(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
4 May 2006 (04.05.2006)

PCT

(10) International Publication Number
WO 2006/045148 A2

(51) International Patent Classification: Not classified

(21) International Application Number: PCT/AU2005/001658

(22) International Filing Date: 28 October 2005 (28.10.2005)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
60/622,612 28 October 2004 (28.10.2004) US
60/622,602 28 October 2004 (28.10.2004) US

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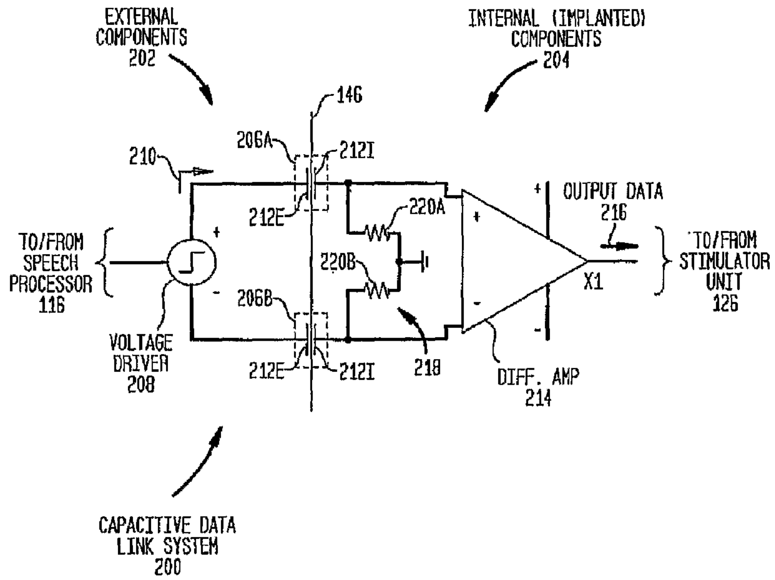
(81) Designated States *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AT, AU, AZ, BA, BB, BG, BR, BW, BY, BZ, CA, CH, CN, CO, CR, CU, CZ, DE, DK, DM, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, HR, HU, ID, IL, IN, IS, JP, KE, KG, KM, KN, KP, KR, KZ, LC, LK, LR, LS, LT, LU, LV, LY, MA, MD, MG, MK, MN, MW, MX, MZ, NA, NG, NI, NO, NZ, OM, PG, PH, PL, PT, RO, RU, SC, SD, SE, SG, SK, SL, SM, SY, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, YU, ZA, ZM, ZW.

(84) Designated States *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, GH, GM, KE, LS, MW, MZ, NA, SD, SL, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European (AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HU, IE, IS, IT, LT, LU, LV, MC, NL, PL, PT, RO, SE, SI, SK, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, ML, MR, NE, SN, TD, TG).

Published:
— *without international search report and to be republished upon receipt of that report*

For two-letter codes and other abbreviations, refer to the "Guidance Notes on Codes and Abbreviations" appearing at the beginning of each regular issue of the PCT Gazette.

(54) Title: TRANSCUTANEOUS CAPACITIVE DATA LINK



(57) Abstract: There is disclosed a transcutaneous data communications link (200) for am implantable medical device, the transcutaneous data communications link comprising: at least two capacitors (206A, 206B), each capacitor having an externally located electrode and an implanted electrode separated by a transcutaneous dielectric (146); an implanted data receiver (204) in communication with each one of the at least two capacitors; an externally located voltage driver (208) arranged to communicate a differential data signal to the implanted data receiver through the at least two capacitors.

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TRANSCUTANEOUS CAPACITIVE DATA LINK

CROSS-REFERENCE TO RELATED APPLICATIONS

[0001] This application is related to U.S. Patent Application Nos. 10/883,809, 10/856,823, 10/333,676, 10/887,894, and 10/887,893 and US. Patents 6,810,289, 6,751,505, and 6,700,982 which are hereby incorporated by reference herein in its entirety.

BACKGROUND

Field of the Invention

[0002] The present invention relates generally to transcutaneous transfer systems and, more particularly, to a transcutaneous capacitive data link.

Related Art

[0003] The use of implantable medical devices to provide therapy to individuals for various medical conditions has become more widespread as the advantages and benefits such devices provide become more widely appreciated and accepted throughout the population. In particular, devices such as hearing aids, implantable pacemakers, defibrillators, functional electrical stimulation devices such as cochlear prostheses, organ assist or replacement devices, and other medical devices, have been successful in performing life saving and/or lifestyle enhancement functions for a number of individuals.

[0004] Medical devices often include one or more sensors, processors, controllers or other functional electrical components that are permanently or temporarily implanted in a patient. Many such implantable devices require power and/or require communications with external systems that are part of or operate in conjunction with the medical device. One common approach to provide for the transcutaneous transfer of power and/or communications with an implantable component is via a transcutaneous transfer system.

[0005] One type of medical device that may include a transcutaneous transfer system is a cochlear™ implant system. Cochlear implant systems provide the benefit of hearing to individuals suffering from severe to profound hearing loss. Hearing loss in such individuals is due to the absence or destruction of the hair cells in the cochlea which transduce acoustic signals into nerve impulses. Cochlear implants essentially simulate the cochlear hair cells by directly delivering electrical stimulation to the auditory nerve fibers. This causes the brain to perceive a hearing sensation resembling the natural hearing sensation normally delivered to the auditory nerve.

[0006] Conventional cochlear implant systems primarily include external components directly or indirectly attached to the body of the patient (sometimes referred to herein as the recipient), and internal components which are implanted in the patient. The external components typically comprise a microphone for detecting sounds, a speech processor that converts the detected sounds into a coded signal, a power source, and an external transmitter antenna coil. The internal components typically comprise an internal receiver antenna coil, a stimulator located within a recess of the temporal bone of the recipient, and an electrode array positioned in the recipient's cochlear.

[0007] Collectively, the external transmitter antenna coil and the internal receiver antenna coil form an inductively-coupled transcutaneous transfer system. The external transmitter antenna coil is usually positioned on the side of a recipient's head directly facing the implanted antenna coil to allow for the coupling of the coils to enable energy and data to be transferred between the external and internal antenna coils. The transfer of energy is controlled to effect the transmission of the coded sound signal and power from the external speech processor to the implanted stimulator unit, and to effect the transmission of telemetry data from the implanted stimulator unit to the external speech processor.

SUMMARY

In accordance with one aspect of the present invention, there is provided a transcutaneous data communications link for am implantable medical device, the transcutaneous data communications link comprising:

at least two capacitors, each capacitor having an externally located electrode and an implanted electrode separated by a transcutaneous dielectric;

an implanted data receiver in communication with each one of the at least two capacitors;

an externally located voltage driver arranged to communicate a differential data signal to the implanted data receiver through the at least two capacitors.

BRIEF DESCRIPTION OF THE DRAWINGS

[0008] FIG. 1 is a perspective view of internal and external components of a cochlear implant system shown in their operational position on a recipient.

[0009] FIG. 2 is a simplified schematic diagram a capacitive data link in accordance with one embodiment of the present invention.

[0010] FIG. 3 is a detailed schematic diagram a capacitive data link in accordance with one embodiment of the present invention.

[0011] FIG. 4 is a simplified schematic diagram of a circuit for driving an external coil of a transcutaneous transfer system.

[0012] FIG. 5 is a simplified schematic diagram of a convention circuit for coupling out telemetry data.

[0013] FIG. 6 is a simplified schematic diagram of a convention circuit for coupling out telemetry data.

[0014] FIG. 7 is a simplified schematic diagram of a circuit for coupling out telemetry data in accordance with another embodiment of the present invention.

[0015] FIG. 8 is a simulation circuit diagram of the embodiment of the circuit illustrated in FIG. 4 in accordance with another embodiment of the present invention.

[0016] FIGS. 9A-9F are a series of waveforms taken at various nodes in the simulation circuit illustrated in FIG. 8.

[0017] FIG. 10 is a simulation circuit diagram of the embodiment of the circuit illustrated in FIG. 6 when the external coil experiences 100mV amplitude telemetry pulses, in accordance with another embodiment of the present invention.

[0018] FIGS. 11A-11B are waveforms taken at various nodes in the simulation circuit illustrated in FIG. 10.

DETAILED DESCRIPTION

[0019] Embodiments of the present invention are directed to the transcutaneous transfer of data using a capacitive link thereby providing for the low-power transmission of data across the skin of a patient without a galvanic connection. Embodiments of the present invention are described below in connection with one embodiment of a hearing implant commonly referred to as a cochlear™ prosthesis (also referred to as a cochlear™ implant system, cochlear™ prosthetic device and the like; generally referred to herein as a "cochlear implant system"). As used herein, the term "cochlear implant system" refers to any partially- or completely-implantable devices that provide electrical stimulation and/or mechanical stimulation to a patient to improve and/or provide hearing sensations. It should be appreciated, however, that the present invention may be implemented in connection with other types of medical implants.

[0020] Cochlear prostheses use direct electrical stimulation of auditory nerve cells to bypass absent or defective hair cells that normally transducer acoustic vibrations into neural activity. Such devices generally use multi-contact electrodes inserted into the scala tympani of the cochlea so that the electrodes may differentially activate auditory neurons that normally encode differential pitches of sound. Such devices are also used to treat a smaller number of patients with bilateral degeneration of the auditory nerve. For such patients, a cochlear prosthetic device provides stimulation of the cochlear nucleus in the brainstem.

[0021] Exemplary cochlear prostheses in which embodiments of the present invention may be implemented include, but are not limited to, those systems described in U.S. Patent Nos. 4,532, 930, 6,537,200, 6,565,503, 6,575,894 and 6,697,674, which are hereby incorporated by reference herein. FIG. 1 is a schematic diagram of an exemplary cochlear implant system 100 in which embodiments of the present invention may be implemented. Cochlear implant system 100 comprises external components 142 which are directly or indirectly attached to the body of the recipient, and internal components 144 which are temporarily or permanently implanted in the recipient. External components 142 typically comprise a microphone 120 for detecting sounds, a speech processor 116 that converts the detected sounds into a coded signal, a power source (not shown), an external transmitter unit 106, and an external portion of a capacitive data link system (not visible in FIG. 1). External transmitter unit 106 comprises an external coil 108 and a magnet 110 secured directly or indirectly to external coil 108. Speech processor 116 processes the output of microphone 120 that is positioned, in the depicted embodiment, on the ear 122 of the recipient.

Speech processor 116 generates a coded signal which is provided to external transmitter unit 106 via cable 118.

[0022] Internal components 144 comprise an internal receiver unit 112, a stimulator unit 126, an electrode array 134, and an internal portion of the above-noted capacitive data link system (also not visible in FIG. 1). Internal receiver unit 112 comprises an internal receiver coil 124 and a magnet 140 fixed relative to internal coil 124. Internal receiver unit 112 and stimulator unit 126 are hermetically sealed within a housing 128. Internal coil 124 receives power from transmitter coil 108. A cable 130 extends from stimulator unit 126 to cochlea 132 and terminates in an electrode array 134. The received signals are applied by array 134 to the basilar membrane 136 thereby stimulating the auditory nerve 138.

[0023] Collectively, transmitter antenna coil 108 (or more generally, external coil 108) and receiver antenna coil 124 (or, more generally internal coil 124) form an inductively-coupled transcutaneous transfer apparatus 102. Transmitter antenna coil 108 transmits energy to the implantable receiver coil 124 via a radio frequency (RF) link 114. Internal coil 124 is typically a wire antenna coil comprised of at least one and preferably multiple turns of electrically insulated single-strand or multi-strand platinum or gold wire. The electrical insulation of internal coil 124 is provided by a flexible silicone moulding (not shown). In use, implantable receiver unit 112 can be positioned in a recess of the temporal bone adjacent ear 122 of the recipient.

[0024] As noted, cochlear implant system 100 comprises a capacitive data link system. A simplified schematic diagram of one embodiment of such a capacitive data link system to transmit data between internal and external components 142, 144 is depicted in FIG. 2. As shown therein, a capacitive data link system 200 comprises external components 202 and internal components 204. External components 202 are worn by the patient, for example, integrated into speech processor 116 or as a separately-worn unit connected to speech processor 116 by a cable. Internal components 204 are implanted in the recipient at a location in which a capacitive link can be established, as described herein. Internal components 204 are operatively coupled to stimulator unit 126, as shown in FIG. 2.

[0025] Briefly, external components 202 comprise an external electrode 212E of each of two capacitors 206A and 206B (collectively and generally referred to herein as capacitors 206). External components 202 also comprise a voltage driver 208 which generates a biphasic voltage signal 210 to differentially drive external electrodes 212E of capacitors 206 as described herein.

[0026] Internal components 204 comprise an internal electrode 212I of each capacitor 206. Each electrode 212I is connected to one input of a differential amplifier 214 through a resistive network 218. It should be appreciated that the embodiment illustrated in FIG. 1 is a simplified schematic. For example, as one of ordinary skill in the art would appreciated, internal components 204 would typically include signal conditioning circuitry to convert data output signal 216 generated by differential amplifier 214 to a form suitable for use by internal components of system 100. Such signal conditioning circuitry may include, for example, a comparator, pulse forming circuitry and related circuitry and/or other circuitry to amplify and shape data out signal 216 as required for the particular application.

[0027] Capacitors 206 each comprise oppositely-spaced electrodes 212E and 212I; that is, the opposing electrodes 212 of each capacitor 206 are aligned with each other along an axis line substantially orthogonal to planes defined by electrodes 212. Such transcutaneous alignment facilitates the capacitive coupling attained in each capacitor 206 during operation of capacitive data link system 200.

[0028] External electrodes 212E are adjacent to and preferably not in contact with skin 146 of the recipient. Accordingly, external electrodes 212E may be encased in a housing formed of a suitable dielectric material. Such housing may provide a desired separation between external electrodes 212E and the recipient.

[0029] Internal electrodes 212I, on the other hand, are galavanically isolated from the body of the recipient for electrical reasons and also for biocompatibility. As such, internal capacitor plates 212I may be encapsulated in, for example, a silicon or silastic film.

[0030] Internal and external electrodes 212 may be formed of any conductive material and may have any dimensions suitable for a particular application. For example, in one embodiment, electrodes 212 comprise a conductive material such as copper or platinum metal and are formed as a flexible coil or film. Thus, should be appreciated that capacitors 206 can be implemented with any conductive material having any dimensions suitable for achieving a capacitive link given the particular patient and where on the patient the capacitor is located. It should also be appreciated that the materials used form the external electrode of a capacitor 206 need not be the same as the materials used to form the internal electrode of that same capacitor 206.

[0031] Preferably, external electrode 212E and internal electrode 212I of each electrode 206 have the same dimensions and surface area. Capacitors 206 are preferably as large as possible to facilitate signal coupling, while taking into consideration the limits imposed on capacitor size due to the size of the recipient's head, the distance between opposing electrodes 212 of each capacitor 206, etc. In one embodiment, electrodes 212 are rectangular and have a surface area of approximately $1cm^3$. It should be appreciated, then, that the surface area and dimensions of electrodes 212 may vary depending on the requirements of the particular application.

[0032] As one of ordinary skill in the art would appreciate, the capacitance of each capacitor 206 is determined by a number of factors such as the dimensions and spacing of its electrodes 212, and the material, here, skin and perhaps hair, between the electrodes of the capacitor. In some embodiments in which capacitors 206 are designed for use in connection with a cochlear implant system such as system 100 introduced above, the capacitance of each capacitor 206 is in the range of approximately 0.1pf – 0.5pf. In alternative embodiments implemented in connection with the same or different application, the capacitance of each capacitor 206 may be different, and based on a variety of factors including the distance and material between electrodes 212.

[0033] As noted, external components 202 include a voltage driver 208. Voltage driver 208 generates differential voltage signal 210 to generate an electric field change on internal electrodes 212I of capacitors 206. Preferably voltage driver 208 generates a pulse waveform, although any biphasic waveform such as a sinusoidal waveform, may be used to differentially drive capacitors 206. In one embodiment, voltage driver 208 generates a 5 volt signal for the implemented TTL circuitry. It should be appreciated, however, that any suitable voltage signal generated by any voltage source now or later developed can be used in alternative embodiments. For example, in one alternative embodiment, voltage driver 208 generates a 3 volt signal.

[0034] In one embodiment, capacitive data link system 200 is powered, for example, by a battery. In such embodiments, the amplitude of voltage signal 210 may be limited. In alternative embodiments, a voltage signal 210 having a greater amplitude can be provided to support greater signal strength. As one of ordinary skill in the art would appreciate, such a voltage boost will likely consume additional power and, therefore require some trade-offs.

[0035] As noted, internal electrodes 212I of capacitors 206 are connected to respective inputs of a discrete differential amplifier 214 through a resistive network 218. Differential amplifier 214 amplifies the difference in electric potential between the two inputs. In this way, common mode variations caused not by external sources are substantially isolated. Preferably, differential amplifier 214 is a transistor differential amplifier implementing JEFTs due to its high input resistance and low input capacitance. As one of ordinary skill in the art would appreciate, differential amplifier 214 may be implemented in a variety of ways.

[0036] Resistive network 218 is provided to adjust the input impedance of differential amplifier 214. In one embodiment, resistors 220A and 220B are approximately 1 MOhm. It should be appreciated that the values of resistors 220 may be selected based on conventional design considerations well-known to those of ordinary skill in the art. In the above exemplary embodiment, the resulting differential voltage across the inputs of amplifier 214 is approximately 20 mV.

[0037] As understood by those of ordinary skill in the art, the current through capacitors 206 is determined by the rise time and the height of voltage signal 210 generated by voltage driver 208. This also determines the amplitude of the data out signal 210. The current through capacitors 206 is also proportional to the size of electrodes 212. As a result, the voltage provided to the inputs of differential amplifier 214 is approximately proportional to its input impedance and this current.

[0038] The present invention advantageously allows for the functional separation of data and power transmission, enabling each to be optimally configured without concern for the potential adverse effects on the other type of transmission. In one embodiment of the present invention, the data rate is approximately 1 megabit per second, or 1 megahertz. It should be appreciated, however, that the data rate can be significantly higher or lower should a different data bandwidth be required.

[0039] FIG. 3 is a detailed schematic diagram of one exemplary embodiment of capacitive data link system 200.

[0040] It should be appreciated that just a few embodiments of the present invention have been described herein. For example, it is contemplates that the capacitive data link system 200 can be extended for bi-directional data transmission. A bi-directional half duplex data link may be achieve, for example, with the addition of a multiplexer and additional driver and receiver components.

[0041] One advantage of certain embodiments of the present invention is that high data transmission rates can be achieved with low power usage. In one embodiment for example, the transmission rate is 1 MHz. It should be appreciated, however, that the transmission rate is determined by a number of factors including, but not limited to, the skin and the hair that are located between the external and internal plates of each of the capacitors 206. A further advantage is that the magnetic power link, as noted, could be optimized purely for power alone, not having to perform the function of data transmission.

[0042] **Coupling Out Telemetry Data In A Transcutaneous Transfer System**

[0043] In another arrangement according to this disclosure, there is described a circuit for driving an external coil of a transcutaneous transfer system. The driver circuit enables the transcutaneous transfer system to automatically switch between a telemetry mode in which data is transferred from an implanted coil to the external coil, and a normal transmit mode in which data and/or power is/are transferred from the external coil to the internal coil. As a result, embodiments of the present invention advantageously need not receive nor be responsive to an external mode control signal.

[0044] FIG. 4 is a simplified schematic diagram of one embodiment of an external coil circuit 200. Circuit 200 is integrated into external transmitter unit 106. As shown in FIG. 4, circuit 200 comprises a variety of components including an indicator 202 which serves as external coil 108 shown in FIG. 1.

[0045] External coil circuit 200 implements a zero volt switching (ZVS) technique to drive transmitter coil 202. A switch 204 in the form of a MOSFET is connected in series between a negative terminal of inductor 202 and ground. The gate of MOSFET 204 is controlled by an externally-controlled signal referred to as RFDATA signal 206. A tuning capacitor 208 is connected in parallel with MOSFET 204. Finally, a positive node of inductor 202 is connected to a supply voltage 210. A coupling capacitor 212 is provided to filter power supply signal 210.

[0046] When transmitting data to internal coil 124, speech processor 106 generates RFDATA signal 206 to switch MOSFET 204 on and off at, for example, 5MHz. This drives the resonant tank (inductor 202 and tuning capacitor 208).

[0047] When transmitting telemetry data from internal coil 124 (FIG. 1) to external coil 202, the internal coil is pulsed. Such pulses appear as brief intervals of increased voltage across inductor 202. For example, in one embodiment of hearing prosthesis 100, such voltages are approximately 100mV, and have a pulse width of 1 to 3 microseconds.

[0048] To process (amplify, count etc.) these pulses, they have to be coupled out in some way. This can be achieved either at the top or at the bottom of transmitter coil 202. However, the top terminal of coil 202 is bypassed by capacitor 212 for voltage supply decoupling. Accordingly, bypass capacitor 212 serves as a path to ground for signal frequencies. Thus, to couple out the telemetry data at the topside of coil 202, the positive terminal of coil 202 must be at high impedance with respect to ground for the signal frequencies.

[0049] On the other hand, the bottom terminal of inductor 202 experiences large voltage transients with respect to ground during active data transmission from external coil 202 to internal coil 124. Such transients of, for example, 50V, make it difficult to couple this signal in to an amplifier without excessive power losses. Thus, to couple out the telemetry data at the bottom side of coil 202, the large voltage transients found on that terminal during normal data transmission mode must be managed.

[0050] FIG. 5 is a simplified schematic diagram of one embodiment of the present invention. In this embodiment, circuit 300 is a 5-wire circuit in which the telemetry is coupled out at the topside of coil 202. As noted, to achieve this, this side of coil 202 must be maintained at a high impedance with respect to ground for signal frequencies. In circuit 300, a second MOSFET 308 (P-channel) decouples the topside of coil 202 from bypass capacitor 306. A signal, TELEON 304 is required to be generated by speech processor 106 to control the state of MOSFET 308. Unfortunately, it is not always possible to provide such a control signal to circuit 300. As will be described in detail below, the embodiments of the invention described below couple out the telemetry without switching the mode from data transmitter to data receiver.

[0051] FIG. 6 is a simplified schematic diagram of one embodiment of the present invention. In this embodiment, the voltage at the negative terminal of inductor 202 is clamped. This protects any amplifier that receives telemetry signal 402 while clamping the large voltage transients leading to large losses in series resistor 422. However, it should be appreciated that increasing series resistor 422 is not preferred because the series resistor then forms a low pass filter together with the capacitance of diodes 416 and 418 of clamping circuit 420 and the telemetry amplifier's input capacitance. This may degrade telemetry signal 402.

[0052] FIG. 7 is a simplified schematic diagram of one embodiment of a circuit 500 implemented in external transmitter circuit 106 of transcutaneous transfer system 100. Circuit 500 comprises the circuit arrangement described above with reference to FIG. 4 in addition to the following components. Significantly, circuit 500 does not clamp (dissipate) large transients as in circuit 400.

Rather, circuit 500 blocks (does not pass) large voltage transients, without dissipation, and only outputs relatively small amplitude signals to telemetry output line 502.

[0053] There are two modes of operation of circuit 500, passive mode and active mode. In passive or receiver mode, telemetry data may be received from internal coil 124 (FIG. 1). In active or transmitter mode, data is sent to internal coil 124. Each of these will be described in turn. In passive mode, RFDATA 206 is at zero volts since no data is being transmitted. As a result, the drain 542 of MOSFET 404 is at VSUPPLY 210. The current flowing through diode 526 is determined by resistors 524 and 528. The current flowing through diode 536 is determined by resistors 534 and 532.

[0054] In passive mode, the anode of diodes 526 and 536 are both close to VSUPPLY 210, and both diodes are slightly forward biased. The forward biased diodes 526, 536 act as low dynamic resistances. Hence a signal path is provided through diodes 526, 536, and, coupling capacitors 530, 538, to telemetry output 502 of circuit 500.

[0055] In active mode, on the negative edge of RFDATA signal 206, MOSFET 204 goes out of conduction and coil 202 starts charging tuning capacitor 208. In effect the drain node 542 of MOSFET 204 rises rapidly to several tens of volts (close to 50V).

[0056] Diode 526 and capacitor 530 pass this high positive transient voltage on to the cathode of diode 536. However, diode 536 blocks the transient voltage, and the output 502 experiences only a small positive transient the size of a diode voltage drop.

[0057] On the positive edge of RFDATA signal 206, the drain node 542 drops from VSUPPLY 210 to GND 214, a decrease of approximately 10V in one embodiment. This negative transient causes diode 526 to block, since the cathode voltage is kept almost constant due to capacitor 530, diode 536 and resistor 532. Thus only a small negative transient the size of a diode drop is experience by output 502. Thus, very large voltage transients are blocked, and thus power is saved.

[0058] And, as noted, no control signal is required to change the operating modes of circuit 500. Rather, speech processor 116 only needs to holds RFDATA 406 low for a short period to permit circuit 500 to settle down and become ready to receive telemetry data from internal coil 124.

[0059] FIG. 8 is a simulation of the circuit 500, with nodes identified throughout the circuit. The waveforms occurring at the identified nodes are depicted FIGS. 9A-9F.

[0060] FIG. 10 is a schematic diagram of a simulation showing an telemetry output of 50mV for a 100mV input signal. It should be appreciated that increasing the bias current of diodes 526 and 536 can increase this. FIGS. 11A-11B are waveforms generated by the simulation at coil 202 (node 1 in FIG. 10) and at telemetry output 402 (node 2 in FIG. 10).

[0061] It will be appreciated by persons skilled in the art that numerous variations and/or modifications may be made to the invention as shown in the specific embodiments without departing from the spirit or scope of the invention as broadly described. The present embodiments are, therefore, to be considered in all respects as illustrative and not restrictive.

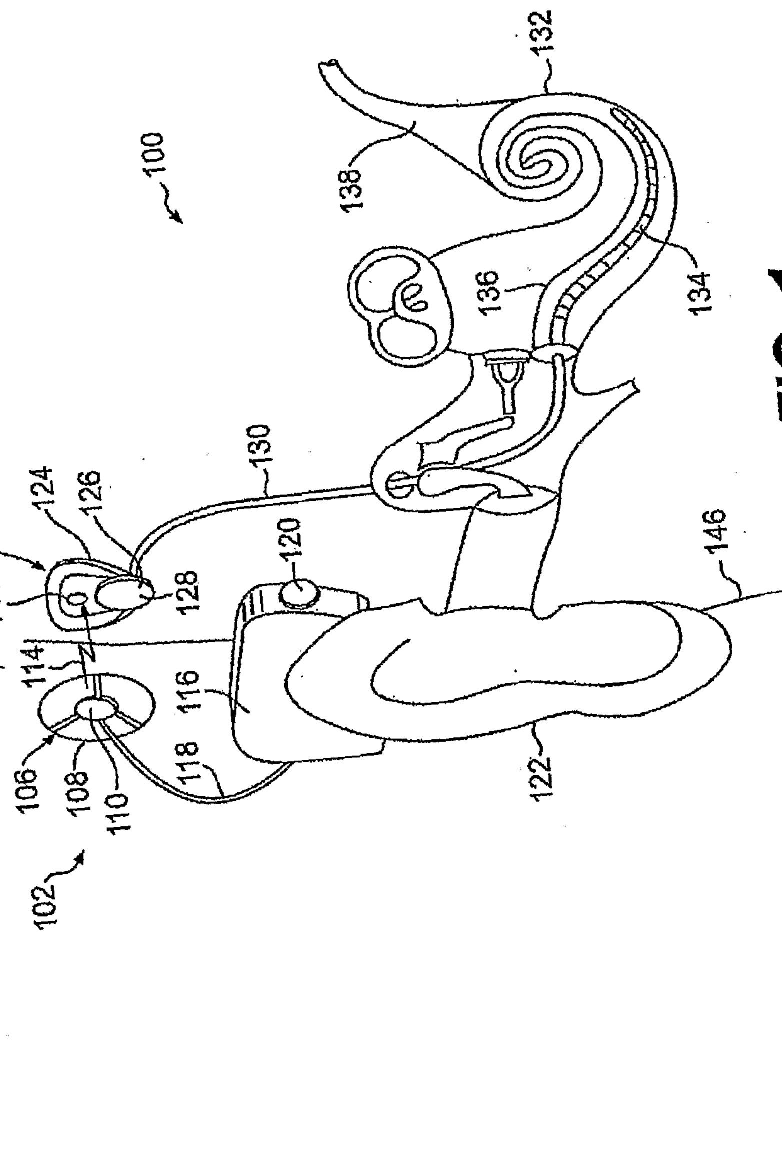
CLAIMS

1. A transcutaneous data communications link for am implantable medical device, the transcutaneous data communications link comprising:

at least two capacitors, each capacitor having an externally located electrode and an implanted electrode separated by a transcutaneous dielectric;

an implanted data receiver in communication with each one of the at least two capacitors;

an externally located voltage driver arranged to communicate a differential data signal to the implanted data receiver through the at least two capacitors.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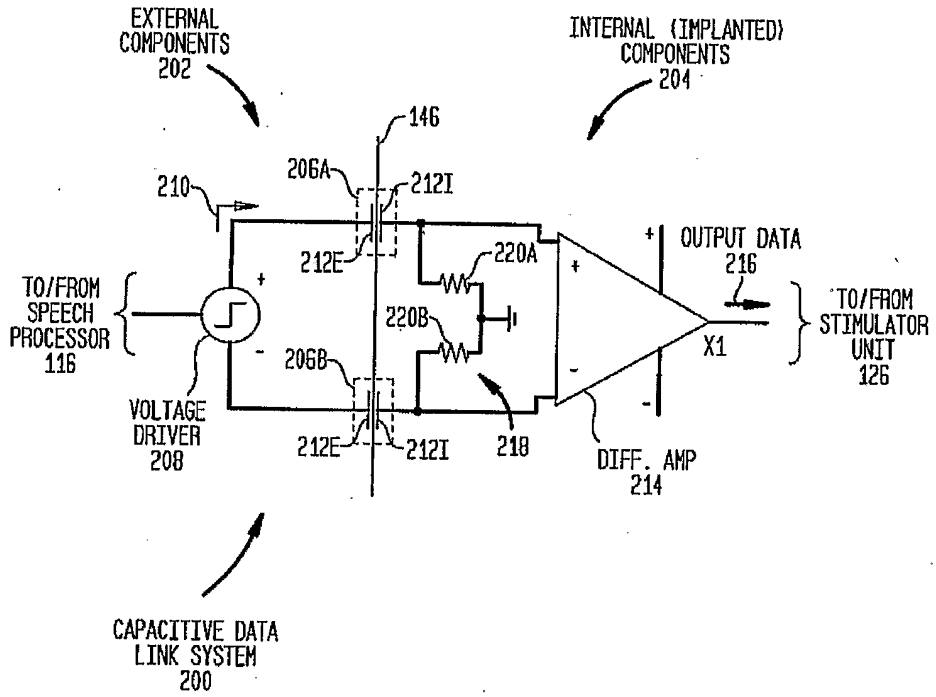
146

FIG. 1

FIG. 2


EXTERNAL
COMPONENTS
202
INTERNAL (IMPLANTED)
COMPONENTS
204
146
206A
212I
210
212E
220A
OUTPUT DATA
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TO/FROM
SPEECH
PROCESSOR
116
220B
X1
TO/FROM
STIMULATOR
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LINK SYSTEM
200

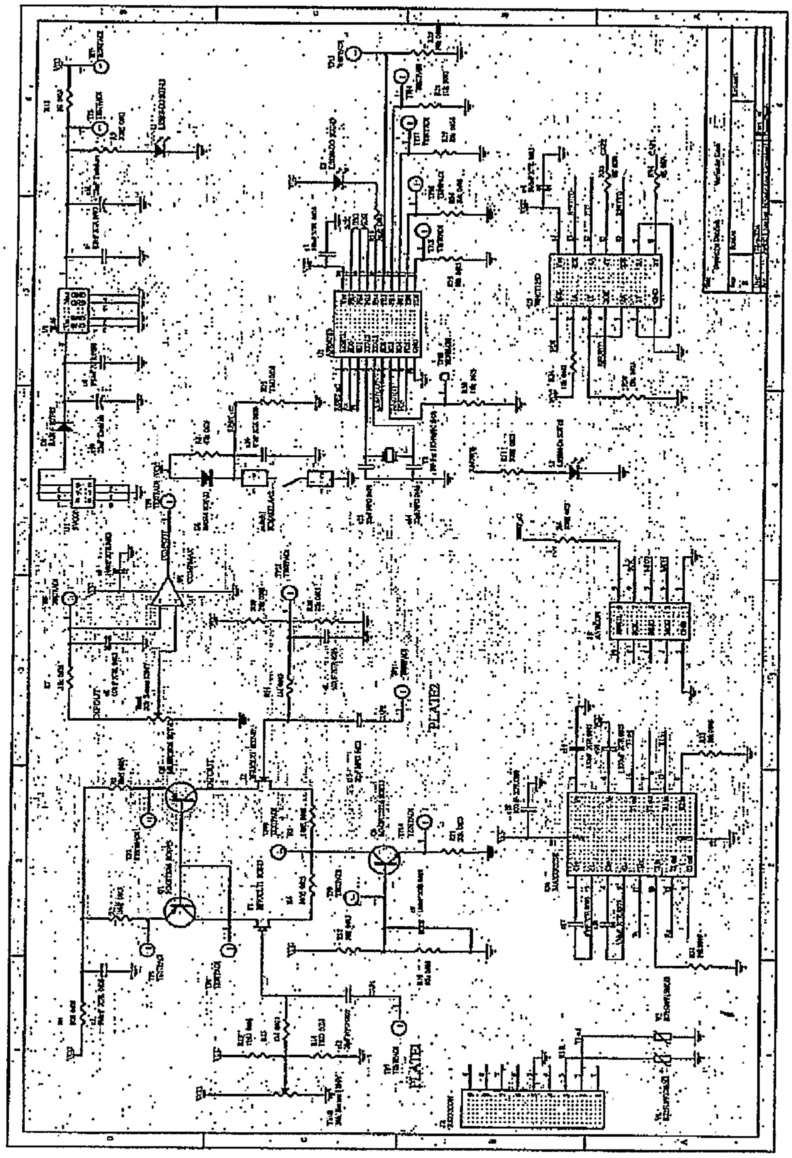
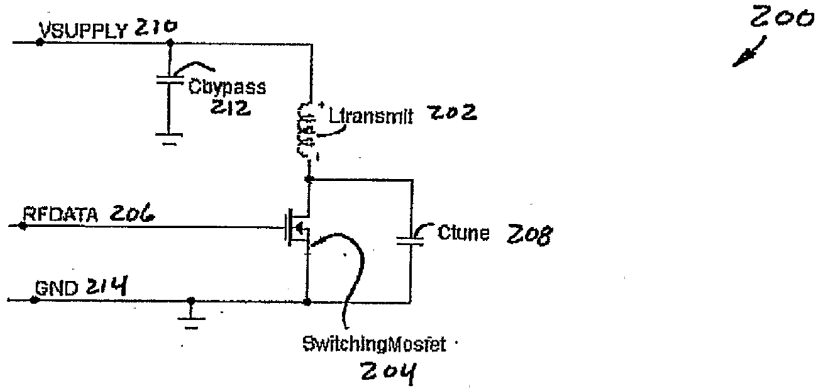


FIG. 3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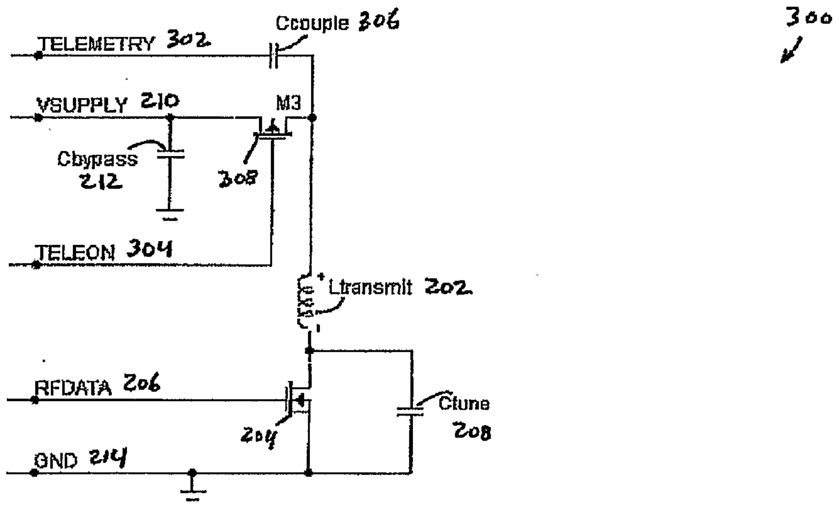
RFDATA 206
Ctune 208
GND 214
SwitchingMosfet 204

FIG. 4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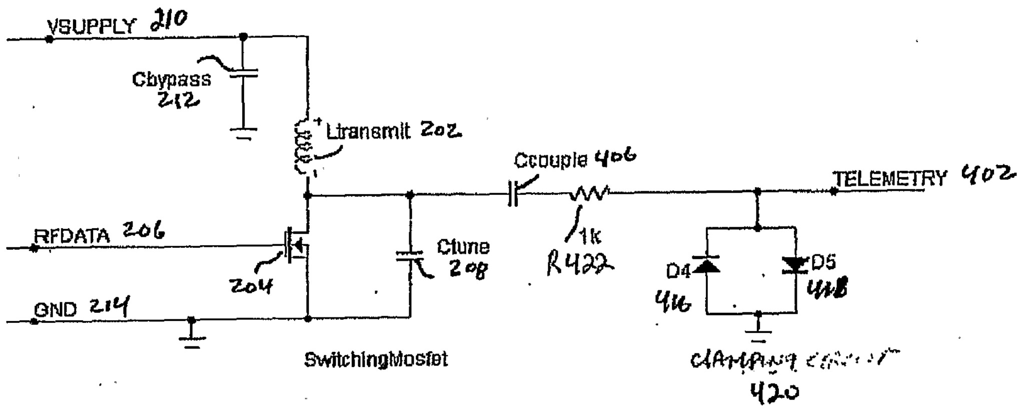
Ctune 208
GND 214

FIG. 5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418
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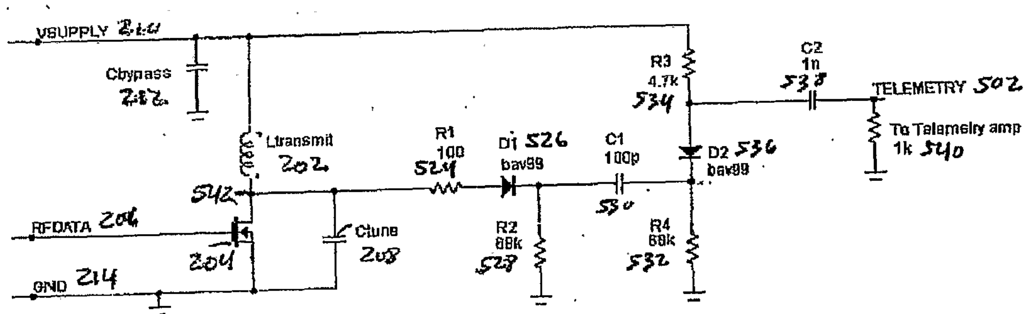
420

FIG. 6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538
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100
524
D1 526
bav99
C1
100p
530
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bav99
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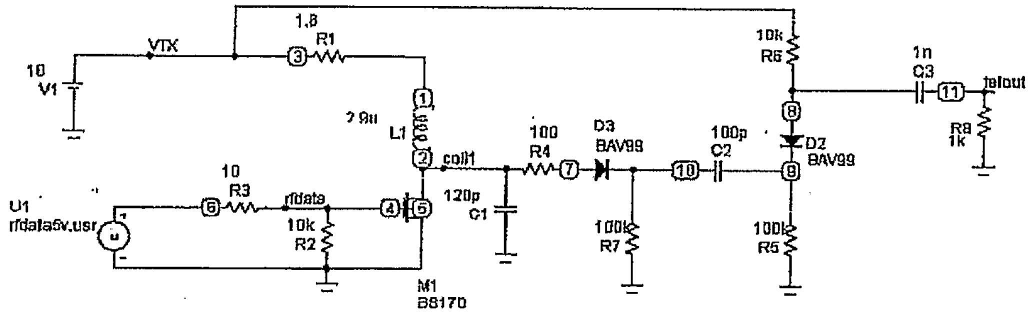
532
GND 214
500

FIG. 7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VTX
10
V1
3
1
2.9u
L1
2
coll1
100
R4
7
D3
BAV99
10
100p
C2
9
10k
R6
8
D2
BAV99
1n
C3
11
telout
R8
1k
10
R3
6
rfdata
4
5
U1
rfdata5v.usr
10k
R2
M1
BS170
120p
C1
100k
R7
100k
R5

FIG. 8

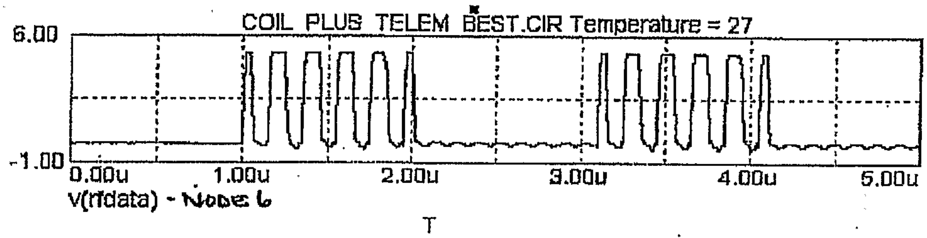

FIG. 9A

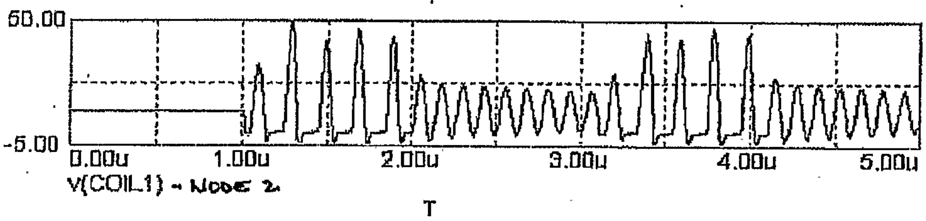

FIG. 9B

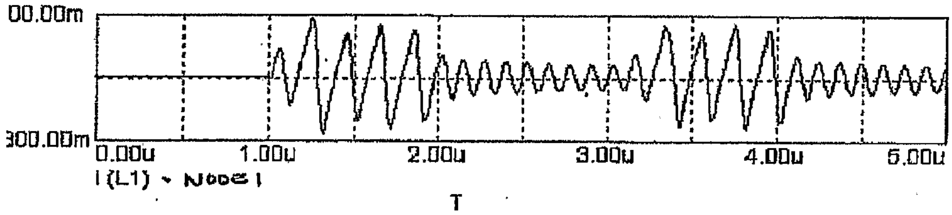

FIG. 9C

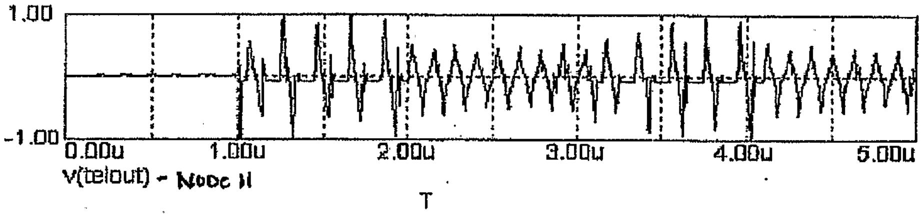

FIG. 9D

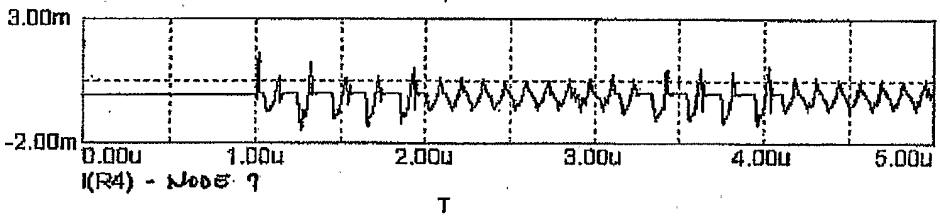

FIG. 9E

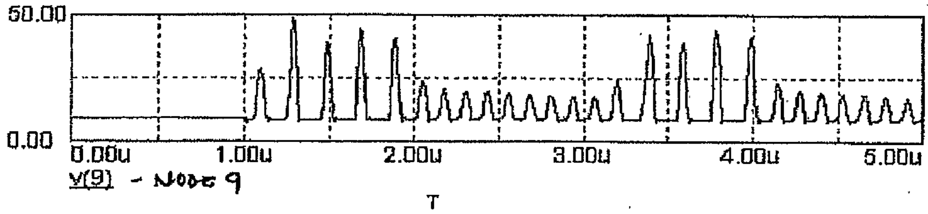

FIG. 9F

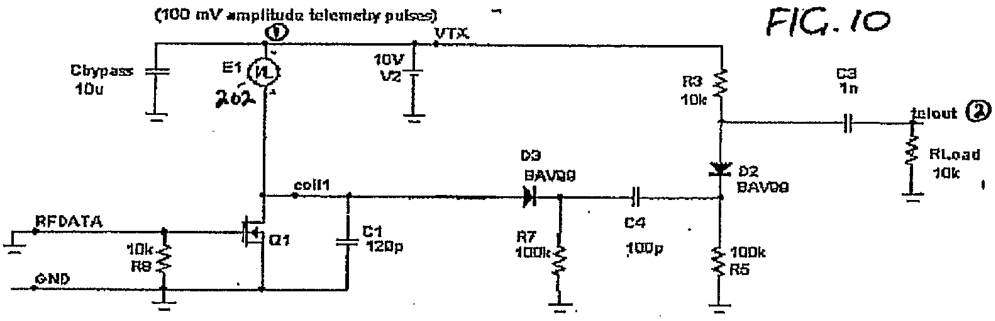

FIG. 10

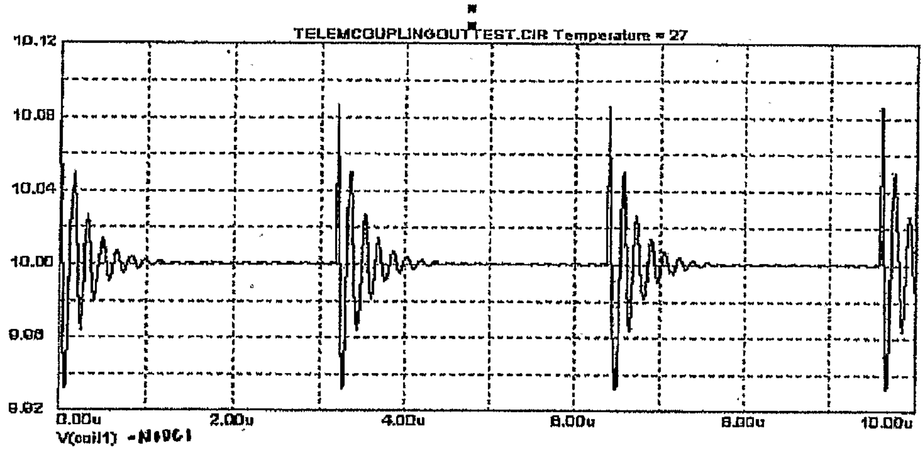

FIG. 11A

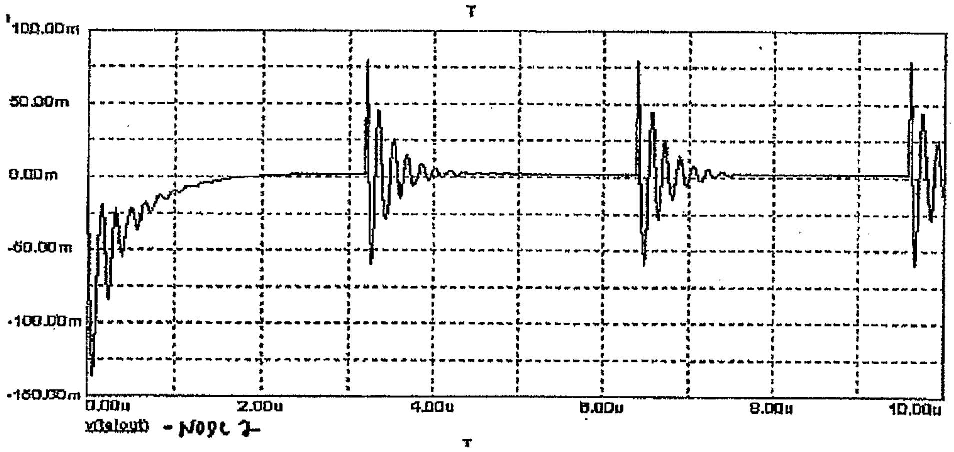

FIG. 11B