APPARATUS FOR SWITCHING NOMINAL AND ATTENUATED POWER BETWEEN ABLATION PROBES

A tissue ablation system comprises an ablation source, such as an RF ablation source, configured for generating a common power signal, and a power multiplexer configured for splitting the power signal into first and second power signals, substantially attenuating the second power signal relative to the first power signal to create nominal and attenuated power signals, and sequentially delivering the nominal power signal to each tissue ablation probe, while delivering the attenuated power signal to the remaining ablation probes to which the nominal power signal is currently not delivered.
APPARATUS FOR SWITCHING NOMINAL AND ATTENUATED POWER BETWEEN ABLATION PROBES

FIELD OF THE INVENTION

The invention relates generally to the structure and use of radiofrequency electrosurgical apparatus for the treatment of tissue. More particularly, the invention relates to an electrosurgical system having multiple ablation probes to treat large volumes of tissue, particularly for the treatment of tumors in the liver and other tissues and organs.

BACKGROUND

The delivery of radio frequency (RF) energy to target regions within tissue is known for a variety of purposes of particular interest to the invention. In one particular application, RF energy may be delivered to diseased regions (e.g., tumors) for the purpose of ablating predictable volumes of tissue with minimal patient trauma. In particular, RF energy translates into ion agitation, which is converted into heat and induces cellular death via coagulation necrosis. Multiple needle electrodes have been designed for the treatment and necrosis of tumors in the liver and other solid tissues. PCT application WO 96/29946 and U.S. Patent No. 6,379,353 disclose such probes. In U.S. Patent No. 6,379,353, a probe system comprises a cannula having a needle electrode array reciprocatably mounted therein. The individual electrodes within the array have spring memory, so that they assume a radially outward, arcuate configuration as they are advanced distally from the cannula. In general, a multiple electrode array creates a larger lesion than that created by a single needle electrode.
In theory, RF ablation can be used to sculpt precisely the volume of necrosis to match the extent of the tumor. By varying the power output and the type of electrical waveform, it is possible to control the extent of heating, and thus, the resulting ablation. However, the size of tissue coagulation created from a single electrode, and to a lesser extent a multiple electrode array, has been limited by heat dispersion. As a consequence, when ablating lesions that are larger than the capability of the above-mentioned devices, the common practice is to stack ablations (i.e., perform multiple ablations) within a given area. This requires multiple electrode placements and ablations facilitated by the use of ultrasound imaging to visualize the electrode in relation to the target tissue. Because of the echogenic cloud created by the ablated tissue, however, this process often becomes difficult to accurately perform. This process considerably increases treatment duration and patient discomfort and requires significant skill for meticulous precision of probe placement.

Increasing generator output, however, has been generally unsuccessful for increasing lesion diameter, because an increased wattage is associated with a local increase of temperature to more than 100°C, which induces tissue vaporization and charring. This, in turn, increases local tissue impedance, limiting RF deposition, and therefore heat diffusion and associated coagulation necrosis. In addition, patient tolerance appears to be at the maximum using currently available 200W generators.

To a large extent, the size and nature of an ablation lesion depends on how the electrode element(s) are arranged. In one arrangement, RF current
may be delivered to an electrode element (whether a single electrode or electrode array) in a monopolar fashion, which means that current will pass from the electrode element to a dispersive electrode attached externally to the patient, e.g., using a contact pad placed on the patient's flank. Alternatively, the RF current may be delivered in a bipolar fashion, which means that current will pass between "positive" and "negative" electrode elements. Bipolar arrangements, which require the RF energy to traverse through a relatively small amount of tissue between the tightly spaced electrodes, are more efficient than monopolar arrangements, which require the RF energy to traverse through the thickness of the patient's body. As a result, bipolar electrode arrays generally create larger and/or more efficient lesions than monopolar electrode arrays. To provide even larger lesions, it is known to operate two electrode arrays in a bipolar arrangement.

Thus, to a certain extent, the use of bipolar electrode arrangements has eliminated the need to "stack" ablations when treating a tumor. The ability to provide uniform heating and the creation of homogenous tissue lesions, however, is particularly difficult with bipolar devices. For example, the two bipolar electrodes may be placed in regions with quite different perfusion characteristics, and the heating around each pole can be quite different. That is, one pole may be located adjacent to a large blood vessel, while the other pole may be located adjacent to tissue, which is less perfused. Thus, the pole located in the less perfused tissue will heat the tissue immediately surrounding the electrode much more rapidly than the tissue surrounding the opposite polar electrode is heated. In such circumstances, the tissue
surrounding one pole may be preferentially heated and necrosed, while the tissue surrounding the other pole will neither be heated nor necrosed sufficiently. In the case where two electrode arrays are used, if the distance between the electrode arrays is too great in an attempt to ablate a longer tissue volume, the energy transmitted between the electrode arrays may thin and not fully ablate the intermediate tissue. As a result, an hour-glass shaped ablation, rather than the desired uniform circular/elliptical ablation, may be created. Because the electrode arrays are, in effect, three-dimensional, portions between the electrode arrays will be closer together than other portions of the electrode arrays, thereby causing a non-uniform current density between the electrode arrays, resulting in a non-uniform ablation. Besides lacking the ability to produce predictable homogenous lesions, bipolar arrangements, which are designed to ablate tissue between the electrodes, are not well-suited for simultaneously ablating multiple tissue regions.

In situations where it is desired to produce large homogenous lesions or simultaneously ablate multiple tissue regions, it is known to arrange multiple probes in a monopolar fashion (i.e., the RF energy generated by each probe is conveyed to a dispersive electrode attached to the skin of the patient. In this case, current flows from each probe to the ground pad. A drawback to this approach is that simultaneously supplying power to multiple probes taxes the power output by the RF generator, which may cause insufficient heating around the probes. Also, because the tissue adjacent the probes is non-uniform (e.g., one probe may be adjacent a blood vessel), the heating pattern
created by the probes will be non-uniform, thereby making it difficult to predict the nature of the resulting lesion.

To address these drawbacks, it is known to use an ablation system that sequentially switches ablation energy between probes, so that at any given time, ablation energy is supplied to only one probe. While this switching technique may result in a more efficient and predictable lesion, it is believed that, during any given time period, the tissue adjacent the probes to which the ablation energy is not currently supplied temporarily cools—especially when the switching speed between the probes is relatively slow, e.g., a few seconds. As a result, the cooled tissue must be reheated when power is again supplied to the adjacent probes, thereby losing some efficiency in the ablation process.

SUMMARY OF THE INVENTION

In accordance with a first embodiment of the invention, a tissue ablation system is provided. The tissue ablation system comprises an ablation source, such as a radio frequency (RF) source, having a common power output. The tissue ablation system further comprises a power splitter having a splitter input and first and second splitter outputs. The splitter input is coupled to the common power output. In one embodiment, the first and second splitter outputs are substantially unattenuated relative to the common power output. For example, an intervening attenuation device, whatever its nature, can be placed between the common power output and the first and second splitter inputs, so that the first and second splitter outputs are both attenuated relative to the common power output.
The tissue ablation system further comprises a power attenuator having an attenuator input and an attenuator output. The attenuator input is coupled to the second splitter output. In one embodiment, the attenuator is configured for attenuating power at a level equal to or greater than 3 dB, e.g., within the range of 3 dB to 6 dB. Although the attenuator may be fixed in one embodiment, the attenuator may be variable in alternative embodiments. For example, the attenuation value of the attenuator may be capable of being set by the user. As another example, the tissue ablation system may further comprise a feedback control circuit configured for receiving a feedback input (e.g., a measured physiological parameter, such as tissue impedance or temperature) and for varying the attenuation value of the attenuator based on the feedback input.

The tissue ablation system further comprises a switch having first and second switch inputs and a plurality of switch outputs. The first switch input is coupled to the first splitter output and the second switch input is coupled to the attenuator output. The plurality of switch outputs are coupleable or coupled to a plurality of tissue ablation probes, which may be included within the tissue ablation system. In one embodiment, the plurality of switch outputs comprises at least three outputs to accommodate at least three ablation probes, although other plural numbers of switch outputs can be used including just two.

The switch is configured for sequentially coupling the first switch input (which is coupled to the first splitter output) to each one of the switch outputs, while coupling the second switch input (which is coupled to the attenuator
output) to the switch output currently not coupled to the first switch input. This configuration allows the power to be continuously delivered to each ablation probe—albeit sometimes at an attenuated level.

The switch may be configured for sequentially coupling the first switch input to each switch output at a fixed rate. Preferably, this fixed switching rate is greater than once per second to maximize the ablation efficiency of the system, but less than thirty seconds to prevent tissue charring. As such, the fixed switching rate may be less than once per one second or greater than once per thirty seconds if desirable. The switch may alternatively be configured for sequentially coupling the first switch input to each switch output at a variable rate. For example, the tissue ablation system may further comprise a feedback control circuit configured for receiving a feedback input (e.g., a measured physiological parameter, such as tissue impedance or temperature) and controlling the switch to couple the first splitter output to each switch output based on the feedback input.

Notably, for the purposes of this specification, two elements that are coupled or coupleable together does not necessarily mean that the two elements must be connected together. Rather, the one element need only be capable of receiving power or derivation of that power from that other element. Thus, the two elements may be coupled or coupleable together even though an intervening element exists between the two elements.

In accordance with a second embodiment of the invention, another tissue ablation system is provided. The tissue ablation system comprises an
ablation source, such as an RF ablation source, configured for generating a common power signal. The tissue ablation system further comprises a power multiplexor configured for splitting the power signal into first and second power signals, and substantially attenuating the second power signal relative to the first power signal. In one embodiment, the first power signal is substantially unattenuated relative to the common power signal. For example, an intervening attenuation device can be used to attenuate the first and second power signals after they are output by the ablation source.

In one embodiment, the power multiplexor is configured for attenuating power at a level equal to or greater than 3 dB, e.g., within the range of 3 dB to 6 dB. In an optional embodiment, the power multiplexor is configured for varying the value that the second power signal is attenuated. In this case, the power multiplexor may optionally be configured for receiving a feedback input (e.g., a measured physiological parameter, such as tissue impedance or temperature) and varying the attenuation value based on the feedback input.

The power multiplexor is further configured for delivering the first power signal to one or more of a plurality of tissue ablation probes (which may be included within the tissue ablation system), while delivering the second power signal to a different one or more of the plurality of ablation probes. The power multiplexor may be configured for varying the number of tissue ablation probes to which it delivers the first and second power signals. In this manner, the tissue ablation system can be adapted to any number of tissue ablation probes desired to be used.
By way of non-limiting example, the power multiplexor may be configured for sequentially delivering the first power signal to different sets of the tissue ablation probes, while delivering the second power signal to the tissue ablation probes to which the first power signal is not currently delivered. The set of the tissue ablation probes to which the first power signal is sequentially delivered can equal any value, but in the embodiment, the probe set only comprises a single tissue ablation probe in order to focus more of the ablation energy towards a single tissue ablation probe at a time. The plurality of tissue ablation probes may equal any plurality number, including just two ablation probes, but in one embodiment, comprises at least three tissue ablation probes.

If the first power signal is sequentially delivered, the power multiplexor may be configured for sequentially deliver it to each tissue ablation probe at a fixed rate. Preferably, this fixed rate is greater than one probe per second to maximize the ablation efficiency of the system, but less than one probe per thirty seconds to prevent tissue charring. As such, the fixed rate may be less than one probe per one second or greater than one probe per thirty seconds if desirable. The power multiplexor may alternatively be configured for sequentially coupling the first power signal to each tissue ablation probe at a variable rate. For example, the power multiplexor may be configured for receiving a feedback input (e.g., a measured physiological parameter, such as tissue impedance or temperature) and delivering the first power signal to each tissue ablation probe based on the feedback input.
BRIEF DESCRIPTION OF DRAWINGS

The drawings illustrate the design and utility of embodiment(s) of the
invention, in which similar elements are referred to by common reference
numerals, and in which:

Fig. 1 is a perspective view of a tissue ablation system constructed in
accordance with one embodiment of the invention;

Fig. 2 is a partially cutaway side view of an ablation probe used in the
tissue ablation system of Fig. 1, wherein an array of electrode tines is
particularly shown retracted;

Fig. 3 is a partially cutaway side view of an ablation probe used in the
tissue ablation system of Fig. 1, wherein an array of electrode tines is
particularly shown deployed;

Fig. 4 is a timing diagram illustrating the switching of ablation energy
between tissue ablation probes used in the tissue ablation system of Fig. 1;

Fig. 5 is a table illustrating the ablation states in which the plurality of
tissue ablation probes used in the tissue ablation system of Fig. 1 can be
placed;

Fig. 6 is a detailed schematic diagram of a power multiplexor used in
the tissue ablation system of Fig. 1;

Fig. 7 is a perspective view of a tissue ablation system constructed in
accordance with another embodiment of the invention;
Fig. 8 is a timing diagram illustrating the switching of ablation energy between tissue ablation probes used in the tissue ablation system of Fig. 7; Fig. 9 is a table illustrating the ablation states in which the plurality of tissue ablation probes used in the tissue ablation system of Fig. 7 can be placed; and Fig. 10 is a detailed schematic diagram of a power multiplexor used in the tissue ablation system of Fig. 7.

DETAILED DESCRIPTION OF THE ILLUSTRATED EMBODIMENTS

Referring generally to Fig. 1, a tissue ablation system 10 constructed in accordance with one embodiment of the invention will be described. The tissue ablation system 10 generally includes a plurality of tissue ablation probes 12 (in this case, three probes 12(1)-(3)) for introduction into the body of a patient for ablative treatment of target tissue, a radio frequency (RF) generator 14 configured for generating RF power, and a power multiplexor 16 configured for receiving the RF power from the RF generator via a standard RF cable 15 and selectively providing the RF power to the ablation probes 12 via standard RF cables 13 in accordance with a particular pattern, such that each ablation probe 12 is always "on" during a tissue ablation procedure—albeit at a power level that may be substantially below the power level of the RF generator 14.

Referring specifically now to Figs. 2 and 3, each probe 12 includes an elongate cannula 18 having a proximal end 20, a distal end 22, and a central lumen 24, a probe shaft 26 slidably disposed within the cannula lumen 24 and
having a proximal end 28 and a distal end 30, and an array of electrode tines 32 carried by the distal end 28 of the probe shaft 26. The cannula 18 may be rigid, semi-rigid, or flexible depending upon the designed means for introducing the cannula 18 to the target tissue. The probe shaft 26 is composed of a suitably rigid material, such as plastic, metal or the like.

Each probe 12 further includes a handle assembly 34, which includes a handle member 36 mounted to the proximal end 26 of the probe shaft 26, and a handle sleeve 38 mounted to the proximal end 20 of the cannula 18. The handle member 36 is slidably engaged with the handle sleeve 38 (and the cannula 18). The handle member 36 and handle sleeve 38 can be composed of any suitable rigid material, such as, e.g., metal, plastic, or the like. The handle assembly 34 also includes an electrical connector 40 mounted within the handle member 36. The electrical connector 40 is electrically coupled to the electrode array 32, e.g., via the probe shaft 26 (which will be electrically conductive) or separate wires (not shown). The electrical connector 40 is configured for mating with the proximal end of a RF cable 13 (shown in Fig. 1). Alternatively, the RF cable 13 may be hardwired within the handle member 36.

It can be appreciated that longitudinal translation of the probe shaft 26 relative to the cannula 18 in a distal direction 42, by holding the handle sleeve 38 and displacing the handle member 36 in the distal direction 42, deploys the electrode array 32 from the distal end 22 of the cannula 18 (Fig. 3), and longitudinal translation of the probe shaft 26 relative to the cannula 18 in a proximal direction 44, by holding the handle sleeve 38 and displacing the
handle member 36 in the proximal direction 44, retracts the probe shaft 26 and the electrode array 32 into the distal end 22 of the cannula 18 (Fig. 2).

In the illustrated embodiment, the RF current is delivered to the electrode array 32 in a monopolar fashion, which means that current will pass from the electrode array 32, which is configured to concentrate the energy flux in order to have an injurious effect on the surrounding tissue, and a dispersive electrode (not shown), which is located remotely from the electrode array 32 and has a sufficiently large area (typically 130 cm² for an adult), so that the current density is low and non-injurious to surrounding tissue. In the illustrated embodiment, the dispersive electrode may be attached externally to the patient, e.g., using a contact pad placed on the patient's flank.

Further details regarding electrode array-type probe arrangements are disclosed in U.S. Patent No. 6,379,353. It should be noted that the tissue ablation probe 12 illustrated in Figs. 2 and 3 is only one type of ablation probe that can be used with the tissue treatment system 10. For example, a single needle electrode probe may be used as well.

Referring back to Fig. 1, the RF generator 14 may be a conventional general purpose electrosurgical power supply operating at a frequency in the range from 300 kHz to 5 MHz, with a conventional sinusoidal or non-sinusoidal wave form. Such power supplies are available from many commercial suppliers, such as Valleylab, Aspen, Bovie, and Ellman. Most general purpose electrosurgical power supplies, however, are constant current, variable voltage devices and operate at higher voltages and powers.
than would normally be necessary or suitable. Thus, such power supplies will usually be operated initially at the lower ends of their voltage and power capabilities, with voltage then being increased as necessary to maintain current flow. More suitable power supplies will be capable of supplying an ablation current at a relatively low fixed voltage, typically below 200 V (peak-to-peak). Such low voltage operation permits use of a power supply that will significantly and passively reduce output in response to impedance changes in the target tissue. The output will usually be from 50 W to 300 W, usually having a sinusoidal wave form, but other wave forms would also be acceptable. Power supplies capable of operating within these ranges are available from commercial vendors, such as Boston Scientific Therapeutics Corporation. Preferred power supplies are model RF-2000 and RF-3000, available from Boston Scientific Corporation.

Referring still to Fig. 1, the power multiplexor 16 comprises a common input connector 17 coupled to an RF output connector 19 of the RF generator 14 via the RF cable 15, and a plurality of output connectors 21 (in this case, three output connectors 21(1)-(3)) coupled to the electrical connectors 40 of the ablation probes 12 via the RF cables 13. Although the power multiplexor 16 is shown external to the RF generator 14, the power multiplexor 16 can alternatively be incorporated into RF generator 14 itself. However, it is believed that an external power multiplexor 16 will allow the tissue ablation system 10 to be more easily implemented with standard RF generators simply by connecting the power multiplexor 16 between the tissue ablation probes 12 and RF generator 14, as well as to provide physicians the flexibility of
optionally using the RF generator 14 in a standard manner by directly connecting a single tissue ablation probe 12 to the RF generator 14 without the use of the power multiplexor 16.

In any event, the power multiplexor 16 is configured for splitting the power signal from the output connector 19 of the RF generator 14 into two power signals, substantially attenuating one of the power signals relative to the other power signal, and presenting the power signals to the output connectors 21, and thus, the ablation probes 12 in accordance with a particular pattern in a manner that provides at least some power to all of the ablation probes 12 during any given time period.

Although it is preferred that one of the power signals remains substantially unattenuated, both power signals may be attenuated if desired as long as one of the power signals is substantially more attenuated than the other power signal. For example, after attenuation, the second power signal may be 1-30 dB less than the first power signal, but preferably is 3-6 dB less than the first power signal. For purposes of brevity and clarity, the power signal that exhibits substantially less power than the other power signal will be considered the "attenuated signal" and the signal that exhibits substantially more power than the other signal will be considered the "nominal signal."

It should be noted the power signals will sometimes be described herein as being presented to the output connectors 21, and at other times will be described herein as being delivered to ablation probes 12. When a power signal is described as being presented to an output connector 21, it follows
that such power signal will be delivered to an ablation probe 12 connected to
the output connector 21. In a similar fashion, when a power signal is
described as being delivered to an ablation probe 12, it follows that such
power signal is presented to the output connector 21 to which the ablation probe 12 is connected.

The particular pattern followed by the power multiplexor 16 in delivering
the power signals to the ablation probes 12 involves sequentially delivering
the nominal power signal to different sets of the ablation probes 12, while
delivering the attenuated power signal to other sets of the ablation probes 12.

In the illustrated embodiment, each set of ablation probes 12 to which the
nominal power signal is delivered contains a single ablation probe 12, and thus, each set of ablation probes 12 to which the attenuated power signal is
delivered contains the remaining two ablation probes 12.

For example, as illustrated in Fig. 4, the power multiplexor 16 delivers
the nominal power signal to the first ablation probe 12(1) during a first time
period T1, while delivering the attenuated power signal to the second and
third ablation probes 12(2), (3). The power multiplexor 16 then delivers the
nominal power signal to the second ablation probe 12(2) during a second time
period T2, while delivering the attenuating power signal to the first and third
ablation probes 12(1), (3). The power multiplexor 16 then delivers the
nominal power signal to the third ablation probe 12(3) during a third time
period T3, while delivering the attenuating power signal to the first and second
ablation probes 12(1), (2). This pattern is then repeated for subsequent time
periods until completion of the ablation procedure. Thus, it can be
appreciated that the plurality of ablation probes 12 may be placed within three
different ablation states, as illustrated in Fig. 5. The timing of the pattern may
be controlled in any one or more of a variety of manners, as will be described
in further detail below.

Although the set of ablation probes 12 to which the nominal power
signal is delivered is described as containing a single ablation probe, and the
set of ablation probes 12 to which the attenuated power signal is delivered is
described as containing the remaining two ablation probes, it should be noted
that other probe set configurations are possible. For example, the nominal
power signal may be sequentially delivered to different pairs of ablation
probes 12, while the attenuated power signal is delivered to the remaining
ablation probe 12. It should be noted, however, that it is preferred that the
nominal power signal be delivered to a single ablation probe at a time in order
to maximize the efficiency of the tissue ablation system 10.

The power multiplexor 16 may be implemented in any one of a variety
of manners in order to effect the power signal attenuation and switching
functions described above. In the embodiment illustrated in Fig. 6, the power
multiplexor 16 comprises a power splitter 46 having a common input terminal
48 coupled to the common input connector 17 of the power multiplexor 16.
The power splitter 46 is configured for splitting the power from the RF
generator 14, so that it is presented at first and second output terminals 50,
52.
The power multiplexor 16 further comprises a power attenuator 54 having an input terminal 56 coupled to one of the first and second output terminals 50, 52 of the power splitter 46, and an output terminal 58 for presentation of the attenuated power signal. In the illustrated embodiment, the input terminal 56 of the power attenuator 54 is coupled to the second output terminal 52 of the power splitter 46—although the input terminal 56 of the power attenuator 54 can alternatively be coupled to the first output terminal 50 of the power splitter 46 instead. The value of the power attenuator 54 may fall within the range of 1-30 dB, preferably within the range of 3-6 dB. In the illustrated embodiment, the attenuator 54 has a fixed attenuation value, although the attenuation value may alternatively be variable, as discussed in further detail below. Thus, it can be appreciated that the nominal power signal is presented at the first output terminal 50 of the power splitter 46, while the attenuated power signal is presented at the output terminal 58 of the power attenuator 54.

The power multiplexor 16 further comprises a switch device 59 comprising a plurality of switch elements 60 (in this case, three switch elements 60(1)-(3)). Each switch element 60 takes the form of a single-pole double-throw (SPDT) switch element that comprises a first input terminal 62 coupled to the first output terminal 50 of the power splitter 46 for receiving the nominal power signal, and a second input terminal 64 coupled to the output terminal 58 of the attenuator 54 for receiving the attenuated power signal. Each switch element 60 further comprises an output terminal 66 coupled to the respective one of the previously described multiplexed output connectors.
21 of the power multiplexor 16 for presentation of either the nominal power signal or the attenuated power signal thereon.

That is, the output terminal 66 of the first switch element 60(1) is coupled to the first output connector 21(1), the output terminal 66 of the second switch element 60(2) is coupled to the second output connector 21(2), and the output terminal 66 of the third switch element 60(3) is coupled to the third output connector 21(3). Each switch element 60 comprises a control terminal 68 for changing the respective switch element 60 between two states, the first of which couples the first input terminal 62 to the output terminal 66 to pass the nominal power signal through the switch element 60 to the respective output connector 21 of the power multiplexor 16, the second of which couples the second input terminal 64 to the output terminal 66 to pass the attenuated power signal through the respective switch element 60 to the respective output connector 21 of the power multiplexor 16.

The power multiplexor 16 further comprises a control circuit 70 coupled to the control terminals 68 of the respective switch elements 60 for controlling the timing of the switch elements 60. As previously stated, at any given time, the nominal power signal will be delivered to only one of the ablation probes 12, while the attenuating power signal will be delivered to the remaining ablation probes 12 in accordance with the switching pattern illustrated in Fig. 4. To this end, the control circuit 70 sequentially configures the switch device 59 between three different states by configuring each switch element 60 to pass the nominal power signal to the respective output connector 21, while
configuring the remaining switch elements 60 to pass the attenuated power signal to the remaining respective output connectors 21.

That is, the control circuit 70 configures the switch device 59 in a first state by configuring the first switch element 60(1) to pass the nominal power signal to the first output connector 21(1), while configuring the second and third switch elements 60(2), (3) to pass the attenuated power signal to the second and third output connectors 21(2), (3). The control circuit 70 configures the switch device 59 in a second state by configuring the second switch element 60(2) to pass the nominal power signal to the second output connector 21(2), while configuring the first and third switch elements 60(1), (3) to pass the attenuated power signal to the first and third output connectors 21(1), (3). The control circuit 70 configures the switch device 59 in a third state by configuring the third switch element 60(3) to pass the nominal power signal to the third output connector 21(3), while configuring the first and second switch elements 60(1), (2) to pass the attenuated power signal to the first and second output connectors 21(1), (2).

It should be noted that sequential configuration of the switch elements 60 to pass the nominal power signal to the respective output connector 21, and thus, the respective ablation probe 12, does not necessarily mean that the switch elements 60 are so configured in a numerical order, i.e., 1, 2, 3, etc. For instance, the first switch element 60 can be configured to pass the nominal power signal to the first output connector 21, then the third switch element 60, and then the second switch element 60.
The control circuit 70 may be configured to control the timing of the switch device 59 in any one of a variety of manners. In the illustrated embodiment, the control circuit 70 is configured for switching the state of the switch device 59 at a fixed frequency. Preferably, the fixed frequency is at least once every second to maximize the efficiency of the tissue ablation system 10, but less than once every thirty seconds to prevent tissue charring otherwise resulting from providing nominal power to an ablation probe for too long. The control circuit 70 may base the switch timing on the operating frequency of the power signal. For example, if the operating frequency is 300 KHz, and it is desired to change the state of the switch device 59 once a second, the control circuit 70 will change the state of the switch device 59 once every 300K cycles. Of course, rather than basing the switching frequency on digital references, such as clock cycles, the switching frequency may be based on analog references, such as capacitor discharge times.

Alternatively, the control circuit 70 may be configured to change the state of the switch device 59 at a variable frequency. For example, the control circuit 70 may change the state of the switch device 59 based on physiological parameters, such as temperature or impedance, that can be sensed and delivered back to the control circuit 70 during the ablation process (shown as dashed lines in Fig. 6). In the case of an impedance measurement, the control circuit can be connected to an electrode (not shown) on each ablation probe 12 and the dispersive electrode to determine the impedance of the intervening tissue. In this case of a temperature measurement, a temperature sensor, such as a thermistor or thermocouple
(not shown) can be mounted to an electrode of each ablation probe 12 and then coupled to the control circuit 70.

In contrast to changing the state of the switch device 59 at a fixed frequency, which may not optimal if the state of the switch device 59 is changed to late or too early, the use of sensed physiological feedback to the control circuit 70 allows the state of the switch device 59 to be changed at the exact time that it needs to be changed. For example, as the measured tissue impedance adjacent an ablation probe 12 that is currently delivered with the nominal power signal begins to exponentially increase or the measured temperature of the tissue adjacent the ablation probe 12 approaches 100°C, thereby indicating that full tissue ablation has been achieved, the control circuit 70 may change the state of the switch device 59, so that the nominal power signal is delivered to the next ablation probe 12, and attenuated power signal is delivered to the previous ablation probe 12, along with the other remaining ablation probe 12. The control circuit 70 can then change the state of the switch device 59 based on the measured impedance and/or temperature adjacent the next ablation probe 12 in the same manner, and so on.

It should be noted that the use of physiological feedback parameters can be used with other probe switching implantations besides those that ensure that power is continuously delivered to each ablation probe. For example, such a feature can be applied to prior art probe switching implementation, wherein power is delivered to one probe at a time, while no power is delivered to the remaining probes.
Although the attenuation value of the attenuator 54 has been described as being fixed, the power multiplexor 16 may be configured to vary the amplitude of the attenuated power signal delivered to the ablation probes 12. In this case, the attenuator 54 has a variable attenuation value that is controlled by the control circuit 70 based on a user input. For example, the power multiplexor 16 may have an attenuation control device (not shown) that can be manipulated by the user based on the characteristics of the tissue to be ablated, e.g., by inputting the attenuation value or by inputting the type of tissue to be ablated. For example, if the tissue is lung tissue, which requires a relatively small amount of power to ablate, the user may input a relatively large attenuation value or simply input the tissue type into the attenuation control device. In contrast, if the tissue is liver tissue or otherwise tissue that is highly vascular, the user may input a relatively small attenuation value or simply input the tissue type into the attenuation control device. Whether the attenuation value is increased or decreased, the control circuit 70 will then respond to whichever type of input is used by varying the attenuation value of the attenuator 54.

Rather than basing the control of the attenuation value on a manual input from the user, the attenuation value may be varied based on a measured physiological parameter, such as tissue impedance or temperature, in order to suit a longer probe dwell time, potentially resulting in a larger ablation. For example, if the measured tissue impedance adjacent an ablation probe 12 to which the attenuated power signal is currently delivered is relatively low or the measured temperature of such tissue is much less than
100°C, tissue ablation will not likely occur in a relatively short time, and therefore, the control circuit 70 may automatically decrease the attenuation value of the attenuator 54, so that the amplitude of the attenuated power signal is increased in order to quickly reach the point at which tissue ablation occurs, thereby decreasing the elapsed time of the ablation procedure. In contrast, as the measured tissue impedance adjacent an ablation probe 12 to which the attenuated power signal is currently delivered begins to exponentially increase or the measured temperature of the tissue adjacent the ablation probe 12 approaches 100°C, the control circuit 70 may automatically increase the attenuation value of the attenuator 54, so that the amplitude of the attenuated power signal is decreased, thereby allowing the ablation probes 12 to have a longer dwell time, potentially creating a larger ablation.

The power multiplexor 16 can be implemented in any suitable manner that facilitates the afore-described attenuation and switching functions. In the illustrated embodiment, the attenuator 54 may be a discrete component that can be obtained from supplier, such as from JFW Industries, Inc. The switch 59 and control circuit 70 can be implemented as a low power switching circuit to minimize the cost of the power multiplexor 16. For example, the switch elements 60 take the form of powered transistors, and the control circuit can take the form of logic circuitry. Alternatively, the switch elements 60 can take the form of discrete components or electromechanical devices, such as relays.

Although the tissue treatment system 10 has been described as comprising an equal number of ablation probes 12 and output connectors 21,
a tissue treatment system can optionally be configured, such that the number of ablation probes used can be less than the number of output ports available. For example, Fig. 7 illustrates a tissue treatment system 110 that is similar to the previously described tissue treatment system 10, with the exception that it comprises a power multiplexor 116 with six output connectors 21(1)-(6), although only three ablation probes 12 are still used in this scenario.

The particular pattern followed by the power multiplexor 116 in delivering the power signals to the ablation probes 12 will depend on the number of used output connectors 21; i.e., the number of output connectors 21 to which ablation probes 12 are connected. The power multiplexor 116 will activate only those output connectors 21 that are used (in this case, output connectors 21(1)-(3)) and sequentially change as many ablation states of the plurality of ablation probes 12 as there are activated output connectors 21. The power multiplexor 116 will only sequentially change three ablation states of the ablation probes 12, which will be accomplished in the same manner illustrated in Fig. 5, and will deliver the nominal and attenuated power signals to the ablation probes 12 in the same manner illustrated in Fig. 4.

If additional ablation probes 12 are used, thereby using additional output connectors 21, the power multiplexor 166 will add additional ablation states between which the ablation probes 12 will change. For example, if three additional ablation probes 12 are added, thereby using all six output connectors 21, the power multiplexor 166 will sequentially change six ablation states of the ablation probes 12 in accordance with the switching pattern illustrated in Fig. 9, and will deliver nominal and attenuated power signals to
the six ablation probes 12 connected to the used output connectors 21 in a
manner similar to that previously described, with the exception that nominal
power signal will be delivered to each of the six ablation probes 12, while
delivering the attenuated power signal to the remaining five ablation probes.

That is, as illustrated in Fig. 8, the power multiplexor 116 delivers the
nominal power signal to the first ablation probe 12(1) during a first time period
T1, while delivering the attenuated power signal to the second to sixth
ablation probes 12(2)-(6). The power multiplexor 116 then delivers the
nominal power signal to the second ablation probe 12(2) during a second time
period T2, while delivering the attenuating power signal to the first and third
through fifth ablation probes 12(1), (3)-(6). This switching pattern continues
for time periods T3-T6 and continuously repeats until the ablation procedure is
completed.

The power multiplexor 116 may be implemented in the same manner
as the power multiplexor 16, with the exception that the power multiplexor 116
comprises a switch 159 having six single pole three position (SP3P) switching
elements 160. In particular, like the previously described switching elements
60, each of the switching elements 160 illustrated in Fig. 10 comprises a first
input terminal 162 coupled to the first output terminal 50 of the power splitter
46 for receiving the nominal power signal, and a second input terminal 164
coupled to the output terminal 58 of the power attenuator 54 for receiving the
attenuated power signal. Unlike the previously described switching elements
60, each of the switching elements 160 comprises a third input terminal 169
coupled to ground.
Each switch element 160 also comprises a control terminal 168 for changing the switch element 160 between three states, the first two of which are similar to the states in which the previously described switch elements 60 can be placed. That is, each switch element 160 can be placed in a first state that connects the first input terminal 162 to the output terminal 166 to pass the nominal power signal through the switch element 160 to the respective output connector 21, and a second state that connects the second input terminal 164 to the output terminal 166 to pass the attenuated power signal through the switch element 160 to the respective output connector 21. Unlike the previously described switch elements 60, each switch element 160 can be placed in a third state that connects the grounded third input terminal 169 to the output terminal 166 to prevent the passing of any power signal through the respective switch element 160 to the respective output connector 21. As a result, the switch element 160 and the respective output connector 21 are deactivated.

The power multiplexor 16 also comprises a control circuit 170 that is similar to the previously described control circuit 70, with the exception that the control circuit 170 is configured to activate the used output connectors 21 by passing one of the nominal or attenuated power signals through the respective switch elements 160, and to deactivate the unused output connectors 21 by grounding the respective switch elements 160. As discussed above, the control circuit 170 will sequentially change as many states of the switch device 159 as there are used output connectors 21.
In this case, the control circuit 170 will sequentially configure the switch device 159 between three different states by configuring each of the first three switch elements 160(1)-(3) to pass the nominal power signal to the respective one of the output connectors 21(1)-(3), while configuring the remaining switch elements 160(1)-(3) to pass the attenuated power signal to the remaining respective output connectors 21(1)-(3). The control circuit 170 will effectively accomplish this function in the same manner that the control circuit 70 sequentially configured the switch device 59 described above. The main difference is that the control circuit 170 will ground the switch elements 160(4)-(6) to prevent the passage of either the nominal power signal or the attenuated power signal to the respective output terminals 21(4)-(6). If additional ablation probes 12 are added, thereby using additional output connectors 21, the control circuit 170 will configure the switch 159 with additional states, allowing either of the nominal power signal or the attenuated power signal to pass through the switch elements 160 to the additional output connectors 21.

The control circuit 170 can determine the number of used output connectors 21 in any one of a variety of manners. For example, the power multiplexor 16 can be provided with a user control (not shown) that allows the user to manually set the number of ablation probes 12, and thus, output connectors 21, to be used. In this case, the user will have to mate the ablation probes 12 to the output connectors 21 in a predetermined pattern. For example, if n number of ablation probes 12 are to be used, the user will have to mate the ablation probes with the first n output connectors 21 of the
power multiplexor 16. The control circuit 70 is configured to respond to the
user entry of number of ablation probes 12 by activating the first n output
connectors 21.

Alternatively, the control circuit 170 can be configured to automatically
sense the output connectors 21 that are to be used when the ablation probes
12 are mated with these output connectors 21. For example, mating of an
ablation probe 12 to an output connector 21 may initiate a closed circuit
condition that can be sensed by the control circuit 170. Accordingly, the
control circuit 170 can then be configured to activate the output connector 21
by grounding it through the switch element 160 when the closed circuit
condition is sensed. In contrast, removal of an ablation probe 12 from an
output connector 21 may initiate an open circuit condition that can be sensed
by the control circuit 170. Accordingly, the control circuit 170 can then be
configured to deactivate the output connector 21 by grounding it through the
switch element 160 when the open circuit condition is sensed. Notably, in this
case, the user need not mate the ablation probes 12 to the output connectors
21 in a predetermined pattern, since the control circuit 70 can automatically
sense the output connectors 21 that are to be used.

It should be noted that the adaptable probe activation feature can be
used with other probe switching implantations besides those that ensure that
power is continuously delivered to each ablation probe. For example, such a
feature can be applied to prior art probe switching implementation, wherein
power is delivered to one probe at a time, while no power is delivered to the
remaining probes.
It should also be noted that although the previously described tissue ablation systems 10, 110 have been described as being mono-polar ablation systems, the attenuation-switching functions described above can be applied to bipolar ablation systems. In these types of systems, RF current is delivered to the electrode tines in a bipolar fashion, which means that current will pass between two electrode tines (“positive” and “negative” electrodes) of a single ablation probe, or between the electrode tines of the two different ablation probes. In the former case, the tissue ablation system will generally perform the attenuation and switching functions in the same manner described above. In the latter case, some of the output connectors will operate as positive poles to which nominal or attenuated power signals are presented and delivered to the connected ablation probes, and some of the output connectors on the power multiplexor will act as negative poles from which nominal or attenuated power signals are received from the connected ablation probes after passage through intervening tissue.

Although the previous tissue ablation systems 10, 110 have been described as providing power to the ablation probes using a single RF source, it should be noted that multiple RF sources can be used to provide the same attenuation and switching functions as a single RF source—although such multiple RF sources may require more complex timing circuitry in order to coordinate the different functions of the RF sources.

For example, one RF generator can be used to sequentially delivered nominal power signal to ablation probes, while delivering no power signal to the remaining ablation probes to which the nominal power signal is currently
not being delivered. In this case, a power multiplexor with no attenuation feature can be used to switch power between the ablation probes. Another RF generator can be used to deliver attenuated power signal to the ablation probes that are currently not being delivered power from the first RF generator. Control circuitry is preferably used to coordinate the timing of both RF generators, so that when the nominal power signal is delivered to one ablation probe, the attenuated power signal is delivered to the other ablation probes.

As another example, a number of RF generators equal to the number of ablation probes can be used. In this case, each RF generator is dedicated to an ablation probe, and alternately provides nominal and attenuated power signals to that ablation probe. Control circuitry is used to maintain a phase difference between the RF generators, so that the nominal power signal is sequentially delivered to each ablation probe, while the attenuated power signal is delivered to the remaining ablation probes to which the nominal power signal is currently not being delivered.
CLAIMS

1. A tissue ablation system, comprising:
   an ablation source having a common power output;
   a power splitter having a splitter input and first and second splitter outputs,
   the splitter input coupled to the common power output;
   a power attenuator having an attenuator input and an attenuator output,
   wherein the attenuator input is coupled to the second splitter output; and
   a switch device having first and second switch inputs respectively coupled
   to the first splitter output and the attenuator output, and a plurality of switch
   outputs coupleable or coupled to a respective plurality of tissue ablation probes,
   wherein the switch device is configured for sequentially coupling the first switch
   input to each one of the switch outputs, while coupling the second switch input to
   the switch outputs currently not coupled to the first switch input.

2. The tissue ablation system of claim 1, wherein the ablation source
   is a radio frequency (RF) ablation source.

3. The tissue ablation system of claims 1 or 2, wherein the first and
   second splitter outputs are substantially unattenuated relative to the common
   power output.

4. The tissue ablation system of any of claims 1 - 3, wherein the
   power attenuator is variable.

5. The tissue ablation system of claim 4, further comprising a
   feedback control circuit configured for receiving a feedback input and varying the
   power attenuator based on the feedback input.
6. The tissue ablation system of any of claims 1 - 5, wherein the power attenuator is configured for attenuating power at a level equal to or greater than 3 dB.

7. The tissue ablation system of any of claims 1 - 5, wherein the power attenuator is configured for attenuating power at a level within a range of 3-6 dB.

8. The tissue ablation system of any of claims 1 - 7, wherein the plurality of switch outputs comprises at least three outputs.

9. The tissue ablation system of any of claims 1 - 8, wherein the switch device is configured for sequentially coupling the first splitter output to each switch output at a fixed rate greater than once per second.

10. The tissue ablation system of any of claims 1 - 8, wherein the switch device is configured for sequentially coupling the first switch input to each switch output at a fixed rate less than once per thirty seconds.

11. The tissue ablation system of claim 1, further comprising a control circuit configured for receiving a feedback input and causing the switch device to sequentially couple the first switch input to each switch output based on the feedback input.

12. The tissue ablation system of any of claims 1 - 11, further comprising the tissue ablation probes.

13. A tissue ablation system, comprising:

an ablation source configured for generating a common power signal;
a power multiplexor configured for splitting the common power signal into
first and second power signals, substantially attenuating the second power signal
relative to the first power signal, and delivering the first power signal to one or
more of a plurality of tissue ablation probes, while delivering the second power
signal to a different one or more of the plurality of ablation probes.

14. The tissue ablation system of claim 13, wherein the ablation source
is a radio frequency (RF) ablation source.

15. The tissue ablation system of claims 13 or 14, wherein the power
multiplexor is configured for not substantially attenuating the first power signal.

16. The tissue ablation system of claims 13 or 14, wherein the power
multiplexor is configured for varying a value that the second power signal is
attenuated.

17. The tissue ablation system of claim 16, wherein the power
multiplexor is configured for receiving a feedback input and varying the
attenuation value based on the feedback input.

18. The tissue ablation system of claims 13 or 14, wherein the power
multiplexor is configured for attenuating the second power signal at a level equal
to or greater than 3 dB.

19. The tissue ablation system of claims 13 or 14, wherein the power
multiplexor is configured for attenuating the second power signal at a level within
a range of 3-6 dB.

20. The tissue ablation system of any of claims 13 - 19, wherein the
plurality of ablation probes comprises at least three tissue ablation probes.
21. The tissue ablation system of any of claims 13 - 19, wherein the power multiplexor is configured for sequentially delivering the first power signal to different sets of the tissue ablation probes, while delivering the second power signal to the tissue ablation probes to which the first power signal is not currently delivered.

22. The tissue ablation system of claim 21, wherein sets of tissue ablation probes to which the first power signal is delivered only contains a single ablation probe.

23. The tissue ablation system of claim 21, wherein the power multiplexor is configured for sequentially delivering the first power signal to each tissue ablation probe set at a rate greater than one probe per second.

24. The tissue ablation system of claim 21, wherein the power multiplexor is configured for sequentially delivering the first power signal to each tissue ablation probe set at a rate less than one probe per thirty seconds.

25. The tissue ablation system of claim 21, wherein the power multiplexor is configured for receiving a feedback input and sequentially delivering the first power signal to each tissue ablation probe set based on the feedback input.

26. The tissue ablation system of claim 13, wherein the power multiplexor is configured for varying the number of tissue ablation probes to which it delivers the first and second power signals.

27. The tissue ablation system of any of claims 13 - 26, further comprising the tissue ablation probes.
**FIG. 4**

![Waveform Diagram](Image)

**FIG. 5**

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FIG. 7

110

Power Multiplier

FIG. 9

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FIG. 8

PROBE 1
100%
50%

PROBE 2
100%
50%

PROBE 3
100%
50%

PROBE 4
100%
50%

PROBE 5
100%
50%

PROBE 6
100%
50%

T1 T2 T3 T4 T5 T6 T7 T8 T9 T10 T11 t
FIG. 10

POWER MULTIPLEXOR

FROM RF GENERATOR

POWER SPLITTER

ATTENUATOR

SWITCH ELEMENT 1

SWITCH ELEMENT 2

SWITCH ELEMENT 3

SWITCH ELEMENT 4

SWITCH ELEMENT 5

SWITCH ELEMENT 6

TO PROBE 1 (USED)

TO PROBE 2 (USED)

TO PROBE 3 (USED)

CONTROL CIRCUIT

UNUSED

UNUSED

UNUSED
INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER
INV. A61B18/12
ADD. A61B18/14

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)
A61B

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)
EPO-Internal, WPI Data

C. DOCUMENTS CONSIDERED TO BE RELEVANT

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<td>US 5 496 312 A (KLICEK ET AL) 5 March 1996 (1996–03–05) the whole document</td>
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Further documents are listed in the continuation of Box C.

See patent family annex.

Date of the actual completion of the international search
12 July 2006

Date of mailing of the international search report
25/07/2006

Name and mailing address of the ISA 
European Patent Office, P.B. 5818 Patentlaan 2 NL–2280 HV Rijswijk
Tel. (+31–70) 340–2040, Tx. 31 651 epo nl, Fax (+31–70) 340–3016

Authorized officer
Sopelana Martinez, J
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