anti-bradycardia pacing in an embodiment of the present invention; and

Fig. 20 is a graph that shows an example of a monophasic waveform for use in anti-bradycardia pacing in an embodiment of the present invention.

15 DETAILED DESCRIPTION

Turning now to FIG. 1, the S-ICD of the present invention is illustrated. The S-ICD consists of an electrically active canister 11 and a subcutaneous electrode 13 attached to the canister. The canister has an electrically active surface 15 that is electrically insulated from the electrode connector block 17 and the canister housing 16 via insulating area 14. The canister can be similar to numerous electrically active canisters commercially available in that the canister will contain a battery supply, capacitor and operational circuitry. Alternatively, the canister can be thin and elongated to conform to the intercostal space. The circuitry will be able to monitor cardiac rhythms for tachycardia and fibrillation, and if detected, will initiate charging the capacitor and then delivering cardioversion /defibrillation energy through the active surface of the housing and to the subcutaneous electrode. Examples of such circuitry are described in U.S. Patent Nos. 4,693,253 and 5,105,810, the entire disclosures of which are herein incorporated by reference. The canister circuitry can provide cardioversion/ defibrillation energy in different types of waveforms. In one embodiment, a 100 uF biphasic waveform is used of approximately 10-20 ms total duration and with the initial phase containing approximately 2/3 of the energy, however, any type of waveform can be utilized

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such as monophasic, biphasic, multiphasic or alternative waveforms as is known in the art.

In addition to providing cardioversion/ defibrillation energy, the circuitry can also provide transthoracic cardiac pacing energy. The optional circuitry will be able to monitor the heart for bradycardia and/or tachycardia rhythms. Once a bradycardia or tachycardia rhythm is detected, the circuitry can then deliver appropriate pacing energy at appropriate intervals through the active surface and the subcutaneous electrode. Pacing stimuli can be biphasic in one embodiment and similar in pulse amplitude to that used for conventional transthoracic pacing.

This same circuitry can also be used to deliver low amplitude shocks on the T-wave for induction of ventricular fibrillation for testing S-ICD performance in treating V-Fib as is described in U.S. Patent No. 5,129,392, the entire disclosure of which is hereby incorporated by reference. Also the circuitry can be provided with rapid induction of ventricular fibrillation or ventricular tachycardia using rapid ventricular pacing. Another optional way for inducing ventricular fibrillation would be to provide a continuous low voltage, i.e., about 3 volts, across the heart during the entire cardiac cycle.

Another optional aspect of the present invention is that the operational circuitry can detect the presence of atrial fibrillation as described in Olson, W. et al. "Onset And Stability For Ventricular Tachyarrhythmia Detection in an Implantable Cardioverter and Defibrillator," Computers in Cardiology (1986) pp. 167-170. Detection can be provided via R-R Cycle length instability detection algorithms. Once atrial fibrillation has been detected, the operational circuitry will then provide QRS synchronized atrial defibrillation/cardioversion using the same shock energy and waveshape characteristics used for ventricular defibrillation/ cardioversion.

The sensing circuitry will utilize the electronic signals generated from the heart and will primarily detect QRS waves. In one embodiment, the circuitry will be programmed to detect only ventricular tachycardias or fibrillations. The detection circuitry will utilize in its most direct form, a rate detection algorithm that triggers charging of the capacitor once the ventricular rate exceeds some predetermined level for a fixed period of time: for example, if the ventricular rate exceeds 240 bpm on average for more than 4 seconds. Once the capacitor is

charged, a confirmatory rhythm check would ensure that the rate persists for at least another 1 second before discharge. Similarly, termination algorithms could be instituted that ensure that a rhythm less than 240 bpm persisting for at least 4 seconds before the capacitor charge is drained to an internal resistor. Detection, confirmation and termination algorithms as are described above and in the art can be modulated to increase sensitivity and specificity by examining QRS beat-to-beat uniformity, QRS signal frequency content, R-R interval stability data, and signal amplitude characteristics all or part of which can be used to increase or decrease both sensitivity and specificity of S-ICD arrhythmia detection function.

In addition to use of the sense circuitry for detection of V-Fib or V-Tach by examining the QRS waves, the sense circuitry can check for the presence or the absence of respiration. The respiration rate can be detected by monitoring the impedance across the thorax using subthreshold currents delivered across the active can and the high voltage subcutaneous lead electrode and monitoring the frequency in undulation in the waveform that results from the undulations of transthoracic impedance during the respiratory cycle. If there is no undulation, then the patient is not respiring and this lack of respiration can be used to confirm the QRS findings of cardiac arrest. The same technique can be used to provide information about the respiratory rate or estimate cardiac output as described in U.S. Patent Nos. 6,095,987, 5,423,326, 4,450,527, the entire disclosures of which are incorporated herein by reference.

The canister of the present invention can be made out of titanium alloy or other presently preferred electrically active canister designs. However, it is contemplated that a malleable canister that can conform to the curvature of the patient's chest will be preferred. In this way the patient can have a comfortable canister that conforms to the shape of the patient's rib cage. Examples of conforming canisters are provided in U.S. Patent No. 5,645,586, the entire disclosure of which is herein incorporated by reference. Therefore, the canister can be made out of numerous materials such as medical grade plastics, metals, and alloys. In the preferred embodiment, the canister is smaller than 60 cc volume having a weight of less than 100 gms for long term wearability, especially in children. The canister and the lead of the S-ICD can also use fractal or wrinkled surfaces to increase surface area to improve defibrillation capability. Because of

the primary prevention role of the therapy and the likely need to reach energies over 40 Joules, a feature of one embodiment is that the charge time for the therapy, is intentionally left relatively long to allow capacitor charging within the limitations of device size. Examples of small ICD housings are disclosed in U.S. Patents Nos. 5,597,956 and 5,405,363, the entire disclosures of which are herein incorporated by reference.

Different subcutaneous electrodes 13 of the present invention are illustrated in FIGS. 1-3. Turning to FIG. 1, the lead 21 for the subcutaneous electrode is preferably composed of silicone or polyurethane insulation. The electrode is connected to the canister at its proximal end via connection port 19 which is located on an electrically insulated area 17 of the canister. The electrode illustrated is a composite electrode with three different electrodes attached to the lead. In the embodiment illustrated, an optional anchor segment 52 is attached at the most distal end of the subcutaneous electrode for anchoring the electrode into soft tissue such that the electrode does not dislodge after implantation.

The most distal electrode on the composite subcutaneous electrode is a coil electrode 27 that is used for delivering the high voltage cardioversion/defibrillation energy across the heart. The coil cardioversion/defibrillation electrode is about 5-10 cm in length. Proximal to the coil electrode are two sense electrodes, a first sense electrode 25 is located proximally to the coil electrode and a second sense electrode 23 is located proximally to the first sense electrode. The sense electrodes are spaced far enough apart to be able to have good QRS detection. This spacing can range from 1 to 10 cm with 4 cm being presently preferred. The electrodes may or may not be circumferential with the preferred embodiment. Having the electrodes non-circumferential and positioned outward, toward the skin surface, is a means to minimize muscle artifact and enhance QRS signal quality. The sensing electrodes are electrically isolated from the cardioversion/defibrillation electrode via insulating areas 29. Similar types of cardioversion/defibrillation electrodes are currently commercially available in a transvenous configuration. For example, U.S. Patent No. 5,534,022, the entire disclosure of which is herein incorporated by reference, discloses a composite electrode with a coil cardioversion/defibrillation electrode and sense electrodes. Modifications to this arrangement is contemplated within the scope of the

invention. One such modification is illustrated in FIG. 2 where the two sensing electrodes 25 and 23 are non-circumferential sensing electrodes and one is located at the distal end, the other is located proximal thereto with the coil electrode located in between the two sensing electrodes. In this embodiment the sense electrodes are spaced about 6 to about 12 cm apart depending on the length of the coil electrode used. FIG. 3 illustrates yet a further embodiment where the two sensing electrodes are located at the distal end to the composite electrode with the coil electrode located proximally thereto. Other possibilities exist and are contemplated within the present invention. For example, having only one sensing electrode, either proximal or distal to the coil cardioversion/ defibrillation electrode with the coil serving as both a sensing electrode and a cardioversion/defibrillation electrode.

It is also contemplated within the scope of the invention that the sensing of QRS waves (and transthoracic impedance) can be carried out via sense electrodes on the canister housing or in combination with the cardioversion/defibrillation coil electrode and/or the subcutaneous lead sensing electrode(s). In this way, sensing could be performed via the one coil electrode located on the subcutaneous electrode and the active surface on the canister housing. Another possibility would be to have only one sense electrode located on the subcutaneous electrode and the sensing would be performed by that one electrode and either the coil electrode on the subcutaneous electrode or by the active surface of the canister. The use of sensing electrodes on the canister would eliminate the need for sensing electrodes on the subcutaneous electrode. It is also contemplated that the subcutaneous electrode would be provided with at least one sense electrode, the canister with at least one sense electrode, and if multiple sense electrodes are used on either the subcutaneous electrode and/or the canister, that the best QRS wave detection combination will be identified when the S-ICD is implanted and this combination can be selected, activating the best sensing arrangement from all the existing sensing possibilities. Turning again to FIG. 2, two sensing electrodes 26 and 28 are located on the electrically active surface 15 with electrical insulator rings 30 placed between the sense electrodes and the active surface. These canister sense electrodes could be switched off and electrically insulated during and shortly after defibrillation/ cardioversion shock delivery. The canister sense electrodes may

also be placed on the electrically inactive surface of the canister. In the embodiment of FIG. 2, there are actually four sensing electrodes, two on the subcutaneous lead and two on the canister. In the preferred embodiment, the ability to change which electrodes are used for sensing would be a programmable feature of the S-ICD to adapt to changes in the patient physiology and size (in the case of children) over time. The programming could be done via the use of physical switches on the canister, or as presently preferred, via the use of a programming wand or via a wireless connection to program the circuitry within the canister.

The canister could be employed as either a cathode or an anode of the S-ICD cardioversion/defibrillation system. If the canister is the cathode, then the subcutaneous coil electrode would be the anode. Likewise, if the canister is the anode, then the subcutaneous electrode would be the cathode.

The active canister housing will provide energy and voltage intermediate to that available with ICDs and most AEDs. The typical maximum voltage necessary for ICDs using most biphasic waveforms is approximately 750 Volts with an associated maximum energy of approximately 40 Joules. The typical maximum voltage necessary for AEDs is approximately 2000-5000 Volts with an associated maximum energy of approximately 200-360 Joules depending upon the model and waveform used. The S-ICD and the US-ICD of the present invention uses maximum voltages in the range of about 50 to about 3500 Volts and is associated with energies of about .5 to about 350 Joules. The capacitance of the devices can range from about 25 to about 200 micro farads.

In another embodiment, the S-ICD and US-ICD devices provide energy with a pulse width of approximately one millisecond to approximately 40 milliseconds. The devices can provide pacing current of approximately one milliamp to approximately 250 milliamps.

The sense circuitry contained within the canister is highly sensitive and specific for the presence or absence of life threatening ventricular arrhythmias. Features of the detection algorithm are programmable and the algorithm is focused on the detection of V-FIB and high rate V-TACH (>240 bpm). Although the S-ICD of the present invention may rarely be used for an actual life-threatening event, the simplicity of design and implementation allows it to be employed in

large populations of patients at modest risk with modest cost by non-cardiac electrophysiologists. Consequently, the S-ICD of the present invention focuses mostly on the detection and therapy of the most malignant rhythm disorders. As part of the detection algorithm's applicability to children, the upper rate range is programmable upward for use in children, known to have rapid supraventricular tachycardias and more rapid ventricular fibrillation. Energy levels also are programmable downward in order to allow treatment of neonates and infants.

Turning now to FIG. 4, the optimal subcutaneous placement of the S-ICD of the present invention is illustrated. As would be evidence to a person skilled in the art, the actual location of the S-ICD is in a subcutaneous space that is developed during the implantation process. The heart is not exposed during this process and the heart is schematically illustrated in the figures only for help in understanding where the canister and coil electrode are three dimensionally located in the left mid-clavicular line approximately at the level of the inframammary crease at approximately the 5th rib. The lead 21 of the subcutaneous electrode traverses in a subcutaneous path around the thorax terminating with its distal electrode end at the posterior axillary line ideally just lateral to the left scapula. This way the canister and subcutaneous cardioversion/defibrillation electrode provide a reasonably good pathway for current delivery to the majority of the ventricular myocardium.

FIG. 5 illustrates a different placement of the present invention. The S-ICD canister with the active housing is located in the left posterior axillary line approximately lateral to the tip of the inferior portion of the scapula. This location is especially useful in children. The lead 21 of the subcutaneous electrode traverses in a subcutaneous path around the thorax terminating with its distal electrode end at the anterior precordial region, ideally in the inframammary crease. FIG. 6 illustrates the embodiment of FIG. 1 subcutaneously implanted in the thorax with the proximal sense electrodes 23 and 25 located at approximately the left axillary line with the cardioversion/defibrillation electrode just lateral to the tip of the inferior portion of the scapula.

FIG. 7 schematically illustrates the method for implanting the S-ICD of the present invention. An incision 31 is made in the left anterior axillary line approximately at the level of the cardiac apex. This incision location is distinct

from that chosen for S-ICD placement and is selected specifically to allow both canister location more medially in the left inframammary crease and lead positioning more posteriorly via the introducer set (described below) around to the left posterior axillary line lateral to the left scapula. That said, the incision can be
5 anywhere on the thorax deemed reasonably by the implanting physician although in the preferred embodiment, the S-ICD of the present invention will be applied in this region. A subcutaneous pathway 33 is then created medially to the inframmary crease for the canister and posteriorly to the left posterior axillary line lateral to the left scapula for the lead.

10 The S-ICD canister 11 is then placed subcutaneously at the location of the incision or medially at the subcutaneous region at the left inframmary crease. The subcutaneous electrode 13 is placed with a specially designed curved introducer set 40 (see FIG. 8). The introducer set comprises a curved trocar 42 and a stiff curved peel away sheath 44. The peel away sheath is curved to allow for placement
15 around the rib cage of the patient in the subcutaneous space created by the trocar. The sheath has to be stiff enough to allow for the placement of the electrodes without the sheath collapsing or bending. Preferably the sheath is made out of a biocompatible plastic material and is perforated along its axial length to allow for it to split apart into two sections. The trocar has a proximal handle 41 and a curved
20 shaft 43. The distal end 45 of the trocar is tapered to allow for dissection of a subcutaneous path 33 in the patient. Preferably, the trocar is cannulated having a central Lumen 46 and terminating in an opening 48 at the distal end. Local anesthetic such as lidocaine can be delivered, if necessary, through the lumen or through a curved and elongated needle designed to anesthetize the path to be used
25 for trocar insertion should general anesthesia not be employed. The curved peel away sheath 44 has a proximal pull tab 49 for breaking the sheath into two halves along its axial shaft 47. The sheath is placed over a guidewire inserted through the trocar after the subcutaneous path has been created. The subcutaneous pathway is then developed until it terminates subcutaneously at a location that, if a straight
30 line were drawn from the canister location to the path termination point the line would intersect a substantial portion of the left ventricular mass of the patient. The guidewire is then removed leaving the peel away sheath. The subcutaneous lead system is then inserted through the sheath until it is in the proper location. Once

the subcutaneous lead system is in the proper location, the sheath is split in half using the pull tab 49 and removed. If more than one subcutaneous electrode is being used, a new curved peel away sheath can be used for each subcutaneous electrode.

5 The S-ICD will have prophylactic use in adults where chronic transvenous/epicardial ICD lead systems pose excessive risk or have already resulted in difficulty, such as sepsis or lead fractures. It is also contemplated that a major use of the S-ICD system of the present invention will be for prophylactic use in children who are at risk for having fatal arrhythmias, where chronic transvenous
10 lead systems pose significant management problems. Additionally, with the use of standard transvenous ICDs in children, problems develop during patient growth in that the lead system does not accommodate the growth. FIG. 9 illustrates the placement of the S-ICD subcutaneous lead system such that the problem that growth presents to the lead system is overcome. The distal end of the
15 subcutaneous electrode is placed in the same location as described above providing a good location for the coil cardioversion/defibrillation electrode 27 and the sensing electrodes 23 and 25. The insulated lead 21, however is no longer placed in a taught configuration. Instead, the lead is serpiginously placed with a specially designed introducer trocar and sheath such that it has numerous waves or bends.
20 As the child grows, the waves or bends will straighten out lengthening the lead system while maintaining proper electrode placement. Although it is expected that fibrous scarring especially around the defibrillation coil will help anchor it into position to maintain its posterior position during growth, a lead system with a distal tine or screw electrode anchoring system 52 can also be incorporated into the
25 distal tip of the lead to facilitate lead stability (see FIG. 1). Other anchoring systems can also be used such as hooks, sutures, or the like.

FIGS. 10 and 11 illustrate another embodiment of the present S-ICD invention. In this embodiment there are two subcutaneous electrodes 13 and 13' of opposite polarity to the canister. The additional subcutaneous electrode 13' is
30 essentially identical to the previously described electrode. In this embodiment the cardioversion/defibrillation energy is delivered between the active surface of the canister and the two coil electrodes 27 and 27'. Additionally, provided in the canister is means for selecting the optimum sensing arrangement between the four

sense electrodes 23, 23', 25, and 25'. The two electrodes are subcutaneously placed on the same side of the heart. As illustrated in FIG. 6, one subcutaneous electrode 13 is placed inferiorly and the other electrode 13' is placed superiorly. It is also contemplated with this dual subcutaneous electrode system that the canister and one subcutaneous electrode are the same polarity and the other subcutaneous electrode is the opposite polarity.

Turning now to FIGS. 12 and 13, further embodiments are illustrated where the canister 11 of the S-ICD of the present invention is shaped to be particularly useful in placing subcutaneously adjacent and parallel to a rib of a patient. The canister is long, thin, and curved to conform to the shape of the patient's rib. In the embodiment illustrated in FIG. 12, the canister has a diameter ranging from about 0.5 cm to about 2 cm without 1 cm being presently preferred. Alternatively, instead of having a circular cross sectional area, the canister could have a rectangular or square cross sectional area as illustrated in FIG. 13 without falling outside of the scope of the present invention. The length of the canister can vary depending on the size of the patient's thorax. In an embodiment, the canister is about 5 cm to about 40 cm long. The canister is curved to conform to the curvature of the ribs of the thorax. The radius of the curvature will vary depending on the size of the patient, with smaller radiuses for smaller patients and larger radiuses for larger patients. The radius of the curvature can range from about 5 cm to about 35 cm depending on the size of the patient. Additionally, the radius of the curvature need not be uniform throughout the canister such that it can be shaped closer to the shape of the ribs. The canister has an active surface, 15 that is located on the interior (concave) portion of the curvature and an inactive surface 16 that is located on the exterior (convex) portion of the curvature. The leads of these embodiments, which are not illustrated except for the attachment port 19 and the proximal end of the lead 21, can be any of the leads previously described above, with the lead illustrated in FIG. 1 being presently preferred.

The circuitry of this canister is similar to the circuitry described above. Additionally, the canister can optionally have at least one sense electrode located on either the active surface of the inactive surface and the circuitry within the canister can be programmable as described above to allow for the selection of the best sense electrodes. It is presently preferred that the canister have two sense

electrodes 26 and 28 located on the inactive surface of the canisters as illustrated, where the electrodes are spaced from about 1 to about 10 cm apart with a spacing of about 3 cm being presently preferred. However, the sense electrodes can be located on the active surface as described above.

5 It is envisioned that the embodiment of FIG. 12 will be subcutaneously implanted adjacent and parallel to the left anterior 5th rib, either between the 4th and 5th ribs or between the 5th and 6th ribs. However other locations can be used.

Another component of the S-ICD of the present invention is a cutaneous test electrode system designed to simulate the subcutaneous high voltage shock electrode system as well as the QRS cardiac rhythm detection system. This test electrode system is comprised of a cutaneous patch electrode of similar surface area and impedance to that of the S-ICD canister itself together with a cutaneous strip electrode comprising a defibrillation strip as well as two button electrodes for sensing of the QRS. Several cutaneous strip electrodes are available to allow for testing various bipole spacings to optimize signal detection comparable to the implantable system.

15 Figures 14 to 18 depict particular US-ICD embodiments of the present invention. The various sensing, shocking and pacing circuitry, described in detail above with respect to the S-ICD embodiments, may additionally be incorporated into the following US-ICD embodiments. Furthermore, particular aspects of any individual S-ICD embodiment discussed above, may be incorporated, in whole or in part, into the US-ICD embodiments depicted in the following figures.

Turning now to Fig. 14, the US-ICD of the present invention is illustrated. The US-ICD consists of a curved housing 1211 with a first and second end. The first end 1413 is thicker than the second end 1215. This thicker area houses a battery supply, capacitor and operational circuitry for the US-ICD. The circuitry will be able to monitor cardiac rhythms for tachycardia and fibrillation, and if detected, will initiate charging the capacitor and then delivering cardioversion/defibrillation energy through the two cardioversion/defibrillating electrodes 1417 and 1219 located on the outer surface of the two ends of the housing. The circuitry can provide cardioversion/defibrillation energy in different types of waveforms. In one embodiment, a 100 uF biphasic waveform is used of approximately 10-20 ms total duration and with the initial phase containing

approximately 2/3 of the energy, however, any type of waveform can be utilized such as monophasic, biphasic, multiphasic or alternative waveforms as is known in the art.

5 The housing of the present invention can be made out of titanium alloy or other presently preferred ICD designs. It is contemplated that the housing is also made out of biocompatible plastic materials that electronically insulate the electrodes from each other. However, it is contemplated that a malleable canister that can conform to the curvature of the patient's chest will be preferred. In this way the patient can have a comfortable canister that conforms to the unique shape of the patient's rib cage. Examples of conforming ICD housings are provided in 10 U.S. Patent No. 5,645,586, the entire disclosure of which is herein incorporated by reference. In the preferred embodiment, the housing is curved in the shape of a 5th rib of a person. Because there are many different sizes of people, the housing will come in different incremental sizes to allow a good match between the size of the rib cage and the size of the US-ICD. The length of the US-ICD will range from 15 about 15 to about 50 cm. Because of the primary preventative role of the therapy and the need to reach energies over 40 Joules, a feature of the preferred embodiment is that the charge time for the therapy, intentionally be relatively long to allow capacitor charging within the limitations of device size.

20 The thick end of the housing is currently needed to allow for the placement of the battery supply, operational circuitry, and capacitors. It is contemplated that the thick end will be about 0.5 cm to about 2 cm wide with about 1 cm being presently preferred. As microtechnology advances, the thickness of the housing will become smaller.

25 The two cardioversion/defibrillation electrodes on the housing are used for delivering the high voltage cardioversion/defibrillation energy across the heart. In the preferred embodiment, the cardioversion/defibrillation electrodes are coil electrodes, however, other cardioversion/defibrillation electrodes could be used such as having electrically isolated active surfaces or platinum alloy electrodes. 30 The coil cardioversion/defibrillation electrodes are about 5-10 cm in length. Located on the housing between the two cardioversion/defibrillation electrodes are two sense electrodes 1425 and 1427. The sense electrodes are spaced far enough apart to be able to have good QRS detection. This spacing can range from 1 to 10

cm with 4 cm being presently preferred. The electrodes may or may not be circumferential with the preferred embodiment. Having the electrodes non-circumferential and positioned outward, toward the skin surface, is a means to minimize muscle artifact and enhance QRS signal quality. The sensing electrodes are electrically isolated from the cardioversion/defibrillation electrode via insulating areas 1423. Analogous types of cardioversion/defibrillation electrodes are currently commercially available in a transvenous configuration. For example, U.S. Patent No. 5,534,022, the entire disclosure of which is herein incorporated by reference, discloses a composite electrode with a coil cardioversion/defibrillation electrode and sense electrodes. Modifications to this arrangement is contemplated within the scope of the invention. One such modification is to have the sense electrodes at the two ends of the housing and have the cardioversion/defibrillation electrodes located in between the sense electrodes. Another modification is to have three or more sense electrodes spaced throughout the housing and allow for the selection of the two best sensing electrodes. If three or more sensing electrodes are used, then the ability to change which electrodes are used for sensing would be a programmable feature of the US-ICD to adapt to changes in the patient physiology and size over time. The programming could be done via the use of physical switches on the canister, or as presently preferred, via the use of a programming wand or via a wireless connection to program the circuitry within the canister.

Turning now to Fig. 15, the optimal subcutaneous placement of the US-ICD of the present invention is illustrated. As would be evident to a person skilled in the art, the actual location of the US-ICD is in a subcutaneous space that is developed during the implantation process. The heart is not exposed during this process and the heart is schematically illustrated in the figures only for help in understanding where the device and its various electrodes are three dimensionally located in the thorax of the patient. The US-ICD is located between the left mid-clavicular line approximately at the level of the inframammary crease at approximately the 5th rib and the posterior axillary line, ideally just lateral to the left scapula. This way the US-ICD provides a reasonably good pathway for current delivery to the majority of the ventricular myocardium.

Fig. 16 schematically illustrates the method for implanting the US-ICD of the present invention. An incision 1631 is made in the left anterior axillary line approximately at the level of the cardiac apex. A subcutaneous pathway is then created that extends posteriorly to allow placement of the US-ICD. The incision can be anywhere on the thorax deemed reasonable by the implanting physician although in the preferred embodiment, the US-ICD of the present invention will be applied in this region. The subcutaneous pathway is created medially to the inframammary crease and extends posteriorly to the left posterior axillary line. The pathway is developed with a specially designed curved introducer 1742 (see Fig. 17). The trocar has a proximal handle 1641 and a curved shaft 1643. The distal end 1745 of the trocar is tapered to allow for dissection of a subcutaneous path in the patient. Preferably, the trocar is cannulated having a central lumen 1746 and terminating in an opening 1748 at the distal end. Local anesthetic such as lidocaine can be delivered, if necessary, through the lumen or through a curved and elongated needle designed to anesthetize the path to be used for trocar insertion should general anesthesia not be employed. Once the subcutaneous pathway is developed, the US-ICD is implanted in the subcutaneous space, the skin incision is closed using standard techniques.

As described previously, the US-ICDs of the present invention vary in length and curvature. The US-ICDs are provided in incremental sizes for subcutaneous implantation in different sized patients. Turning now to Fig. 18, a different embodiment is schematically illustrated in exploded view which provides different sized US-ICDs that are easier to manufacture. The different sized US-ICDs will all have the same sized and shaped thick end 1413. The thick end is hollow inside allowing for the insertion of a core operational member 1853. The core member comprises a housing 1857 which contains the battery supply, capacitor and operational circuitry for the US-ICD. The proximal end of the core member has a plurality of electronic plug connectors. Plug connectors 1861 and 1863 are electronically connected to the sense electrodes via pressure fit connectors (not illustrated) inside the thick end which are standard in the art. Plug connectors 1865 and 1867 are also electronically connected to the cardioverter/defibrillator electrodes via pressure fit connectors inside the thick end. The distal end of the core member comprises an end cap 1855, and a ribbed fitting

1859 which creates a water-tight seal when the core member is inserted into opening 1851 of the thick end of the US-ICD.

The S-ICD and US-ICD, in alternative embodiments, have the ability to detect and treat atrial rhythm disorders, including atrial fibrillation. The S-ICD and US-ICD have two or more electrodes that provide a far-field view of cardiac electrical activity that includes the ability to record the P-wave of the electrocardiogram as well as the QRS. One can detect the onset and offset of atrial fibrillation by referencing to the P-wave recorded during normal sinus rhythm and monitoring for its change in rate, morphology, amplitude and frequency content. For example, a well-defined P-wave that abruptly disappeared and was replaced by a low-amplitude, variable morphology signal would be a strong indication of the absence of sinus rhythm and the onset of atrial fibrillation. In an alternative embodiment of a detection algorithm, the ventricular detection rate could be monitored for stability of the R-R coupling interval. In the examination of the R-R interval sequence, atrial fibrillation can be recognized by providing a near constant irregularly irregular coupling interval on a beat-by-beat basis. A R-R interval plot during AF appears "cloudlike" in appearance when several hundred or thousands of R-R intervals are plotted over time when compared to sinus rhythm or other supraventricular arrhythmias. Moreover, a distinguishing feature compared to other rhythms that are irregularly irregular, is that the QRS morphology is similar on a beat-by-beat basis despite the irregularity in the R-R coupling interval. This is a distinguishing feature of atrial fibrillation compared to ventricular fibrillation where the QRS morphology varies on a beat-by-beat basis. In yet another embodiment, atrial fibrillation may be detected by seeking to compare the timing and amplitude relationship of the detected P-wave of the electrocardiogram to the detected QRS (R-wave) of the electrocardiogram. Normal sinus rhythm has a fixed relationship that can be placed into a template matching algorithm that can be used as a reference point should the relationship change.

In other aspects of the atrial fibrillation detection process, one may include alternative electrodes that may be brought to bear in the S-ICD or US-ICD systems either by placing them in the detection algorithm circuitry through a programming maneuver or by manually adding such additional electrode systems to the S-ICD or US-ICD at the time of implant or at the time of follow-up evaluation. One may

also use electrodes for the detection of atrial fibrillation that may or may not also be used for the detection of ventricular arrhythmias given the different anatomic locations of the atria and ventricles with respect to the S-ICD or US-ICD housing and surgical implant sites.

5 Once atrial fibrillation is detected, the arrhythmia can be treated by delivery of a synchronized shock using energy levels up to the maximum output of the device therapy for terminating atrial fibrillation or for other supraventricular arrhythmias. The S-ICD or US-ICD electrode system can be used to treat both atrial and ventricular arrhythmias not only with shock therapy but also with pacing
10 therapy. In a further embodiment of the treatment of atrial fibrillation or other atrial arrhythmias, one may be able to use different electrode systems than what is used to treat ventricular arrhythmias. Another embodiment, would be to allow for different types of therapies (amplitude, waveform, capacitance, etc.) for atrial arrhythmias compared to ventricular arrhythmias.

15 The core member of the different sized and shaped US-ICD will all be the same size and shape. That way, during an implantation procedures, multiple sized US-ICDs can be available for implantation, each one without a core member. Once the implantation procedure is being performed, then the correct sized US-ICD can be selected and the core member can be inserted into the US-ICD and
20 then programmed as described above. Another advantage of this configuration is when the battery within the core member needs replacing it can be done without removing the entire US-ICD.

 Post-shock bradycardia is a common after-effect of shocking the heart for cardioversion/defibrillation therapy. Symptoms related to low blood pressure may
25 result from post-shock bradycardia whenever the heart rate falls below approximately 30 to approximately 50 beats per minute. Accordingly, it is often desirable to provide anti-bradycardia pacing to correct the symptoms resulting from bradycardia.

 To ensure adequate pacing capture of the heart through a subcutaneous only
30 lead system, pacing therapy can be considerably enhanced (i.e., require less energy and voltage) by using either a monophasic or a biphasic waveform for pacing.

 Fig. 19 is a graph that shows an embodiment of the example of a biphasic waveform for use in anti-bradycardia pacing applications in subcutaneous

implantable cardioverter-defibrillators (“S-ICD”) in an embodiment of the present invention. As shown in Fig. 19, the biphasic waveform is plotted as a function of current versus time.

5 In an embodiment, the biphasic waveform 1902 comprises a positive portion 1904, a negative portion 1906 and a transition portion 1908. In an embodiment, both the positive portion 1904 and the negative portion 1906 are substantially rectangular in shape. The positive portion 1904 of the biphasic waveform 1902 comprises an initial positive current 1910, a positive fixed current 1912 and a final positive current 1914. The negative portion 1906 of the biphasic waveform 1902 comprises an initial negative current 1916, a negative fixed current 1918 and a final negative current 1920. In an embodiment, the polarities of the biphasic waveform 1902 can be reversed such that the negative portion 1906 precedes the positive portion 1904 in time.

15 As shown in Fig. 19, the biphasic waveform 1902 is initially at zero current. Upon commencement of the anti-bradycardia pacing, a current of positive polarity is provided and the biphasic waveform 1902 rises to the initial positive current 1910. Next, the current of the biphasic waveform 1902 remains at a constant level along the positive fixed current 1912. The positive portion 1904 of the biphasic waveform 1902 is then truncated and a negative current is provided. 20 The biphasic waveform 1902 then undergoes a relatively short transition portion 1908 where the current is approximately zero. Next, the biphasic waveform 1902 is increased (in absolute value) in the opposite (negative) polarity to the initial negative current 1916. After reaching its maximum negative current (in absolute value), the current of the biphasic waveform 1902 remains at a constant level along the negative fixed current 1918. After the negative portion 1906 of the biphasic waveform 1902 is truncated at the final negative current 1914, the biphasic waveform 1902 returns to zero. 25

The total amount of time that the biphasic waveform 1902 comprises is known as the “pulse width.” In an embodiment, the pulse width of the biphasic waveform can range from approximately 1 millisecond to approximately 40 milliseconds. The total amount of energy delivered is a function of the pulse width and the absolute value of the current. 30

An example of one embodiment of the biphasic waveform 1902 will now be described. In this embodiment, the amplitude of the initial positive current 1910 can range from approximately one to approximately 250 milliamps. Similarly, the amplitude of the initial negative current 1916 can range from approximately one to approximately 250 milliamps.

In the example, the pulse width of the biphasic waveform 1902 can range from approximately 1 millisecond to approximately 40 milliseconds. In addition, the implantable cardioverter-defibrillator employs biphasic anti-bradycardia pacing at rates of approximately 20 to approximately 120 stimuli/minute for severe bradycardia episodes although programming of higher pacing rates up to 120 stimuli/minute is also possible.

Fig. 20 is a graph that shows an embodiment of the example of a monophasic waveform for use in anti-bradycardia pacing applications in subcutaneous implantable cardioverter-defibrillators ("S-ICD") in an embodiment of the present invention. As shown in Fig. 20, the monophasic waveform is plotted as a function of current versus time.

In an embodiment, the monophasic waveform 2002 comprises an initial positive current 2004, a positive fixed current 2006 and a final positive current 2008. In an embodiment, the monophasic waveform 2002 is substantially rectangular in shape. In an embodiment, the polarities of the monophasic waveform 2002 can be reversed such that the waveform 2002 is negative in polarity.

As shown in Fig. 20, the monophasic waveform 2002 is initially at zero current. Upon commencement of the anti-bradycardia pacing, a current of positive polarity is provided and the monophasic waveform 2002 rises to the initial positive current 2004. Next, the current of the monophasic waveform 2002 remains at a constant level along the positive fixed current 1906. The monophasic waveform 2002 is then truncated.

The total amount of time that the monophasic waveform 2002 comprises is known as the "pulse width." In an embodiment, the pulse width of the monophasic waveform can range from approximately 1 millisecond to approximately 40 milliseconds. The total amount of energy delivered is a function of the pulse width and the absolute value of the current.

An example of one embodiment of the monophasic waveform 2002 will now be described. In this embodiment, the amplitude of the initial positive current 2004 can range from approximately one to approximately 250 milliamps.

5 In the example, the pulse width of the monophasic waveform 2002 can range from approximately 1 millisecond to approximately 40 milliseconds. In addition, the implantable cardioverter-defibrillator employs monophasic anti-bradycardia pacing at rates of approximately 20 to approximately 120 stimuli/minute for severe bradycardia episodes although programming of higher pacing rates up to 120 stimuli/minute is also possible. In order to maintain these rates, in one embodiment of the invention, the power supply continues to operate to maintain a sufficient voltage to deliver a constant current.

10 Although it possible for the present invention to provide standard VVI pacing at predetermined or preprogrammed rates, one embodiment provides anti-bradycardia pacing only for bradycardia or post-shock bradycardia. To avoid frequent anti-bradycardia pacing at 50 stimuli/minute but to provide this rate in case of emergencies, a hysteresis detection trigger can be employed at lower rates, typically in the range of approximately 20 to approximately 40 stimuli/minute. For example, a default setting may be set at approximately 20 stimuli/minute (i.e., the equivalent of a 3 second pause), and the invention providing VVI pacing at a rate of approximately 50 stimuli/minute only when such a pause occurs. In another embodiment, the invention can provide physiologic pacing in a VVIR mode of operation in response to a certain activity, respiration, pressure or oxygenation sensor.

20 The S-ICD and US-ICD devices and methods of the present invention may be embodied in other specific forms without departing from the teachings or essential characteristics of the invention. The described embodiments are therefore to be considered in all respects as illustrative and not restrictive, the scope of the invention being indicated by the appended claims rather than by the foregoing description, and all changes which come within the meaning and range of equivalency of the claims are therefore to be embraced therein.

30

What is claimed is:

- 1 . A power supply for an implantable cardioverter-defibrillator for subcutaneous positioning between the third rib and the twelfth rib and using a lead system that does not directly contact a patient's heart or reside in the intrathoracic blood vessels and for providing anti-bradycardia pacing energy to the heart, the power supply comprising:
- 5
- a capacitor subsystem for storing the anti-bradycardia pacing energy for delivery to the patient's heart; and
- a battery subsystem electrically coupled to the capacitor subsystem for providing the anti-bradycardia pacing energy to the capacitor subsystem.
- 10
- 2 . The power supply of claim 1, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately one milliamp to approximately 250 milliamps.
- 15
- 3 . The power supply of claim 2, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately one milliamp to approximately 50 milliamps.
- 20
- 4 . The power supply of claim 2, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 50 milliamps to approximately 100 milliamps.
- 25
- 5 . The power supply of claim 2, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 100 milliamps to approximately 150 milliamps.
- 30
- 6 . The power supply of claim 2, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 150 milliamps to approximately 200 milliamps.

7 . The power supply of claim 2, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 200 milliamps to approximately 250 milliamps.

5 8 . The power supply of claim 1, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 1 millisecond to approximately 40 milliseconds.

9 . The power supply of claim 8, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 1 millisecond to approximately 10 milliseconds.

10 10 . The power supply of claim 8 wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 10 milliseconds to approximately 20 milliseconds.

11 . The power supply of claim 8 wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 20 milliseconds to approximately 30 milliseconds.

20

12 . The power supply of claim 8 wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 30 milliseconds to approximately 40 milliseconds.

25 13 . The power supply of claim 1, wherein the anti-bradycardia pacing energy comprises a biphasic waveform further comprising a portion that is positive in polarity and a portion that is negative in polarity.

14 . The power supply of claim 1, wherein the anti-bradycardia pacing energy comprises a biphasic waveform that is provided at a rate of approximately 20 to approximately 120 stimuli/minute.

5 15 . The power supply of claim 25, wherein the biphasic waveform is provided after a patient's heart rate is greater than or equal to approximately 20 beats/minute.

10 16 . The power supply of claim 1, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately one milliamp to approximately 250 milliamps.

15 17 . The power supply of claim 16, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately one milliamp to approximately 50 milliamps.

20 18 . The power supply of claim 16, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 50 milliamps to approximately 100 milliamps.

 19 . The power supply of claim 16, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 100 milliamps to approximately 150 milliamps.

25 20 . The power supply of claim 16, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 150 milliamps to approximately 200 milliamps.

21 . The power supply of claim 16, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 200 milliamps to approximately 250 milliamps.

5 22 . The power supply of claim 1, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 1 millisecond to approximately 40 milliseconds.

10 23 . The power supply of claim 22, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 1 millisecond to approximately 10 milliseconds.

15 24 . The power supply of claim 22, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 10 milliseconds to approximately 20 milliseconds.

20 25 . The power supply of claim 22, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 20 milliseconds to approximately 30 milliseconds.

 26 . The power supply of claim 22, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 30 milliseconds to approximately 40 milliseconds.

25 27 . The power supply of claim 1, wherein the anti-bradycardia pacing energy comprises a monophasic waveform that is either positive or negative in polarity.

28 . The power supply of claim 1, wherein the anti-bradycardia pacing energy comprises a monophasic waveform that is provided at a rate of approximately 20 to approximately 120 stimuli/minute.

5 29 . The power supply of claim 28, wherein the monophasic waveform is provided after a patient's heart rate is greater than or equal to approximately 20 beats/minute.

10 30 . Current output system for an implantable cardioverter-defibrillator using a lead system that does not directly contact a patient's heart or reside in the intrathoracic blood vessels and for providing anti-bradycardia pacing energy to the heart, the power supply comprising:
 an energy storage system for storing the anti-bradycardia pacing energy for delivery to the patient's heart; and
15 an energy source system electrically coupled to the capacitor subsystem for providing the anti-bradycardia pacing energy to the capacitor subsystem.

20 31 . Current output system of claim 30, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately one milliamp to approximately 250 milliamps.

25 32 . Current output system of claim 31, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately one milliamp to approximately 50 milliamps.

 33 . Current output system of claim 31, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 50 milliamps to approximately 100 milliamps.

34 . Current output system of claim 31, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 100 milliamps to approximately 150 milliamps.

5 35 . Current output system of claim 31, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 150 milliamps to approximately 200 milliamps.

10 36 . Current output system of claim 31, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 200 milliamps to approximately 250 milliamps.

15 37 . Current output system of claim 30, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 1 millisecond to approximately 40 milliseconds.

20 38 . Current output system of claim 37, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 1 millisecond to approximately 10 milliseconds.

 39 . Current output system of claim 37, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 10 milliseconds to approximately 20 milliseconds.

25 40 . Current output system of claim 37, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 20 milliseconds to approximately 30 milliseconds.

41 . Current output system of claim 37, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 30 milliseconds to approximately 40 milliseconds.

5 42 . Current output system of claim 30, wherein the anti-bradycardia pacing energy comprises a biphasic waveform further comprising a positive voltage portion and a negative voltage portion.

10 43 . Current output system of claim 30, wherein the anti-bradycardia pacing energy comprises a biphasic waveform that is provided at a rate of approximately 20 to approximately 120 stimuli/minute.

15 44 . Current output system of claim 43, wherein the biphasic waveform is provided after a patient's heart rate is greater than or equal to approximately 20 beats/minute.

20 45 . Current output system of claim 30, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately one milliamp to approximately 250 milliamps.

 46 . Current output system of claim 45, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately one milliamp to approximately 50 milliamps.

25 47 . Current output system of claim 45, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 50 milliamps to approximately 100 milliamps.

48. Current output system of claim 45, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 100 milliamps to approximately 150 milliamps.

5 49. Current output system of claim 45, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 150 milliamps to approximately 200 milliamps.

10 50. Current output system of claim 45, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 200 milliamps to approximately 250 milliamps.

15 51. Current output system of claim 30, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 1 millisecond to approximately 40 milliseconds.

20 52. Current output system of claim 51, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 1 millisecond to approximately 10 milliseconds.

53. Current output system of claim 51, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 10 milliseconds to approximately 20 milliseconds.

25 54. Current output system of claim 51, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 20 milliseconds to approximately 30 milliseconds.

55 . Current output system of claim 51, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 30 milliseconds to approximately 40 milliseconds.

5 56 . Current output system of claim 30, wherein the anti-bradycardia pacing energy comprises a monophasic waveform that is either positive or negative in polarity.

10 57 . Current output system of claim 30, wherein the anti-bradycardia pacing energy comprises a monophasic waveform that is provided at a rate of approximately 20 to approximately 120 stimuli/minute.

15 58 . Current output system of claim 57, wherein the monophasic waveform is provided after a patient's heart rate is greater than or equal to approximately 20 beats/minute.

59 . An implantable cardioverter-defibrillator for subcutaneous positioning between the third rib and the twelfth rib within a patient, the implantable cardioverter-defibrillator comprising:

20 a housing having an electrically conductive surface on an outer surface of the housing;

a lead assembly electrically coupled to the housing and having an electrode, wherein the lead assembly does not directly contact the patient's heart or reside in the intrathorasic blood vessels;

25 a capacitor subsystem located within the housing and electrically coupled to the electrically conductive surface and the electrode for storing anti-bradycardia pacing energy and for delivering the anti-bradycardia pacing energy to the patient's heart through the electrically conductive surface and the electrode; and

30 a battery subsystem electrically coupled to the capacitor subsystem for providing the anti-bradycardia pacing energy to the capacitor subsystem.

60. The implantable cardioverter-defibrillator of claim 59, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately one milliamp to approximately 250 milliamps.

5 61. The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately one milliamp to approximately 50 milliamps.

10 62. The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 50 milliamps to approximately 100 milliamps.

15 63. The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 100 milliamps to approximately 150 milliamps.

20 64. The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 150 milliamps to approximately 200 milliamps.

 65. The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 200 milliamps to approximately 250 milliamps.

25 66. The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 1 millisecond to approximately 40 milliseconds.

67 . The implantable cardioverter-defibrillator of claim 66, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 1 millisecond to approximately 10 milliseconds.

5 68 . The implantable cardioverter-defibrillator of claim 66, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 10 milliseconds to approximately 20 milliseconds.

10 69 . The implantable cardioverter-defibrillator of claim 66, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 20 milliseconds to approximately 30 milliseconds.

15 70 . The implantable cardioverter-defibrillator of claim 66, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 30 milliseconds to approximately 40 milliseconds.

20 71 . The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a biphasic waveform further comprising a portion that is positive in polarity and a portion that is negative in polarity.

 72 . The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a biphasic waveform that is provided at a rate of approximately 20 to approximately 120 stimuli/minute.

25 73 . The implantable cardioverter-defibrillator of claim 72, wherein the biphasic waveform is provided after a patient's heart rate is greater than or equal to approximately 20 beats/minute.

74 . The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately one milliamp to approximately 250 milliamps.

5 75 . The implantable cardioverter-defibrillator of claim 74, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately one milliamp to approximately 50 milliamps.

10 76 . The implantable cardioverter-defibrillator of claim 74, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 50 milliamps to approximately 100 milliamps.

15 77 . The implantable cardioverter-defibrillator of claim 74, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 100 milliamps to approximately 150 milliamps.

20 78 . The implantable cardioverter-defibrillator of claim 74, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 150 milliamps to approximately 200 milliamps.

 79 . The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 200 milliamps to approximately 250 milliamps.

25 80 . The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 1 millisecond to approximately 40 milliseconds.

81 . The implantable cardioverter-defibrillator of claim 80, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 1 millisecond to approximately 10 milliseconds.

5 82 . The implantable cardioverter-defibrillator of claim 80, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 10 milliseconds to approximately 20 milliseconds.

10 83 . The implantable cardioverter-defibrillator of claim 80, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 20 milliseconds to approximately 30 milliseconds.

15 84 . The implantable cardioverter-defibrillator of claim 80, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 30 milliseconds to approximately 40 milliseconds.

20 85 . The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a monophasic waveform that is either positive or negative in polarity.

 86 . The implantable cardioverter-defibrillator of claim 60, wherein the anti-bradycardia pacing energy comprises a monophasic waveform that is provided at a rate of approximately 20 to approximately 120 stimuli/minute.

25 87 . The implantable cardioverter-defibrillator of claim 86, wherein the monophasic waveform is provided after a patient's heart rate is greater than or equal to approximately 20 beats/minute.

88 . A method for supplying power for an implantable cardioverter-defibrillator for subcutaneous positioning between the third rib and the twelfth rib and using a lead system that does not directly contact a patient's heart or reside in the intrathoracic blood vessels and for providing anti-bradycardia pacing energy to the heart, the method comprising:

generating anti-bradycardia pacing energy;
storing the anti-bradycardia pacing energy; and
delivering the anti-bradycardia pacing energy to the patient's heart.

89 . The method of claim 88, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately one milliamp to approximately 250 milliamps.

90 . The method of claim 89, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately one milliamp to approximately 50 milliamps.

91 . The method of claim 89, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 50 milliamps to approximately 100 milliamps.

92 . The method of claim 89, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 100 milliamps to approximately 150 milliamps.

93 . The method of claim 89, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 150 milliamps to approximately 200 milliamps.

94 . The method of claim 89, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a peak current that is approximately 200 milliamps to approximately 250 milliamps.

5 95 . The method of claim 88, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 1 millisecond to approximately 40 milliseconds.

10 96 . The method of claim 95, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 1 millisecond to approximately 10 milliseconds.

15 97 . The method of claim 95, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 10 milliseconds to approximately 20 milliseconds.

20 98 . The method of claim 95, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 20 milliseconds to approximately 30 milliseconds.

 99 . The method of claim 95, wherein the anti-bradycardia pacing energy comprises a biphasic waveform having a pulse width that is approximately 30 milliseconds to approximately 40 milliseconds.

25 100 . The method of claim 88, wherein the anti-bradycardia pacing energy comprises a biphasic waveform further comprising a portion that is positive in polarity and a portion that is negative in polarity.

101. The method of claim 88, wherein the anti-bradycardia pacing energy comprises a biphasic waveform that is provided at a rate of approximately 20 to approximately 120 stimuli/minute.

5 102. The method of claim 101, wherein the biphasic waveform is provided after a patient's heart rate is greater than or equal to approximately 20 beats/minute.

10 103. The method of claim 88, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately one milliamp to approximately 250 milliamps.

15 104. The method of claim 103, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately one milliamp to approximately 50 milliamps.

20 105. The method of claim 103, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 50 milliamps to approximately 100 milliamps.

 106. The method of claim 103, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 100 milliamps to approximately 150 milliamps.

25 107. The method of claim 103, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 150 milliamps to approximately 200 milliamps.

108. The method of claim 103, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a peak current that is approximately 200 milliamps to approximately 250 milliamps.

5 109. The method of claim 88, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 1 millisecond to approximately 40 milliseconds.

10 110. The method of claim 109, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 1 millisecond to approximately 10 milliseconds.

15 111. The method of claim 109, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 10 milliseconds to approximately 20 milliseconds.

20 112. The method of claim 109, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 20 milliseconds to approximately 30 milliseconds.

 113. The method of claim 109, wherein the anti-bradycardia pacing energy comprises a monophasic waveform having a pulse width that is approximately 30 milliseconds to approximately 40 milliseconds.

25 114. The method of claim 88, wherein the anti-bradycardia pacing energy comprises a monophasic waveform that is either positive or negative in polarity.

115. The method of claim 88, wherein the anti-bradycardia pacing energy comprises a monophasic waveform that is provided at a rate of approximately 20 to approximately 120 stimuli/minute.

5 116. The method of claim 115, wherein the monophasic waveform is provided after a patient's heart rate is greater than or equal to approximately 20 beats/minute.

10 117. The method of claim 88, wherein the implantable cardioverter-defibrillator is subcutaneously positioned between the third and fourth ribs.

118. The method of claim 88, wherein the implantable cardioverter-defibrillator is subcutaneously positioned between the fourth and sixth ribs.

15 119. The method of claim 88, wherein the implantable cardioverter-defibrillator is subcutaneously positioned between the sixth and eighth ribs.

120. The method of claim 88, wherein the implantable cardioverter-defibrillator is subcutaneously positioned between the eighth and tenth ribs.

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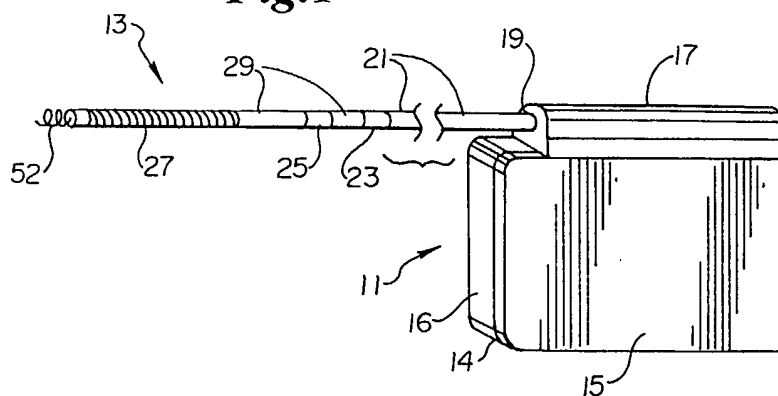
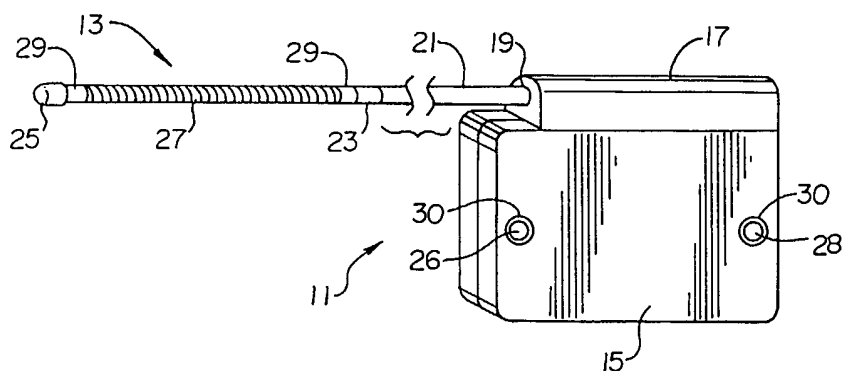
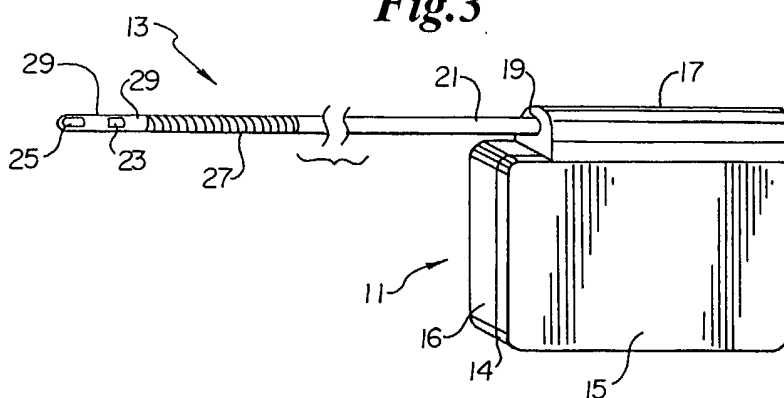
121. The method of claim 88, wherein the implantable cardioverter-defibrillator is subcutaneously positioned between the tenth and twelfth ribs.

122 . The method of claim 88, wherein the implantable cardioverter-defibrillator provides anti-bradycardia pacing energy to the heart for treatment of atrial fibrillation.

5 123 . The method of claim 88, wherein the implantable cardioverter-defibrillator provides anti-bradycardia pacing energy to the heart for treatment of ventricular fibrillation.

10 124 . The power supply of claim 1, wherein the battery subsystem and the capacitor system provide a sufficient voltage to provide an anti-bradycardia pacing energy comprising an approximately constant current.

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Fig.1**Fig.2****Fig.3**

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Fig.4

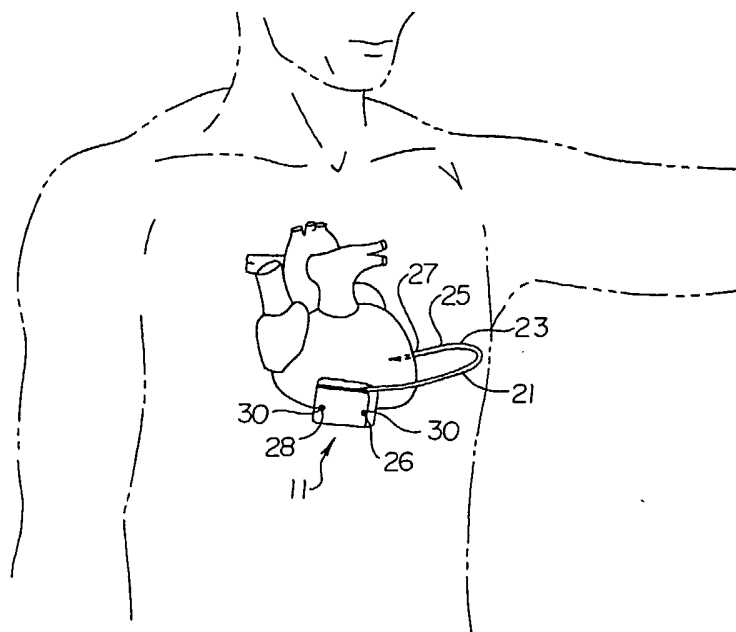
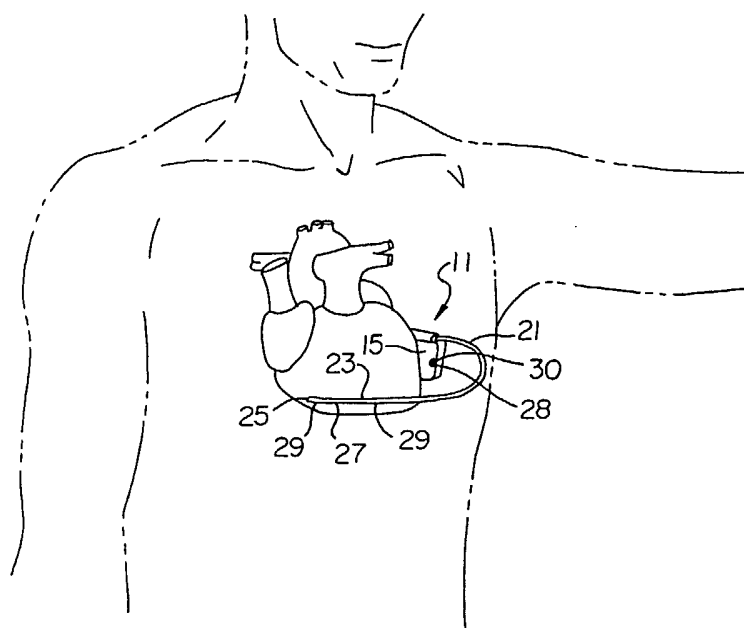


Fig.5



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Fig.6

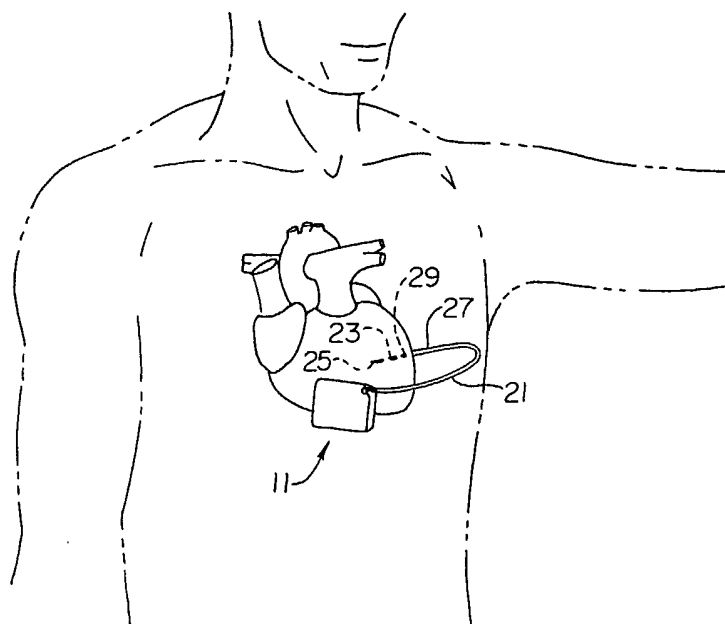
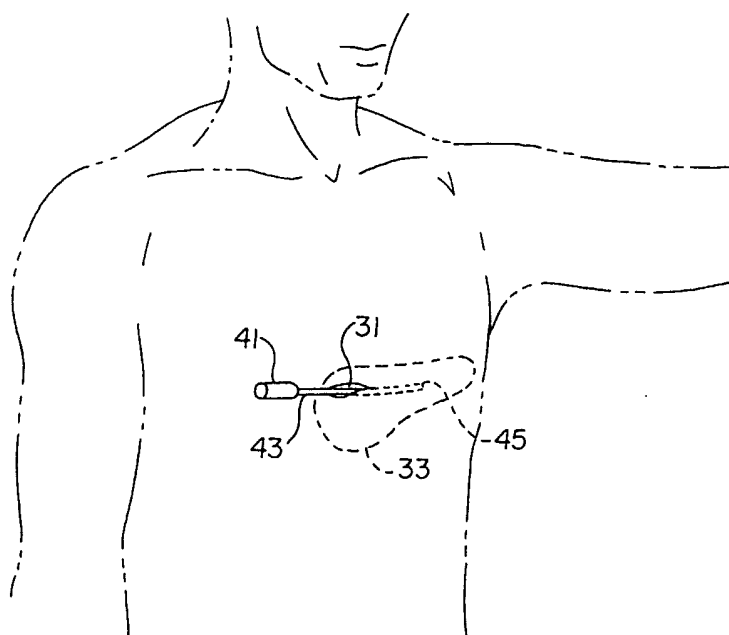


Fig.7



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Fig.8

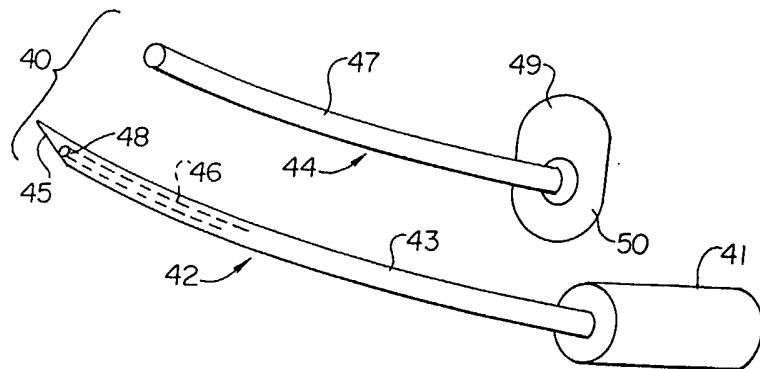
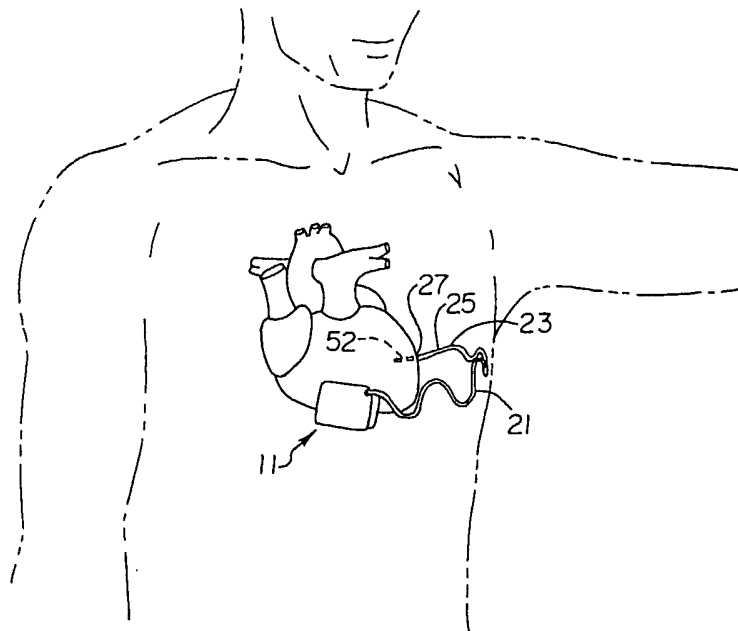
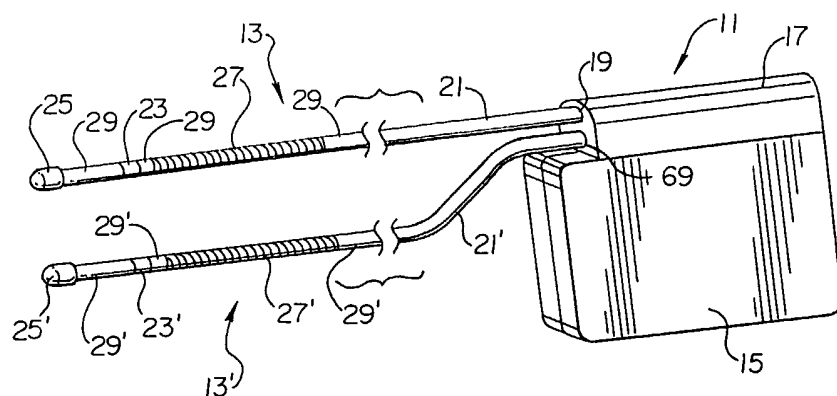
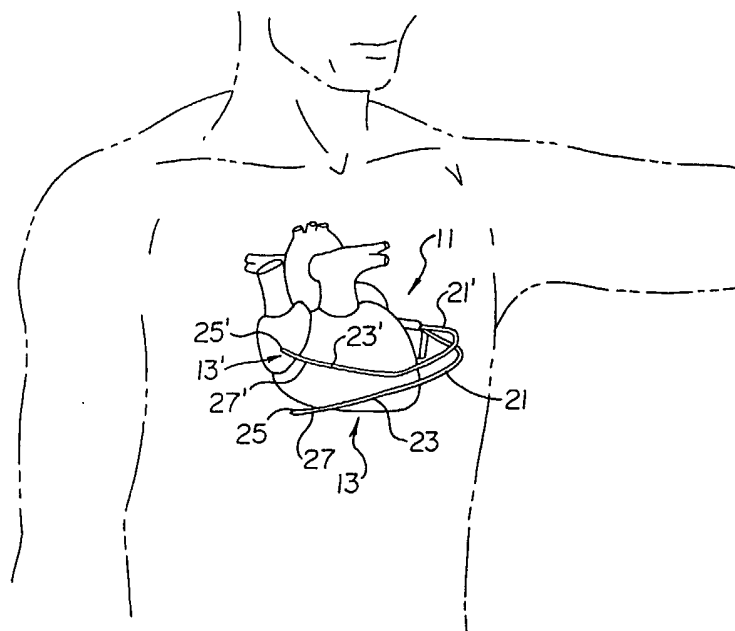


Fig.9



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Fig.10**Fig.11**

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Fig.12

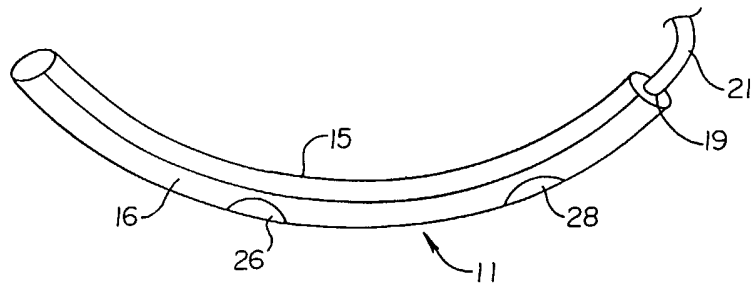


Fig.13

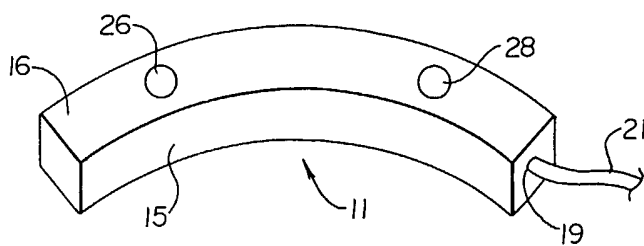
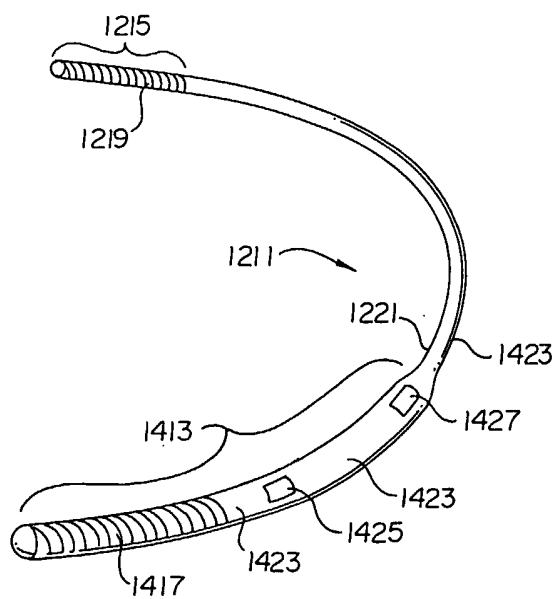


Fig.14



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Fig.15

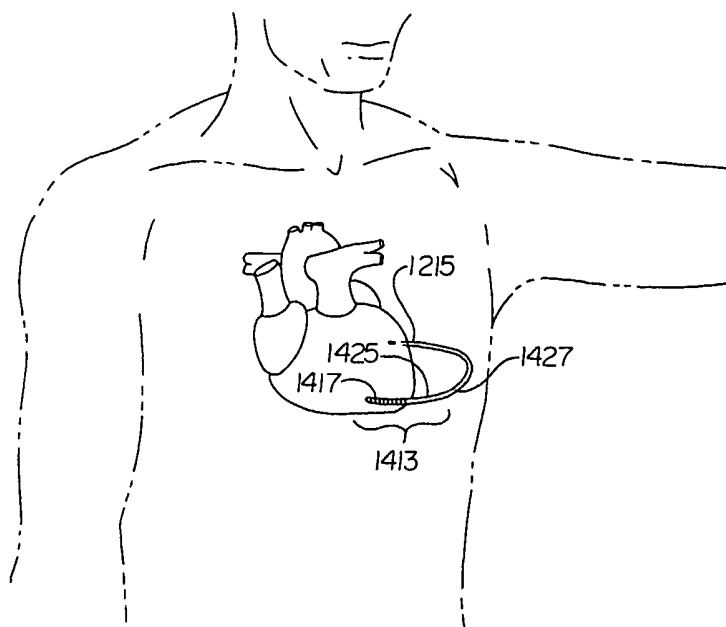
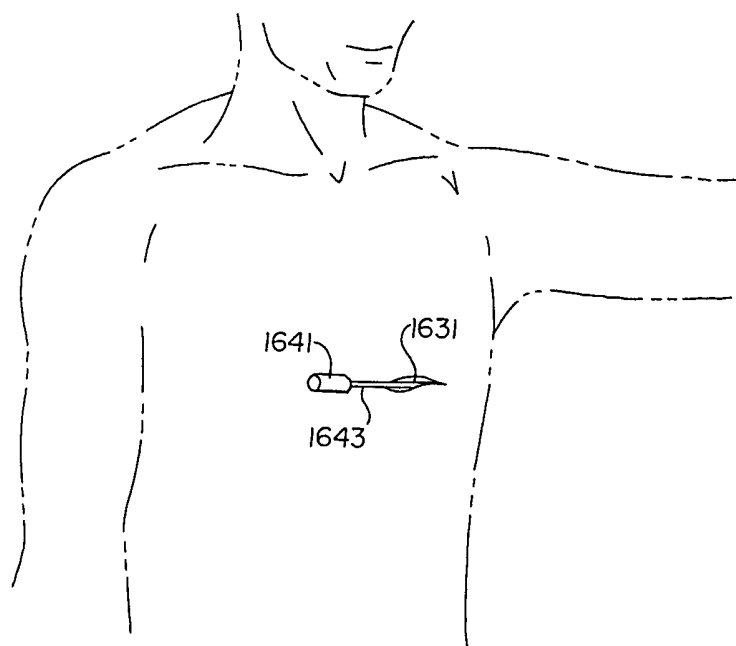
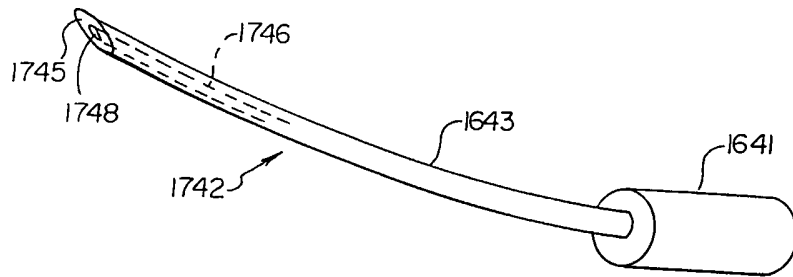
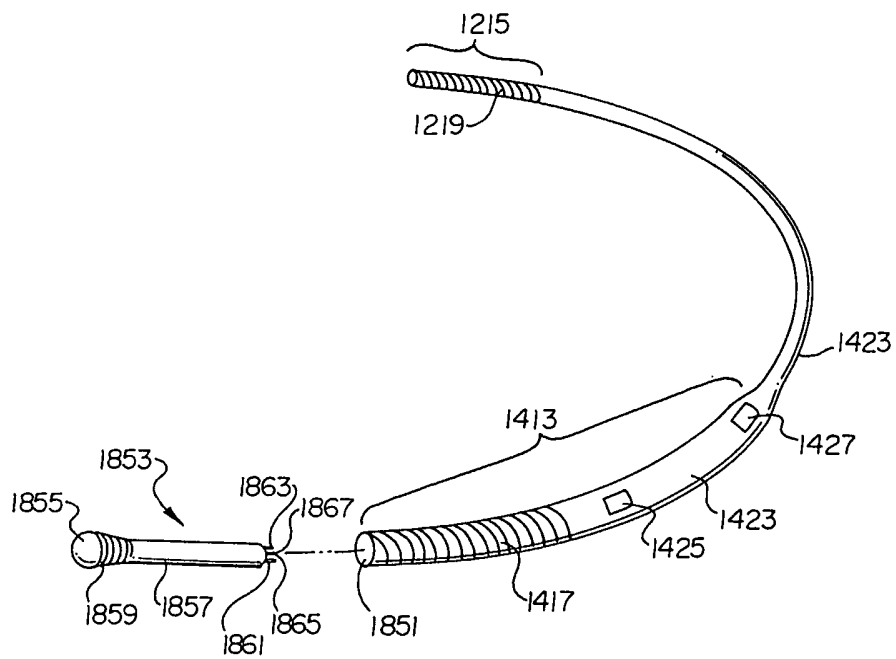


Fig.16



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Fig.17**Fig.18**

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Fig. 19

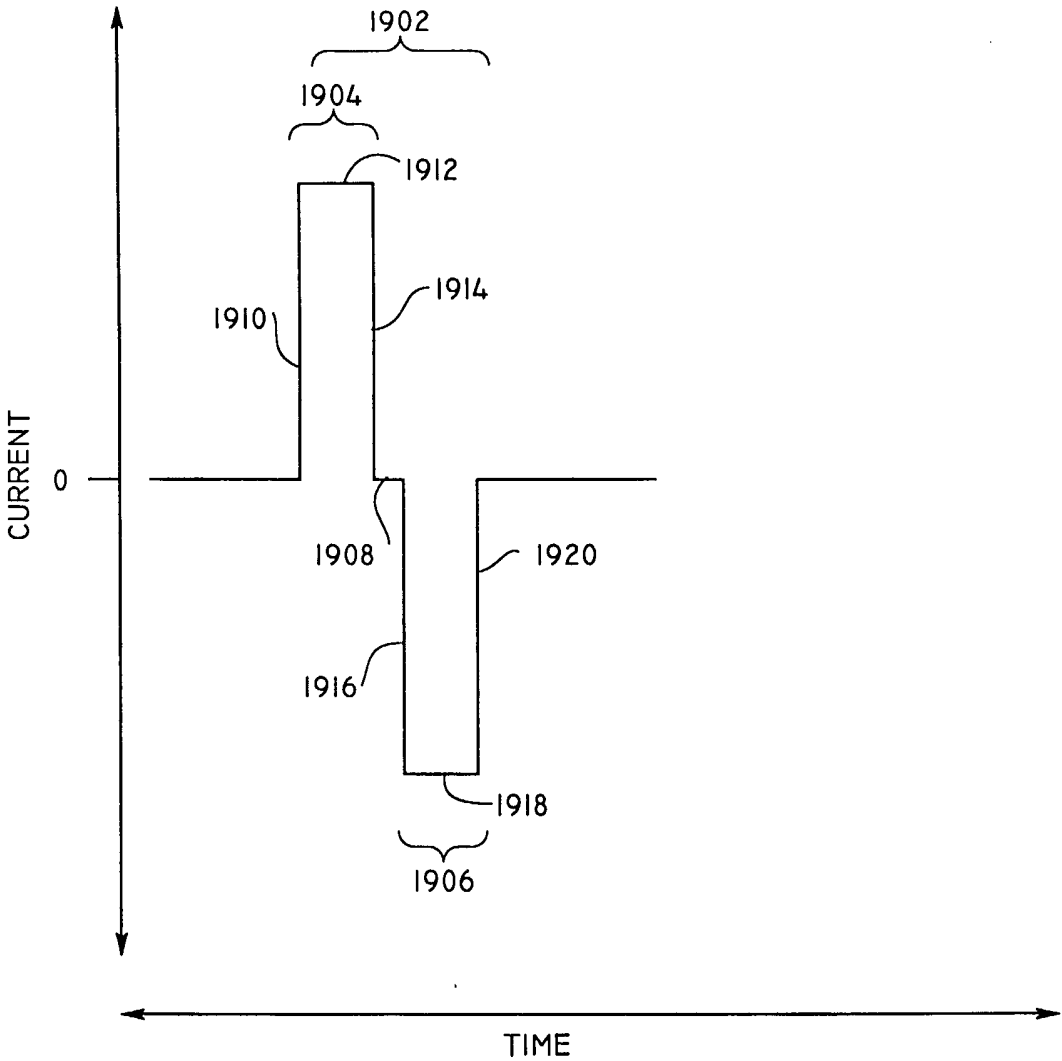
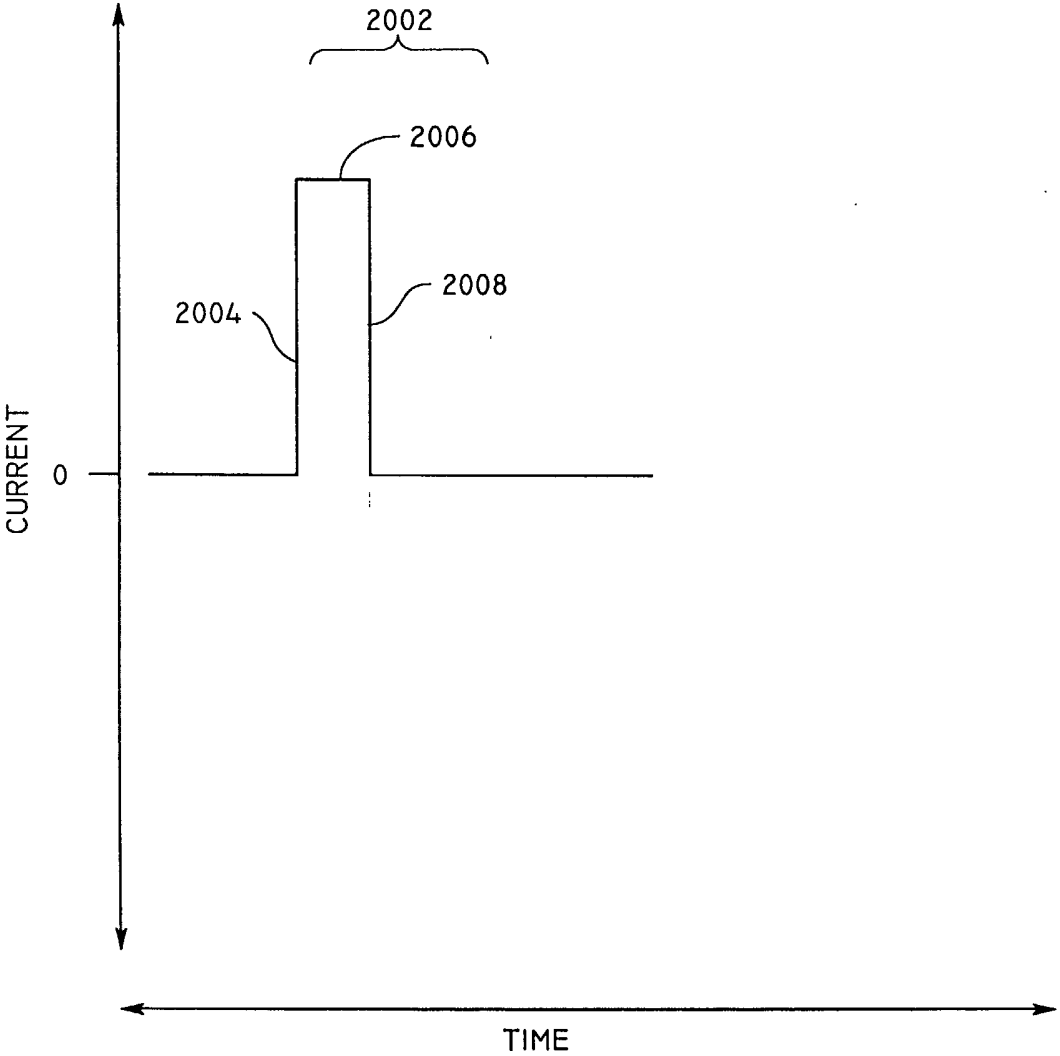


Fig.20



INTERNATIONAL SEARCH REPORT

International Application No

PCT/IB 02/04498

A. CLASSIFICATION OF SUBJECT MATTER
IPC 7 A61N1/39

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

IPC 7 A61N

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

EP0-Internal

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category *	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 5 184 616 A (WEISS STEVEN M) 9 February 1993 (1993-02-09) column 6, line 3 - line 33 column 9, line 1 - line 21 column 14, line 11 - column 15, line 8 column 16, line 49 - column 17, line 32	1-58,124
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Y	US 5 713 926 A (KENKNIGHT BRUCE H ET AL) 3 February 1998 (1998-02-03) column 5, line 12 - line 22	59-87
X	EP 0 518 599 A (TELECTRONICS NV) 16 December 1992 (1992-12-16) abstract	1,30

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☒ Further documents are listed in the continuation of box C.☒ Patent family members are listed in annex.

* Special categories of cited documents:

A document defining the general state of the art which is not considered to be of particular relevance

E earlier document but published on or after the international filing date

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O document referring to an oral disclosure, use, exhibition or other means

P document published prior to the international filing date but later than the priority date claimed

T later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

X document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

Y document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

& document member of the same patent family

Date of the actual completion of the international search

9 April 2003

Date of mailing of the international search report

16/04/2003

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Authorized officer

Petter, E

INTERNATIONAL SEARCH REPORT

Intern al Application No
PCT/IB 02/04498

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT		
Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US 5 413 591 A (KNOLL MARK W) 9 May 1995 (1995-05-09) column 4, line 30 - line 39 ---	1, 30, 59
A	WO 99 03534 A (SCHULER ELEANOR L ;CPR MEDICAL INC (US); SCOTT DALE L (US)) 28 January 1999 (1999-01-28) abstract -----	1-123

INTERNATIONAL SEARCH REPORT

International application No.
PCT/IB 02/04498

Box I Observations where certain claims were found unsearchable (Continuation of item 1 of first sheet)

This International Search Report has not been established in respect of certain claims under Article 17(2)(a) for the following reasons:

1. ☒ Claims Nos.: 88-123
because they relate to subject matter not required to be searched by this Authority, namely:
Rule 39.1(iv) PCT - Method for treatment of the human or animal body by therapy
2. ☐ Claims Nos.:
because they relate to parts of the International Application that do not comply with the prescribed requirements to such an extent that no meaningful International Search can be carried out, specifically:
3. ☐ Claims Nos.:
because they are dependent claims and are not drafted in accordance with the second and third sentences of Rule 6.4(a).

Box II Observations where unity of invention is lacking (Continuation of item 2 of first sheet)

This International Searching Authority found multiple inventions in this international application, as follows:

1. ☐ As all required additional search fees were timely paid by the applicant, this International Search Report covers all searchable claims.
2. ☐ As all searchable claims could be searched without effort justifying an additional fee, this Authority did not invite payment of any additional fee.
3. ☐ As only some of the required additional search fees were timely paid by the applicant, this International Search Report covers only those claims for which fees were paid, specifically claims Nos.:
4. ☐ No required additional search fees were timely paid by the applicant. Consequently, this International Search Report is restricted to the invention first mentioned in the claims; it is covered by claims Nos.:

Remark on Protest

- ☐ The additional search fees were accompanied by the applicant's protest.
- ☐ No protest accompanied the payment of additional search fees.

INTERNATIONAL SEARCH REPORT

Information on patent family members

International Application No

PCT/IB 02/04498

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