

[54] PORTABLE ELECTROANESTHESIA DEVICE WITH AUTOMATIC POWER CONTROL

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[58] Field of Search 128/1 C, 422

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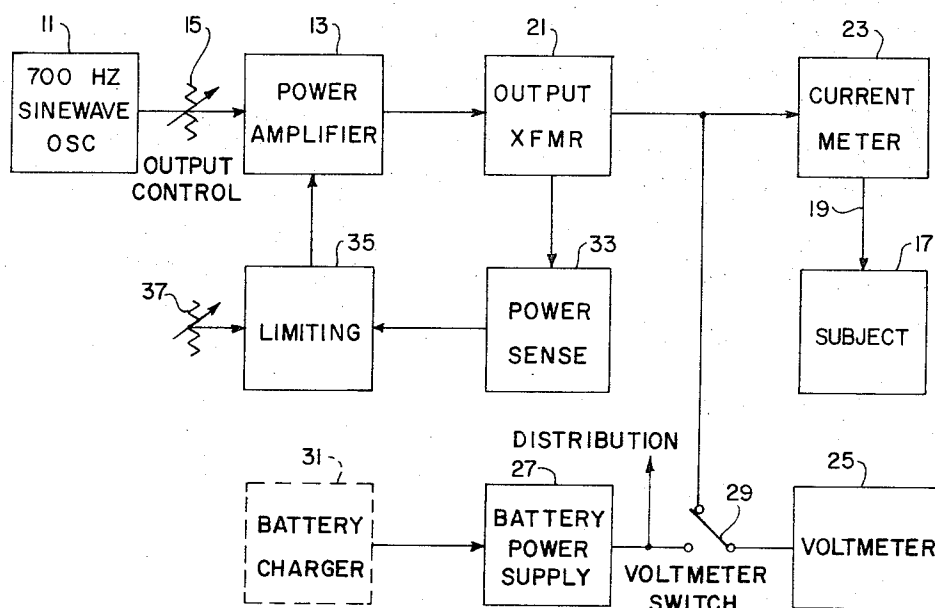
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[57] ABSTRACT

A portable, battery-powered, self-contained electroanesthesia device. A signal generator provides a sinusoidal, constant magnitude signal of fixed, preset frequency with relatively high accuracy and without recalibration each time the device is energized. The preset frequency characteristically produces electroanesthetic induction in a live subject with relatively high efficiency. An amplifier increases the power level sufficiently for producing electroanesthetic induction of the subject. Electrodes supply the amplified signal to tissue of the subject, there being provision for matching the impedance of said tissue of the subject to the amplifier for maximizing transfer of the amplified signal to said tissue. Circuitry is provided also for monitoring and limiting the power level of the signal supplied by said electrodes to the tissue.

7 Claims, 3 Drawing Figures



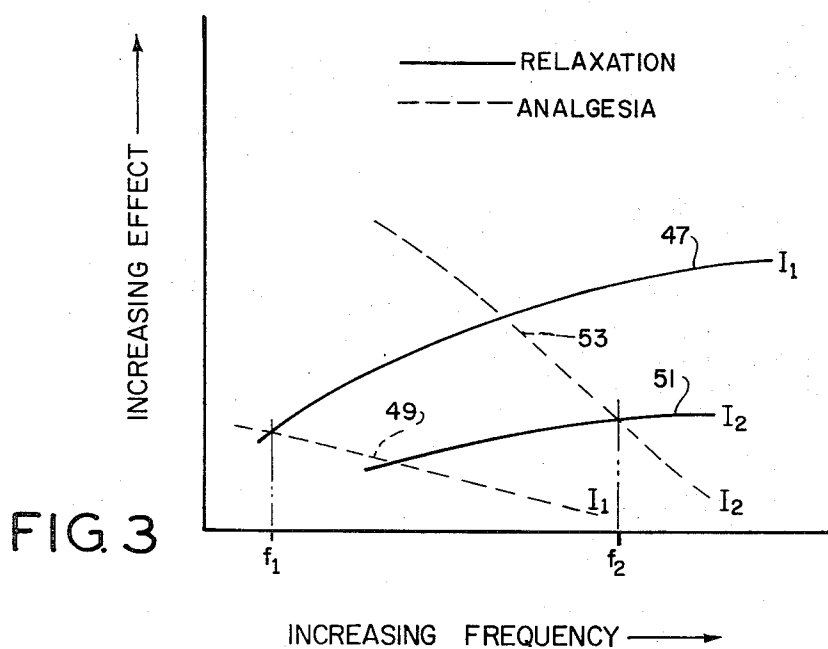
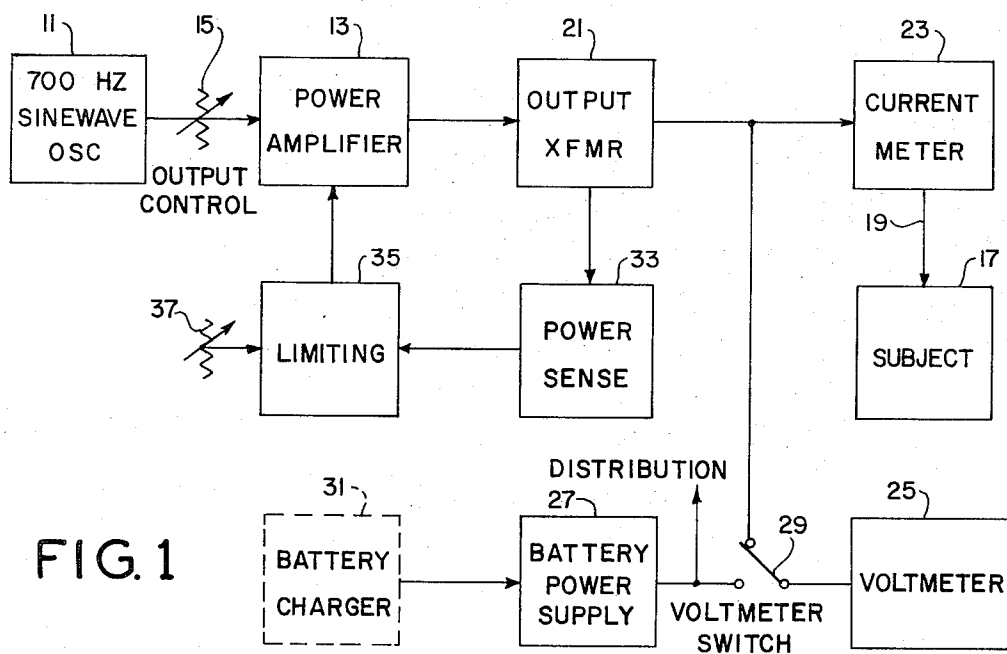
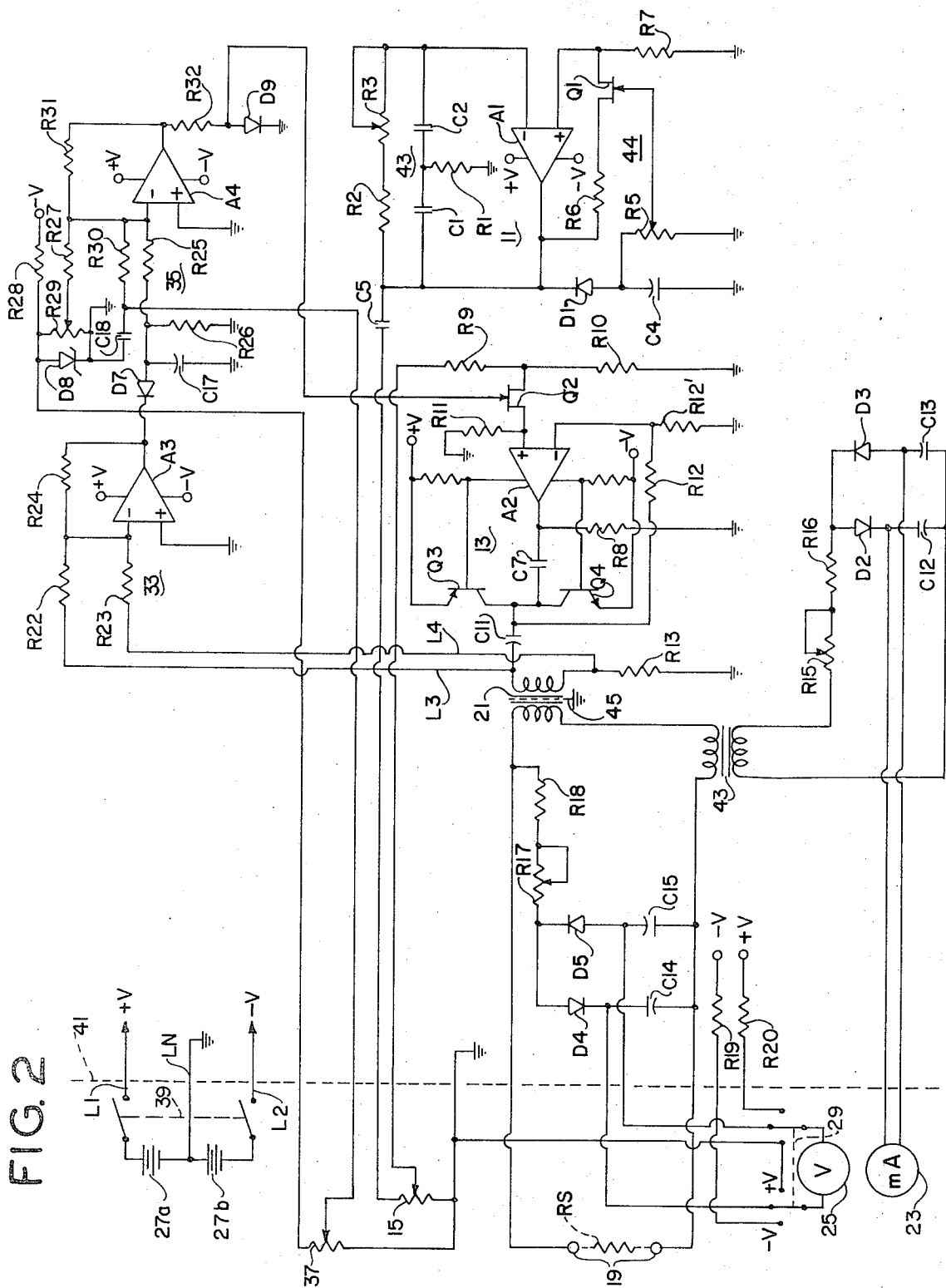


FIG. 2



PORTABLE ELECTROANESTHESIA DEVICE WITH AUTOMATIC POWER CONTROL

BACKGROUND OF THE INVENTION

This invention relates to electrically induced anesthesia, i.e., electroanesthesia, and more particularly to devices for supplying a periodically varying current to tissue of a living subject for causing anesthesia.

It has long been proposed to produce anesthesia in human subjects as well as in lesser animals by causing an electric current to flow through tissue of the subject and various techniques for accomplishing this have been under investigation for many years. The literature is replete with many reports of such investigations.

The use of alternating currents of various single or mixed frequencies and of various waveforms, such as square-wave, triangular-wave, and so forth, has been suggested. However, it has been observed that a frequency of 700 Hz provides satisfactory electroanesthetic induction in a human subject. The use of a sinusoidal signal at such a frequency has been suggested in an article by William B. Wood et al., entitled "The Cardiovascular Effects of Cranially-Impressed Electric Currents of Anesthetic Intensity" in *Anesthesia and Analgesia Current Researches*, Vol. 43, No. 3 (May-June), 1964. It has also been observed that a sinusoidal signal of substantially exactly 700 Hz or a multiple thereof characteristically produces electroanesthetic induction in a subject with relatively high efficiency. That is, less current (and thus less power) is required to effect anesthesia at such a frequency than at other frequencies.

Devices of an experimental nature have been constructed for supplying signals of various waveforms and frequencies. An electroanesthesia instrument has also been commercially available, this instrument generating two sinusoidal signals, one of a quite low frequency and the other of a higher variable frequency, the signals being mixed together to provide a signal of complex waveform which is amplified for being applied to electrodes secured to (or embedded in) tissue of a subject.

Devices heretofore suggested or available have not been suitable for other than laboratory or operating room use because they have not, as a practical and realistic matter, been truly portable in character. This is so either because such devices have required an a.c. utility power connection (with consequent risk to the subject of shock, etc.) or, at best, have needed heavy and cumbersome storage batteries in the absence of utility power because of their considerable power requirements. Thus, prior art devices have not been self-contained, as required for true portability (and as is important for field or emergency use).

Moreover, because the frequency supplied by such prior art devices is subject to variation, the amount of power required to effect anesthesia is also subject to variation to the extent that, at some frequencies, considerable power may be required to achieve anesthesia, further gravitating against portability. Furthermore, this has required that the user establish by trial and error a frequency producing the most satisfactory anesthesia. Alternatively, recalibration of the device has been required if it be desired, for example, to produce a signal at the highly effective frequency of 700 Hz.

BRIEF SUMMARY OF THE INVENTION

Among the several objects of the invention may be noted the provision of an electroanesthesia device which is truly portable in character, which is battery powered, and which is self contained; the provision of such a device which does not require heavy, cumbersome storage batteries but is instead powered by small rechargeable cells contained within the device; the provision of such a device adapted to provide electroanesthetic induction of a subject with high efficiency and which, for this purpose, does not require recalibration each time it is energized; the provision of such a device which, in use, does not constitute a hazard for the subject; the provision of such a device which prevents excessive power from being supplied to the subject thereby to protect the subject; the provision of such a device which is not prone to damage if output electrodes thereof become shorted; the provision of such a device which employs semiconductor circuitry, which is simply and inexpensively constructed, and which is long lasting and reliable in operation. Other objects and features will be in part apparent and in part pointed out hereinafter.

Briefly, a portable, battery-powered and self-contained electroanesthesia device includes means for generating a signal of substantially sinusoidal waveform at a fixed, preset frequency with relatively high accuracy and without recalibration each time said device is energized, the preset frequency being such as will characteristically produce electroanesthetic induction in a live subject with relatively high efficiency. The sinusoidal signal is of substantially constant magnitude. Means is provided for amplifying the signal to a power level suitable for producing electroanesthetic induction of the subject. The device includes means for supplying the amplified signal to the tissue of the subject to cause electroanesthetic induction of the subject as well as means for matching the impedance of said tissue of the subject to the amplifying means for maximizing transfer of the amplified signal to this tissue. The device further comprises circuitry for monitoring the power level of the signal supplied by said means to said tissue. Preferably, the power monitoring circuitry comprises means for automatically limiting the power supplied to the subject to a preset maximum level.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is a block diagram of circuitry of a portable, battery-powered electroanesthesia device of the invention;

FIG. 2 is a schematic diagram of circuitry of the preferred embodiment; and

FIG. 3 is a graph useful in explaining operation of the device;

Corresponding reference characters indicate corresponding parts throughout the several views of the drawings.

DESCRIPTION OF THE PREFERRED EMBODIMENT

Referring now to FIG. 1, a block diagram illustrates major components of a portable battery-powered electroanesthesia device of the present invention. At 11 is indicated an oscillator, i.e., a signal generator, providing a substantially sinusoidal, constant magnitude signal at a fixed, preset frequency, preferably substantially

precisely 700 Hz. This frequency characteristically produces electroanesthetic induction in a live subject with relatively high efficiency. This sinusoidal-waveform 700 Hz signal is provided each time the circuit is energized without any need for recalibration of the device. Oscillator 11 is preferably solid state in design, the output signal therefrom being supplied to a solid state power amplifier circuit 13. Manually variable impedance means 15 is connected between oscillator 11 and amplifier 13 for controlling the power level of the signal supplied by electrodes to the electroanesthesia subject 17 by suitable electrodes 19. Also included is an impedance-matching or output transformer 21 having a turns ratio preselected to match the effective impedance of the tissue of subject 17 across which the electrodes are applied to the output impedance of amplifier 13, thereby to maximize transfer of the amplified signal to tissue of subject 17.

The circuit includes means for monitoring the power level of the signal supplied by electrodes 19 to tissue of subject 17. The latter means may be regarded as comprising, in a first aspect, a current meter 23 and a voltmeter 25 which together measure and indicate the magnitude of the current and the voltage of the signal supplied to subject 17. Voltmeter 25 is also used to indicate the potential of a battery power supply 27 (i.e., self-contained nickel cadmium cells) which power the instrument. A voltmeter switch 29 selects which potential is to be measured. A conventional battery charger circuit 31 is preferably, but not necessarily, provided for recharging the cells from line-voltage household current.

The power level monitoring means also comprises, in a second aspect, a provision for automatically limiting the power of the signal supplied from amplifier 13 by electrodes 19 to subject 17. This power monitoring means includes power sensing circuitry 33 which compares the voltage and current of the signal supplied by the electrodes and includes also means for generating a control signal which varies as a function of the compared voltage and current and which is thereby a function of the power level of the signal supplied by electrodes 19. Limiting circuitry 35 is responsive to the control signal for controlling the power level delivered by amplifier 13 via electrodes 19 to subject 17. A manually variable impedance means 37 is provided for preselecting a maximum power level to which the signal supplied by electrodes 19 is to be limited.

Referring now to FIG. 2, all of the elements to the right of the dashed line indicated at 41 are advantageously mounted on a single, small printed circuit board, there being suitable interconnections between this circuit board and remaining components of the circuitry which are not located on the circuit board, such as the battery cells, level controls 15 and 37, as well as voltmeter 25 and ammeter 23. All components including the cells are suitably mounted within a small conventional instrument housing of a few inches in each dimension having a face panel on which are located the off-on and level controls, meters 23 and 25, and suitable terminals for connection of electrodes 19.

Battery power supply 27 is seen to constitute two sets of cells 27a and 27b, the junction of which constitutes the circuit ground and the other ends of which provide power supply potentials +V and -V, respectively, at levels suitable for semiconductor circuitry to the vari-

ous portions of the circuit on the printed circuit board via leads L1, L2 and LN.

The oscillator circuitry 11 includes a differential operational amplifier A1 of a monolithic integrated circuit variety, such as the commercially available Type 741. Similar operational amplifiers used in the device are designated A2-A4. A so-called bridge-T feedback circuit 43 is interconnected between the output and inverting input of amplifier A1, including capacitors C1 and C2 and resistors R1-R3. The values of these elements are chosen such that the output of amplifier A1 delivers a sinusoidal frequency of substantially precisely 700 Hz, resistance R3 being constituted by a trimming potentiometer for providing fine adjustment of the frequency to this value.

A second feedback circuit 44 for amplifier A1 causes the output signal of amplifier A1 to be maintained substantially at the constant magnitude. For this purpose, a diode D1 rectifies the sinusoidal output of amplifier A1 and stores the rectified voltage across a capacitor C4. A potentiometer R5 connected across capacitor C4 provides at its wiper a voltage for biasing the gate of a field effect transistor (FET) Q1, whose source and drain terminals are connected in a circuit including a resistor R6 between the output and noninverting input of amplifier A1. The noninverting input of amplifier A1 is also connected through a resistor R7 to the circuit ground. FET Q1 operates in effect as a voltage-control resistor to control the gain of amplifier A1 and thereby to maintain its output voltage substantially at a constant value.

The values of the elements of oscillator circuit 11 are carefully chosen so that the output frequency will not vary more than about ± 5 Hz from the desired 700 Hz operating frequency over an operating temperature of, for example, 32°-140° F. Thus the 700 Hz frequency is supplied within these tolerances each time the device is energized, without requiring any recalibration or adjustment.

The output from oscillator 11 is supplied through a capacitor C5 and through the output power adjustment control means 15 to power amplifier 13. Means 15 is shown to constitute a potentiometer whose wiper provides the output signal from oscillator 11 at a preselected magnitude across a scaling circuit comprising resistors R9 and R10. The scaled-down output signal at the junction of these two resistors is supplied through the source and drain terminals of an FET Q2 to the noninverting input of operational amplifier A2. A resistor R11 is connected between the noninverting input of amplifier A2 and the circuit ground. The power amplification circuit 13 comprises also a pair of power output transistors Q3 and Q4 of complementary symmetry, Q3 being of the PNP type and Q4 being of the NPN type. A capacitor C7 is connected between the output of amplifier A2 and the junction of the collectors of transistors Q3 and Q4, the emitters of these two transistors being supplied with the power supply potentials +V and -V, respectively. A resistor R8 is connected between the output of amplifier A2 and the circuit ground.

Negative feedback for amplifier 13 is constituted by resistors R12 and R' respectively interconnected connected between the noninverting input of amplifier A2, the junction of the collectors of transistors Q3 and Q4, and ground, thereby to provide amplifier 13 with a voltage gain of about 17. Thus, also, it will be seen that

transistors Q3 and Q4 are within the feedback loop. The amplified 700 Hz sinusoidal signal is supplied through a capacitor C11 to the primary winding of impedance-matching transformer 21. The other side of the primary winding of transformer 21 is connected to the circuit ground through a resistor R13 of relatively low resistance, e.g., about 0.56 ohms, for a purpose which shortly will be made clear.

Transformer 21 has a primary-to-secondary winding ratio of 1:3.5. One side of the secondary of transformer 21 is connected to one of the electrodes 19, the other side being connected through the primary of a current transformer 43 to the other of electrodes 19. The tissue resistance of the electroanesthesia subject 17 is represented in the drawing as a resistor R_s connected between electrodes 19, and may be assumed to vary over an exemplary range of approximately 100–800 ohms. It will be understood that the subject resistance may vary according to the type of subject (i.e., type of animal), the location of electrodes 19 and other factors. Electrodes 19 may be of various cutaneous or subcutaneous types, as those with particular expertise in electroanesthetic techniques will be aware, the design and placement of such electrodes 19 being outside the scope of the present description.

If the apparatus includes the battery charger circuit 31 shown in FIG. 1 operating from line voltage, i.e., household utility service, is important to provide isolation between the primary and secondary windings of transformer 21 in order to ensure against a shock hazard to the electroanesthesia subject. For this purpose, transformer 21 is shown as including a Faraday shield 45 for preventing stray a.c. voltages from being coupled from the primary winding to the secondary winding, as during recharging.

Current transformer 43 is of a conventionally constructed ferrite toroidal type, its secondary winding providing an a.c. voltage which is proportional to the magnitude of the current flowing in its primary winding. Connected in a circuit with this secondary winding are a calibration potentiometer R15 and a resistor R16 and a pair of oppositely oriented diodes D2 and D3 which rectify the a.c. voltage appearing across the secondary winding of transformer 43 and charge respective capacitors C12 and C13. Current meter 23 is connected between the junctions of these respective pairs of diodes and capacitors to provide a reading in milliamperes of the current supplied to the subject.

A voltmeter circuit similarly includes a calibration potentiometer R17 and resistor R18, as well as diodes D4 and D5 and capacitors C14 and C15. Voltmeter switch 29 is of a DP3T type and is adapted to switch voltmeter 25 across the junctions of these respective pairs of diodes and capacitors for indicating the potential of the signal supplied to the subject. The positive and negative battery supply potentials +V and -V are supplied to one side of a respective pair of resistors R19 and R20 interconnected with switch 29, the latter being also operable to select which of these two potentials is indicated by voltmeter 25.

Turning now to the power sensing circuitry 33 and limiting circuitry 35, a lead L3 interconnects the top side of the primary winding of transformer 21 with the inverting input of operational amplifier A3 through a resistor R22 to provide a signal at this input of the operational amplifier which is a function of the voltage across the primary winding of transformer 21. Simi-

larly, a lead L4 is connected through a resistor R23 from the top of resistor R13 to this inverting input of amplifier A3 to supply thereto a voltage which is proportional to current through resistor R13, and which is thus also proportional to the current applied to the subject. The noninverting input of amplifier A3 is grounded and another resistor R24 provides a feedback connection between the output and the inverting input of this operational amplifier. Thus the latter is seen to be connected as a summing amplifier which takes, in effect, a weighted sum of the voltage and current of the signal supplied to the primary winding of transformer 21 to approximate the product of this voltage and current and thereby to approximate the amount of power delivered by the device to the subject.

The voltage at the output of amplifier A3 is therefore a control signal which varies as a function of the voltage and current and is thereby a function of the power level of the signal supplied by the electrodes 19. This signal is rectified by a diode D7 and supplied through a resistor R25 to the inverting input of comparator amplifier A4. A resistor R26 and a capacitor C17 are connected between the anode of diode D7 and the circuit ground and remove the a.c. component of the signal. Circuitry for providing a reference voltage to the inverting input of amplifier A4 comprises resistors R27 and R29, the latter a calibration potentiometer. A resistor R28 and a zener diode D8 form a reference voltage supply that excites the internal reference potentiometer R29 and/or the external overload limit control 37 (to the left of the dash line 41). Control 37 is constituted by a potentiometer whose wiper is connected to the junction between resistor R30 and capacitor C18. The position of this wiper accordingly determines the magnitude of the bias voltage applied to the inverting input of amplifier A4. A feedback connection is provided between the output and inverting input of amplifier A4 by resistor R31. Thus, depending on the setting of the limiting control potentiometer 37, the voltage at the output of amplifier A4 becomes increasingly negative with increasing power when the voltage proportional to power exceeds the voltage from limiting control 37.

The gate of FET Q2 is interconnected through a resistor R32 to the output of amplifier A4 and through a clamping diode D9 to the circuit ground. Accordingly, conduction through the source and drain electrodes of FET Q2 is limited when the output of amplifier A4 becomes increasingly negative, the FET acting in effect as a voltage-controlled resistor. Thus, should the power supplied by the apparatus to the electroanesthesia subject increase beyond a preselected maximum level determined by setting of control 37, an increasingly negative voltage is supplied by operational amplifier A4 to the gate of FET Q2 to reduce the magnitude of the signal supplied to the noninverting input of amplifier A2, and thereby to limit automatically to the preselected maximum level the power supplied by electrodes 19.

Operation of the present device can be better understood by referring to FIG. 3 in which the increasing effect of relaxation and analgesia of an electroanesthesia subject are plotted as functions of increasing frequency of a sinusoidal electroanesthesia current for different magnitudes of current through tissue of the subject. This graph indicates that for current of a first magnitude I_1 , relaxation increases in magnitude with increasing frequency, as shown by the solid-line curve 47,

while the degree of analgesia (insensitivity to pain) decreases with increasing frequency, as shown by the dashed-line curve 49. A frequency f_1 may be regarded as an optimum frequency with regard to both relaxation and analgesia, in that both relaxation and analgesia are relatively high at this frequency. If the current be increased to a still greater magnitude I_2 , relaxation also increases with increasing frequency, as indicated by the solid-line curve 51, while analgesia decreases with increasing frequency, as shown by the steeply sloped dashed-line curve 53. Again, there is an optimum frequency f_2 at which both relaxation and analgesia are optimized. Greater current through the subject requires greater power and thereby requires greater battery capacity in a portable device of the character described. Thus, it will be seen that certain frequencies, i.e., 700 Hz, 1,400 Hz, and so forth, have particularly good results, permitting electroanesthesia to be achieved with the least amount of power. Since the power required to effect anesthesia is minimized, small rechargeable cells are able to power the present device for several hours of continuous operation.

If the current supplied to the subject were to become excessive, there could be danger to the subject. However, the power limiting circuitry of the invention automatically limits the maximum amount of power supplied to the subject, preventing injury. As an example, the normal operating power can be about 150 mW, and the power under overload conditions can be limited to no greater than about 300 mW, for example. Of course, by changing the setting of limiting control 37, the power may be limited to some value less than 300 mW. This limiting feature also protects the instrument in the event that electrodes 19 are shorted together, the power being automatically limited to 300 mW or less as determined by control 37 to prevent damage to the instrument. If desired, an indicating device such as a light emitting diode may be employed for signalling an overload condition of this type.

The following example illustrates the invention.

EXAMPLE

Electroanesthesia was induced in seven mongrel dogs weighing 6–7.5 kg. using a prototype model of the present invention capable of supplying the sinusoidal signal of 700 Hz at a potential of 12 v. and a current of 12 ma. Cutaneous circular electrodes of 1.8 cm. in diameter were placed bitemporally on the dogs and anesthesia was induced with the device. The magnitudes of the current and voltage required for loss of toe-pad reflect were recorded, and the corresponding impedance and wattage were then calculated from these figures for each animal. The length of time required for induction in each case, as well as indications of smoothness or discomfort, were recorded. It was found that, on the average, 1–1.5 minutes were required for each animal to reach a state of anesthesia using the device. The following table indicates the amount of power required for each of the seven subjects:

TABLE

Dog	Corrected Power (mW)
1	44.0
2	43.2
3	27.3
4	14.2
5	54.2
6	13.1
7	13.9

The average power required for these seven subjects was 27.99 mW. A prior art device was also used to achieve anesthesia for these seven subjects and was found to require 4–8 minutes on the average to achieve the same state of anesthesia as the device of this invention, and required a substantially higher average power.

In view of the above, it will be seen that the several objects of the invention are achieved and other advantageous results attained.

As various changes could be made in the above constructions without departing from the scope of the invention, it is intended that all matter contained in the above description or shown in the accompanying drawings shall be interpreted as illustrative and not in a limiting sense.

What is claimed is:

1. A portable, battery-powered and self-contained electroanesthesia device comprising:

means for generating a substantially constant magnitude signal of substantially sinusoidal waveform at a fixed, preset frequency with relatively high accuracy and without requiring recalibration each time said device is energized, said preset frequency characteristically producing electroanesthetic induction in a live subject with relatively high efficiency;

means for amplifying said signal to a power level suitable for producing electroanesthetic induction of the subject;

means for supplying the amplified signal to tissue of the subject to cause electroanesthetic induction of the subject;

means for matching the impedance of said tissue of the subject to said amplifying means for maximizing transfer of the amplified signal to said tissue;

means for monitoring the power level of the signal supplied to said tissue; and

means for automatically limiting the power of the signal supplied to said tissue to a maximum level, said power limiting means being adapted to sense the voltage applied to and the current drawn by said live subject.

2. A portable, battery-powered electroanesthesia device as set forth in claim 1, said preset frequency being substantially equal to a harmonic of 700 Hz.

3. A portable, battery-powered electroanesthesia device as set forth in claim 1 wherein the power limiting means includes means for summing voltages which vary as functions of the voltage and the current of the signal supplied to said tissue, means for generating a control signal which varies as a function of the summed voltages and which is thereby a function of the power level of the signal supplied to said tissue, and means responsive to said control signal for controlling the power level of the signal supplied to said tissue.

4. A portable, battery-powered electroanesthesia device as set forth in claim 3, said power limiting means further comprising manually variable means for preselecting a maximum level to which the power of the signal supplied to said tissue is limited.

5. A portable, battery-powered electroanesthesia device as set forth in claim 3 wherein said means responsive to said control signal comprises a field effect transistor.

6. A portable, battery-powered electroanesthesia device as set forth in claim 4 further comprising manually

variable means for controlling the power level normally supplied to said tissue, and manually variable means for adjusting the maximum power level.

7. A portable, battery-powered electroanesthesia device as set forth in claim 1 wherein the signal generating means comprises an oscillator including a resistance-capacitance feedback circuit causing oscillation

at said preset frequency to supply said sinusoidal signal and a second feedback circuit comprising a resistance-capacitance controlled field effect transistor for causing said signal to be maintained substantially at said constant magnitude.

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