Medical Device

The present invention relates to the field of medical devices and in particular, although not exclusively, to medical cauterization and cutting devices. The invention also relates to drive circuits and methods for driving such medical devices.

Many surgical procedures require cutting or ligating blood vessels or other internal tissue. Many surgical procedures are performed using minimally invasive techniques, a hand-held cauterization device is used by the surgeon to perform the cutting or ligating. The existing hand-held cauterization devices require a desk top power supply and control electronics that are connected to the device through an electrical supply line. Figure 10 illustrates such an existing hand-held cauterization device currently in use.

It has been known for a number of years that these existing devices are cumbersome and difficult to use during a surgical operation due to the large size of the supply and control electronics and due to the tethering of the hand-held cauterization device to the supply and control electronics. It has also been known for a number of years that these problems would be overcome by providing a battery powered hand-held cauterization device in which the power and control electronics are mounted within the device itself, such as within the handle of the device. However, it is not a simple matter of miniaturising the electronics. The power that has to be supplied to the device during the surgical procedure and the current design of the electronics is such that large capacitors, inductors and transformers as well as heat sinks and fans are required. Figure 11 illustrates in more detail the different parts of the supply and control electronics that are used in the existing design. Whilst it is possible to reduce the size of the sensing and control electronics, other parts of the circuitry cannot be miniaturised in this way.

In particular, the existing electronics design uses circuitry for providing an adjustable 24 Volt power supply; FETs and associated drive circuitry; a transformer for increasing the supply voltage; and filtering circuitry to remove harmonics from the square wave voltage levels that are generated by the FET switches and the transformer. Given the voltage levels and the power levels used to drive the cauterization device, the transformers and

output filters all have to be relatively bulky devices and large heat sinks and a fan are required to cool the FET switches.

The present invention aims to provide an alterative circuit design that will allow the miniaturisation of the circuitry so that it can be built into the hand-held cauterization device, whilst still being able to provide the power and control required for the medical procedure.

The present invention provides a medical device comprising an end effector having at least one electrical contact a radio frequency, RF, generation circuit for generating an RF drive signal and to provide the RF drive signal to the at least one electrical contact and wherein the RF generation circuit comprises a resonant circuit. In one embodiment, the radio frequency generation circuit comprises switching circuitry that generates a cyclically varying signal, such as a square wave signal, from a DC supply and the resonant circuit is configured to receive the cyclically varying signal from the switching circuitry. The DC supply is preferably provided by one or more batteries that can be mounted in a housing (such as a handle) of the device.

According to another aspect, the invention provides a medical device comprising: a handle for gripping by a user; an end effector coupled to the handle, the end effector having at least one electrical contact; battery terminals for connecting to one or more batteries; a radio frequency, RF, generation circuit coupled to said battery terminals and operable to generate an RF drive signal and to provide the RF drive signal to the at least one electrical contact of said end effector; wherein the frequency generation circuit comprises: switching circuitry for generating a cyclically varying signal (which may be a square wave pulse width modulated signal) from a potential difference across said battery terminals; and a resonant drive circuit coupled to said switching circuitry and operable to filter the cyclically varying signal generated by the switching circuitry; and wherein the RF drive signal is obtained using an output signal from said resonant circuit.

The medical device may also comprise a control circuit (which may comprise hardware and/or software) that varies the frequency of the RF drive signal. The control circuit may vary the frequency based on a measurement of the RF drive signal in order to control at

least one of the power, voltage and/or current delivered to the at least one electrical contact of the end effector. In a preferred embodiment, the measurement is obtained from a sampling circuit that operates synchronously with respect to the frequency of the RF drive signal. The frequency at which the sampling circuit samples the sensed signal may be an integer fraction of the frequency of the RF drive signal.

In one embodiment, the control circuit varies the frequency of the RF drive signal around (preferably just above or just below) the resonant frequency of the resonant circuit. The resonant characteristic of the resonant circuit may vary with a load connected to the at least one electrical contact and the control circuit may be arranged to vary the RF drive frequency to track changes in the resonant characteristic of the resonant circuit.

According to another aspect, the invention provides a medical device comprising: a handle for gripping by a user; an end effector coupled to the handle and having at least one electrical contact; a radio frequency, RF, generation circuit operable to generate an RF drive signal and to provide the RF drive signal to the at least one electrical contact; and a control circuit operable to vary the frequency of the RF drive signal to control at least one of the power, the voltage and the current provided to the at least one contact of the end effector.

The RF generation circuit may comprise a signal generator that generates a cyclically varying signal at the RF frequency; and a frequency dependent attenuator that attenuates the cyclically varying signal in dependence upon the frequency of the cyclically varying signal. The frequency dependent attenuator may be a lossless attenuator and may comprise a resonant circuit having a resonant frequency at or near the RF frequency of the cyclically varying signal.

The present invention also provides a medical device comprising: a handle for gripping by a user; an end effector coupled to the handle and having at least one electrical contact; a radio frequency, RF, generation circuit operable to generate an RF drive signal and to provide the RF drive signal to the at least one electrical contact; an input for receiving a sensed signal that varies with the RF drive signal applied to the at least one electrical contact; a sampling circuit for sampling the sensed signal received at said input; a measurement circuit operable to make measurements of the RF drive signal using samples obtained from the sampling

circuit; and a control circuit operable to control the RF generation circuit in dependence upon the measurements made by the measurement circuit, to vary the frequency of the generated RF drive signal; wherein the sampling circuit is operable to sample the sensed signal at a sampling frequency that varies in synchronism with the frequency of the RF drive signal.

The invention also provides a method of operating a medical device comprising generating an RF signal and applying the RF signal to at least one electrode of an end effector of the medical device and controlling the frequency of the generated RF signal to control at least one of the power, current, and voltage applied to the at least one electrode.

According to another aspect, the invention provides a method of cauterising a vessel or tissue, the method comprising: gripping the vessel or tissue with an end effector of a medical device; applying an RF signal to at least one electrode of the end effector that is in contact with the vessel or tissue; and controlling the frequency of the RF signal to control at least one of the power, current, and voltage applied to the tissue to perform the cauterisation.

The above methods may use the above described medical device, although that is not essential.

The controlling step may vary the frequency of the RF signal to control the power applied to the tissue or vessel, and the method may further comprise obtaining measurements of the impedance of the tissue or vessel and varying the desired power applied to the tissue or vessel in dependence upon the obtained impedance measurements.

These and various other features and aspects of the invention will become apparent from the following detailed description of embodiments which are described with reference to the accompanying Figures in which:

Figure 1 illustrates a hand-held cauterization device that has batteries and drive and control circuitry mounted into a handle portion of the device;

Figure 2 is a part block part schematic diagram illustrating the main components of the RF drive circuitry and control circuitry used in one embodiment of the invention;

Figure 3 is a block diagram illustrating the main components of a controller used to control the operation of the RF drive circuitry illustrated in Figure 2;

Figure 4 is a timing diagram illustrating the RF drive signals applied to the cauterization device and illustrating a way in which synchronous samples may be obtained to measure the drive signals;

Figure 5a is a plot illustrating limits that are placed on voltage and current supplied to the cauterization device illustrated in Figure 1;

Figure 5b illustrates a resulting power plot obtained by combining the current and voltage plots illustrated in Figure 5a;

Figure 6 is a plot illustrating the way in which the resonant characteristics of the RF drive circuit illustrated in Figure 2 varies with different loads;

Figure 7 is a flow chart illustrating the operation of a frequency control algorithm used to control the frequency of the RF drive signals applied to the cauterization device;

Figure 8 is a plot illustrating one way in which the power limit can be varied by the control electronics during a surgical procedure;

Figure 9 is a part block part schematic diagram illustrating the main components of another RF drive circuit and control circuit embodying the invention;

Figure 10 illustrates the form of a prior art hand-held cauterization device which is connected to power supply and control electronics via a power supply line; and

Figure 11 is a plan view illustrating the different components of the existing electronics used to drive and control the hand-held cauterization device illustrated in Figure 10.

Medical Device

Many surgical procedures require cutting or ligating blood vessels or other vascular tissue. With minimally invasive surgery, surgeons perform surgical operations through a small incision in the patient's body. As a result of the limited space, surgeons often have difficulty controlling bleeding by clamping and/or tying-off transected blood vessels. By utilizing electrosurgical forceps, a surgeon can cauterize, coagulate/desiccate, and/or simply reduce or slow bleeding by controlling the electrosurgical energy applied through jaw members of the electrosurgical forceps.

Figure 1 illustrates the form of an electrosurgical medical device 1 that is designed for minimally invasive medical procedures, according to one embodiment of the present invention. As shown, the device 1 is a self contained device, having an elongate shaft 3 that has a handle 5 connected to the proximal end of the shaft 3 and an end effector 7 connected to the distal end of the shaft 3. In this embodiment, the end effector 7 comprises medical forceps 9 and a cutting blade (not shown) that are controlled by the user manipulating control levers 11 and 13 of the handle 5.

During a surgical procedure, the shaft 3 is inserted through a trocar to gain access to the patient's interior and the operating site. The surgeon will manipulate the forceps 9 using the handle 5 and the control levers 11 and 13 until the forceps 9 are located around the vessel to be cauterised. Electrical energy at an RF frequency (it has been found that frequencies above about 50kHz do not affect the human nervous system) is then applied, in a controlled manner, to the forceps 9 to perform the desired cauterisation. As shown in Figure 1, in this embodiment, the handle 5 houses batteries 15 and control electronics 17 for generating and controlling the electrical energy required to perform the cauterisation. In this way, the device 1 is self contained in the sense that it does not need a separate control box and supply wire to provide the electrical energy to the forceps 9.

RF Drive Circuitry

Figure 2 is a part schematic part block diagram illustrating the RF drive and control circuitry 20 used in this embodiment to generate and control the RF electrical energy supplied to the forceps 9. As will be explained in more detail below, in this embodiment,

the drive circuitry 20 is a resonant based circuit and the control circuitry operates to control the operating frequency of the drive signal so that it is varied around the resonant frequency of the drive circuit, which in turn controls the amount of power supplied to the forceps 9. The way that this is achieved will become apparent from the following description.

As shown in Figure 2, the drive circuitry 20 comprises the above described batteries 15 that are arranged to supply, in this example, 0V and 24V rails. An input capacitor (C_{in}) 21 is connected between the 0V and the 24V rails for providing a low source impedance. A pair of FET switches 23-1 and 23-2 (both of which are N-channel in this embodiment to reduce power losses) is connected in series between the 0V rail and the 24V rail. FET gate drive circuitry 25 is provided that generates two drive signals – one for driving each of the two FETs 23. The FET gate drive circuitry 25 generates drive signals that causes the upper FET (23-1) to be on when the lower FET (23-2) is off and vice versa. This causes the node 27 to be alternately connected to the 24V rail (when FET 23-1 is switched on) and the 0V rail (when the FET 23-2 is switched on). Figure 2 also shows the internal parasitic diodes 28-1 and 28-2 of the corresponding FETs 23, which conduct during any periods that the FETs 23 are open.

As shown in Figure 2, the node 27 is connected to a capacitor-inductor-inductor resonant circuit 28 formed by capacitor C_s 29, inductor L_s 31 and inductor L_m 33. The FET gate driving circuitry 25 is arranged to generate drive signals at a drive frequency (f_d) that opens and closes the FET switches 23 at around the resonant frequency of the resonant circuit 28. As a result of the resonant characteristic of the resonant circuit 28, the square wave voltage at node 27 will cause a substantially sinusoidal current at the drive frequency (f_d) to flow within the resonant circuit 28. As illustrated in Figure 2, the inductor L_m 33 is the primary of a transformer 35, the secondary of which is formed by inductor L_{sec} 37. The transformer 35 up-converts the drive voltage (V_d) across inductor L_m 33 to the load voltage (V_L) that is applied to the load (represented by the load resistance R_{load} 39 in Figure 2) corresponding to the impedance of the forceps' jaws and any tissue or vessel gripped by the forceps 9. As shown in Figure 2, a pair of DC blocking capacitors C_{bl} 40-1 and 40-2 is provided to prevent any DC signal being applied to the load 39.

In this embodiment, the amount of electrical power supplied to the forceps 9 is controlled by varying the frequency of the switching signals used to switch the FETs 23. This works because the resonant circuit 28 acts as a frequency dependent (lossless) attenuator. The closer the drive signal is to the resonant frequency of the resonant circuit 28, the less the drive signal is attenuated. Similarly, as the frequency of the drive signal is moved away from the resonant frequency of the circuit 28, the more the drive signal is attenuated and so the power supplied to the load reduces. In this embodiment, the frequency of the switching signals generated by the FET gate drive circuitry 25 is controlled by a controller 41 based on a desired power to be delivered to the load 39 and measurements of the load voltage (V_L) and of the load current (i_L) obtained by conventional voltage sensing circuitry 43 and current sensing circuitry 45. The way that the controller 41 operates will be described in more detail below.

Controller

Figure 3 is a block diagram illustrating the main components of the controller 41. In this embodiment, the controller 41 is a micro-processor based controller and so most of the components illustrated in Figure 3 are software based components. However, a hardware based controller 41 may be used instead. As shown, the controller 41 includes synchronous I,Q sampling circuitry 51 that receives the sensed voltage and current signals from the sensing circuitry 43 and 45 and obtains corresponding samples which are passed to a power, V_{rms} and I_{rms} calculation module 53. The calculation module 53 uses the received samples to calculate the RMS voltage and RMS current applied to the load 39 (forceps 9 and tissue/vessel gripped thereby) and from them the power that is presently being supplied to the load 39. The determined values are then passed to a frequency control module 55 and a medical device control module 57. The medical device control module 57 uses the values to determine the present impedance of the load 39 and based on this determined impedance and a pre-defined algorithm, determines what set point power (P_{set}) should be applied to the frequency control module 55. The medical device control module 57 is in turn controlled by signals received from a user input module 59 that receives inputs from the user (for example pressing buttons or activating the control levers 11 or 13 on the handle 5) and also controls output devices

(lights, a display, speaker or the like) on the handle 5 via a user output module 61.

The frequency control module 55 uses the values obtained from the calculation module 53 and the power set point (P_{set}) obtained from the medical device control module 57 and predefined system limits (to be explained below), to determine whether or not to increase or decrease the applied frequency. The result of this decision is then passed to a square wave generation module 63 which, in this embodiment, increments or decrements the frequency of a square wave signal that it generates by 1kHz, depending on the received decision. As those skilled in the art will appreciate, in an alternative embodiment, the frequency control module 55 may determine not only whether to increase or decrease the frequency, but also the amount of frequency change required. In this case, the square wave generation module 63 would generate the corresponding square wave signal with the desired frequency shift. In this embodiment, the square wave signal generated by the square wave generation module 63 is output to the FET gate drive circuitry 25, which amplifies the signal and then applies it to the FET 23-1. The FET gate drive circuitry 25 also inverts the signal applied to the FET 23-1 and applies the inverted signal to the FET 23-2.

Drive Signals and Signal Measurements

Figure 4 is a signal plot illustrating the switching signals applied to the FETs 23; a sinusoidal signal representing the measured current or voltage applied to the load 39; and the timings when the synchronous sampling circuitry 51 samples the sensed load voltage and load current. In particular, Figure 4 shows the switching signal (labelled PWM1H) applied to upper FET 23-1 and the switching signal (labelled PWM1L) applied to lower FET 23-2. Although not illustrated for simplicity, there is a dead time between PWM1H and PWM1L to ensure that that both FETs 23 are not on at the same time. Figure 4 also shows the measured load voltage/current (labelled OUTPUT). Both the load voltage and the load current will be a sinusoidal waveform, although they may be out of phase, depending on the impedance of the load 39. As shown, the load current and load voltage are at the same drive frequency (f_d) as the switching signals (PWM1H and PWM1L) used to switch the FETs 23. Normally, when sampling a sinusoidal signal, it is necessary to sample the signal at a rate corresponding to at least twice the

frequency of the signal being sampled – i.e. two samples per period. However, as the controller 41 knows the frequency of the switching signals, the synchronous sampling circuit 51 can sample the measured voltage/current signal at a lower rate. In this embodiment, the synchronous sampling circuit 51 samples the measured signal once per period, but at different phases in adjacent periods. In Figure 4, this is illustrated by the "I" sample and the "Q" sample. The timing that the synchronous sampling circuit 51 makes these samples is controlled, in this embodiment, by the two control signals PWM2 and PWM3, which have a fixed phase relative to the switching signals (PWM1H and PWM1L) and are out of phase with each other (preferably by quarter of the period as this makes the subsequent calculations easier). As shown, the synchronous sampling circuit 51 obtains an "I" sample on every other rising edge of the PWM2 signal and the synchronous sampling circuit 51 obtains a "Q" sample on every other rising edge of the PWM3 signal. The synchronous sampling circuit 51 generates the PWM2 and PWM3 control signals from the square wave signal output by the square wave generator 63 (which is at the same frequency as the switching signals PWM1H and PWM1L). Thus when the frequency of the switching signals is changed, the frequency of the sampling control signals PWM2 and PWM3 also changes (whilst their relative phases stay the same). In this way, the sampling circuitry 51 continuously changes the timing at which it samples the sensed voltage and current signals as the frequency of the drive signal is changed so that the samples are always taken at the same time points within the period of the drive signal. Therefore, the sampling circuit 51 is performing a "synchronous" sampling operation instead of a more conventional sampling operation that just samples the input signal at a fixed sampling rate defined by a fixed sampling clock.

The samples obtained by the synchronous sampling circuitry 51 are then passed to the power, V_{rms} and I_{rms} calculation module 53 which can determine the magnitude and phase of the measured signal from just one "I" sample and one "Q" sample of the load current and load voltage. However, in this embodiment, to achieve some averaging, the calculation module 53 averages consecutive "I" samples to provide an average "I" value and consecutive "Q" samples to provide an average "Q" value; and then uses the average I and Q values to determine the magnitude and phase of the measured signal (in a conventional manner). As those skilled in the art will appreciate, with a drive

frequency of about 400kHz and sampling once per period means that the synchronous sampling circuit 51 will have a sampling rate of 400kHz and the calculation module 53 will produce a voltage measure and a current measure every 0.01ms. The operation of the synchronous sampling circuit 51 offers an improvement over existing products, where measurements can not be made at the same rate and where only magnitude information is available (the phase information being lost).

Limits

As with any system, there are certain limits that can be placed on the power, current and voltage that can be delivered to the forceps 9. The limits used in this embodiment and how they are controlled will now be described.

In this embodiment, the RF drive circuitry 20 is designed to deliver a power limited sine wave into tissue with the following requirements:

1) Supplied with a nominally 24V DC supply
2) Substantially sinusoidal output waveform at approximately 400kHz
3) Power limited output of 45W
4) Current limited to $1.4A_{rms}$ and voltage limited to $85V_{rms}$

The last two requirements are represented graphically in Figures 5a and 5b. In particular, Figure 5a illustrates idealised plots of voltage and current for loads between 1 Ohm and 10k Ohms on a logarithmic scale; and Figure 5b illustrates the power delivered to the load 39 for loads between 1 Ohm and 10k Ohms.

The frequency control module 55 maintains data defining these limits and uses them to control the decision about whether to increase or decrease the excitation frequency.

Resonant Characteristic and Frequency Control

As mentioned above, the amount of electrical power supplied to the forceps 9 is controlled by varying the frequency of the switching signals used to switch the FETs 23. This is achieved by utilising the fact that the impedance of the resonant circuit 28 changes rapidly with frequency. Therefore by changing the frequency of the switching signals, the magnitude

of the current through the resonant circuit 28, and hence through the load 39, can be varied as required to regulate the output power.

As those skilled in the art will appreciate, the resonant circuit 28 is coupled to a load 39 whose impedance will vary during the surgical procedure. Indeed the medical device control module 57 uses this variation to determine whether the tissue or vessel has been cauterised, coagulated/desiccated. The varying impedance of the load 39 changes the frequency characteristic of the RF drive circuit 20 and hence the current that flows through the resonant circuit 28. This is illustrated in Figure 6, which is a plot 65 illustrating the way in which the current through the resonant circuit 28 varies with the drive frequency for a fixed value of load impedance. As the impedance of the load 39 increases, the resonant characteristic 65 will change shape (the peak may grow or reduce in height) and will move to the left and as the impedance of the load decreases it will change its shape and move to the right. Therefore, the frequency control module 55 must operate quickly enough to track the changes in the resonant characteristic 65. This is easily achievable in this embodiment, where power, current and voltage measurements are available every 0.01ms. In general terms, measurements would only be required at a rate of about once every 0.1s to track the changes. However sudden changes in the resonant characteristic 65 can occur, which the frequency control module 55 cannot track. When this happens, the frequency control module 55 resets the operating frequency to a value where it knows that it will be on one side of the characteristic.

As the impedance of the resonant circuit 28 increases sharply both above and below resonance, it is possible to operate the RF drive circuit 20 either above or below the resonant frequency. In this embodiment, the frequency control module 55 controls the operation of the drive circuit 20 so that it operates slightly above the resonant frequency as this should lead to lower switching losses through the FETs 23.

Figure 7 illustrates the processing performed in this embodiment by the calculation module 53 and the frequency control module 55. As shown, at the beginning of the process in step s1, the control module 55 turns on the RF drive signal at the system defined maximum frequency by passing an initialisation signal to the square wave generation module 63. Provided the control module 55 has not received, in step s3, a power down signal from the

medical device control module 57, the processing proceeds to step s5 where the calculation module 53 obtains the voltage and current samples from the synchronous sampling circuitry 51. In step s7 the calculation module 53 calculates the square of the voltage and the square of the current and the delivered power by multiplying the measured voltage by the measured current. These calculated values are then passed to the frequency control module 55 which compares, in step s9, the values with the defined limits for the applied voltage, current and power. The voltage and current limits are static limits that are defined in advance. However, the power limit depends on the medical procedure and is defined by the power set point (P_{set}) provided by the medical device control module 57. If each of the measured values is below the corresponding limit then, in step s11, the frequency control module 55 decides to decrease the drive frequency and a decrease command is passed to the square wave generator 63. At the start of the processing, the drive frequency is set to a defined maximum value (in this embodiment 500kHz), which will always be above the resonant peak of the characteristic 65, regardless of the load impedance. Therefore, regardless of the load 39, the initial operating frequency should be on the right hand side of the resonant plot shown in Figure 6. By decreasing the drive frequency, the drive frequency will get closer to the resonant frequency of the resonant circuit 28. As a result, the applied current will increase and more power will be delivered to the load 39. The processing then returns to step s3 and the above process is repeated.

Therefore, the current and power applied to the load 39 should increase until one of the limits is reached. At this point, the control module 55 will determine, in step s9, that a limit has been reached and so will proceed to step s13, where the control module 55 decides to increase the drive frequency and sends the square wave generation module 63 an increase command. This will cause the drive frequency to move away from the resonant frequency of the circuit 28 and so the current and power delivered to the load 39 will reduce. The processing will then return to step s3 as before.

Thus, by starting on one side of the resonant peak and slowly moving the drive frequency towards and away from the resonant peak, the current and power level applied to the load 39 can be controlled within the defined limits even as the impedance of the load changes and the resonant characteristic 65 of the resonant circuit 28 changes as the tissue/vessel is cauterised.

As those skilled in the art will appreciate, it would also be possible to start on the left hand side of the resonant peak and increase the drive frequency to increase the delivered power and decrease the drive frequency to decrease the delivered power.

Medical Device Control Module

As mentioned above, the medical device control module 57 controls the general operation of the cauterisation device 1. It receives user inputs via the user input module 59. These inputs may specify that the jaws of the forceps 9 are now gripping a vessel or tissue and that the user wishes to begin cauterisation. In response, in this embodiment, the medical device control module 57 initiates a cauterisation control procedure. Initially, the medical device control module 57 sends an initiation signal to the frequency control module 55 and obtains current and power measurements from the calculation module 53. The medical device control module 57 then checks the obtained values to make sure that the load 39 is not open circuit or short circuit. If it is not, then the medical device control module 57 starts to vary the power set point to perform the desired cauterisation. Figure 8 is a plot illustrating the way in which the medical device control module 57 may vary the set point power to achieve the desired cauterisation procedure. Various other techniques and other power delivery algorithms may also be used.

As shown in Figure 8, during an initial period 71 the medical device control module 57 pulses the set point power between zero and about 10 Watts. Then during a main cauterisation period 73 (which typically lasts for about 5 seconds) the medical device control module 57 pulses the set point power between zero and 50 Watts. During this period, the medical control device receives the power and voltage measurements from the calculation module 53 and calculates from them the impedance of the load 39. The medical device control module 57 determines that the cauterisation is complete when the calculated impedance exceeds a threshold. Finally, the medical device control module 57 performs a terminating procedure during a terminating period 75. During the terminating procedure, the medical device control module 57 varies the set point power and checks that cauterisation has been achieved (by checking the impedance of the load using the measured power and current values) and re-enters the main cauterisation period again if it determines that cauterisation has not been completed.

Resonant Circuit Design

The way that the values of the inductors and capacitors were chosen in this embodiment will now be described. As those skilled in the art will appreciate, other design methodologies may be used.

The complex impedance of the circuit shown in Figure 2 can be approximated by the following equation:

$$Z = j2\pi fL_s + \frac{1}{j2\pi fC_s} + \frac{j2\pi fL_m R_{load_ref}}{j2\pi fL_m + R_{load_ref}} + R_s \quad (1)$$

Where:

R_{load_ref} is the load resistance referred to the primary (by the square of the turns ratio);

R_s represents the equivalent series resistance of the inductor, transformer capacitor and switching devices.

All other component non-idealities are ignored and the transformer is considered to be ideal as a first approximation.

Assuming that R_s is small, when the load is open circuit (ie R_{load_ref} is infinite) the resonant frequency can be shown to be:

$$f_{min} = \frac{1}{2\pi\sqrt{(L_s + L_m)C_s}} \quad (2)$$

Similarly, when the load is short circuit (ie R_{load_ref} is zero) the resonant frequency can be shown to be:

$$f_{max} = \frac{1}{2\pi\sqrt{L_s C_s}} \quad (3)$$

Assuming R_s is small: at each frequency between f_{min} and f_{max} there is a value of the load, R_{load}, at which the greatest power can be dissipated in the load. This maximum power can be shown to be large at frequencies near f_{min} and f_{max}, and has a minimum at the critical frequency, fc. We refer to this power as P_{max_fc}. Starting with (1) it can be shown that the following relationship holds:

$$L_m = \frac{2V_s^2}{2\pi f_c P_{max_fc}} \quad (4)$$

where V_s is the supply voltage.

It can be shown that the load at which equation (4) holds is given by:

$$R_{load_ref} = 2\pi f L_m \quad (5)$$

Furthermore from (1) a relationship between f_{min}, f_c and f_{max} can be established:

$$f_{min} = \sqrt{(2\pi f_c)^2 \frac{(2\pi f_{max})^2 + (2\pi f_c)^2}{3(2\pi f_{max})^2 - (2\pi f_c)^2}} \quad (6)$$

From (6) it can be shown that $f_{min}<f_c<f_{max}$. If the circuit is to operate at f_c, then equation (4) gives an upper bound on the worst-case power delivered across a range of loads.

From (1), it can be shown that the efficiency of the circuit at resonance may be written as:

$$\eta = \frac{\left(\frac{(2\pi f_c)^2 L_m^2 R_{load}}{(2\pi f_c)^2 L_m^2 + R_{load}^2}\right)}{\mathrm{Re}(Z)} = \frac{(2\pi f_c)^2 L_m^2 R_{load_ref}}{R_s\left((2\pi f_c)^2 L_m^2 + R_{load_ref}^2\right) + (2\pi f_c)^2 L_m^2 R_{load_ref}} \quad (7)$$

From (7) it may be shown that the efficiency is a maximum when $R_{load_ref} = 2\pi f L_m$, i.e. when (5) holds. Therefore the system is designed to operate around the point of maximum efficiency.

Design procedure

For this specific embodiment of the design the following parameters were chosen:

- Battery voltage of 24V however battery voltage droops with discharge and load so V_{s_sq}=18V (square wave peak to peak voltage) was used
- P_{load} = 45W (maximum power into the load)
- V_{load} = 85Vrms (maximum voltage into the load)
- I_{load} = 1.4Arms (maximum current into the load)
- f_c = 430kHz (centre or critical switching frequency)
- f_{max} = 500kHz (maximum switching frequency, which is the upper resonant frequency)
- f_{min} ≈ 380kHz (approximate minimum switching frequency - needs to be calculated)

Given these values, f_{min} can be computed using (6):

$$f_{min} = \sqrt{(2\pi 430k)^2 \frac{(2\pi 500k)^2 + (2\pi 430k)^2}{3(2\pi 500k)^2 - (2\pi 430k)^2}}$$

$$f_{min} = 377kHz$$

Resonant circuits produce sinusoidal waveforms therefore the input square wave voltage (V_{s_sq}) needs to be converted into the RMS of the fundamental switching frequency (V_s).

$$V_s = \frac{4}{\pi}\frac{Vs_sq}{2\sqrt{2}}$$

$$= \frac{4}{\pi}\frac{18V}{2\sqrt{2}}$$

$$= 8.1V_{rms}$$

The power into the load (P_{load}) is set by L_m. Using (4) the transformer magnetising inductance (L_m) can be determined. This ensures that at the critical frequency, f_c, the required power is delivered:

$$L_m = \frac{2V_s^{\,2}}{2\pi f_c P_{load}}$$

$$= \frac{2 \times 8.1 V_{rms}^{\,2}}{2\pi \times 430kHz \times 45W}$$

$$= 1.08\mu H$$

L_s can then be calculated (derived from equations 2 & 3):

$$L_s = \frac{L_m}{\frac{f_{max}^{\,2}}{f_{min}^{\,2}} - 1}$$

$$= \frac{1.08\mu H}{\frac{500kHz}{377kHz} - 1}$$

$$= 1.43\mu H$$

Following from this C_s can be calculated (from equation 3):

$$C_s = \frac{1}{L_s (2\pi f_{max})^2}$$

$$= \frac{1}{1.43uH (2\pi 500kHz)^2}$$

$$= 71nF$$

To maintain regulation, the circuit is run above resonance so actual values of C_s will be typically 20% higher to bring the operating point back down (if below resonance was chosen C_s would have to be reduced).

As previously mentioned, the efficiency is maximised when R_{load_ref} is equal to the magnetising reactance at the critical frequency (equation 5). It is desirable, therefore, to operate about the middle of the constant power range (shown in Figure 5b). R_{load_upper} is the load resistance at which constant power changes to constant voltage. Similarly, R_{load_lower} is the load resistance at which constant power changes to constant current.

$$R_{load_upper} = \frac{V_{load}^2}{P}$$

$$= \frac{85V_{rms}^2}{45W}$$

$$= 161\Omega$$

$$R_{load_lower} = \frac{P}{I_{load}^2}$$

$$= \frac{45W}{1.4A^2}$$

$$= 23\Omega$$

Take the geometric mean of these load resistances to find R_{load_c} (centre or critical load resistance)

$$R_{load_c} = \sqrt{R_{load_upper} R_{load_lower}}$$

$$= 60\Omega$$

As discussed, for maximum efficiency, R_{load_ref} should match the impedance of the primary-referred magnetising reactance at f_c. Hence R_{load} should equal the secondary-referred magnetising reactance. L_{sec} can therefore be calculated as follows:

$$L_{sec} = \frac{R_{load_c}}{2\pi f_c}$$

$$= \frac{60}{2\pi 430kHz}$$

$$= 22.2\mu H$$

Finally the transformer turns ratio can be calculated:

$$N = \sqrt{\frac{L_{sec}}{L_m}}$$

$$= \sqrt{\frac{22.2\mu H}{1.08\mu H}}$$

$$= 4.5$$

For any particular design it may be necessary to adjust the values due to the following reasons:

- to maximise efficiency
- compensate non ideal effect of components (e.g. series resistance, parasitic capacitance & inductance, non ideal transformer characteristics such as leakage inductance)
- make the design practical (e.g. use standard values of capacitors and a whole number of turns
- allow margin to meet the requirements due to component tolerances, temperature etc

In this specific embodiment, the component values were optimised to:

Cs = 82nF

Lm = 1.1uH

Ls = 1.4uH

N = 5 which gives Lsec = 24uH

The following subsections briefly describe how these component values were physically implemented.

Capacitor selection

A low loss capacitor is desired to minimise losses and to ensure the component doesn't get too hot. Ceramic capacitors are ideal and the dielectric type of COG/NPO were used in this embodiment. The capacitor voltage rating is also important as it shouldn't be exceeded under all load conditions. Ten 250V 8.2nF 1206 COG/NPO ceramics capacitors in parallel were used in this embodiment.

Inductor and transformer

In this embodiment, Ferroxcube 3F3 E32/6/20 e-core/plate combination was used as a ferrite core. Ferroxcube 3F3 is supplied by Ferroxcube, a subsidiary of Yageo Corporation, Taiwan. It is a high frequency ferrite material optimised for frequencies between 200kHz and 500kHz. By using this material the core losses are minimised. Core losses increase strongly with increasing flux density. In an inductor, for a particular required energy storage, the flux density increases with decreasing air gap (the air gap is the separation between the e-core & plate). Therefore the air gap and the number of turns can be increased to decrease core losses but this has to be balanced with the actual inductance value required and increased resistive losses introduced with the longer wire/track length.

The same issues apply to the transformer except core losses are due to the output voltage and the number of turns. Since the output voltage is fixed the number of turns is the only variable that can be changed but again this has to be balanced with resistive losses. Once the number of turns is set the air gap can then be adjusted to set Lm. Whatever core is used, it is best practise to fill the winding space with as much copper as possible to minimise resistive losses. In the transformer the volume of windings is preferably about the same in the primary and secondary to balance the losses.

The resistive losses can usually be easily calculated but since the circuit is operating at about 400kHz skin depth becomes an issue. The skin depth in copper at 400kHz is only about 0.1mm so a solid conductor thicker than this doesn't result in all the copper being used. Litz wire (stranded insulated copper wire twisted together where each strand is thinner

than the skin depth) can be used to reduce this effect. In this embodiment 2 oz PCB tracks (about 0.07mm thick copper tracks) were used for the windings of both the inductor (L_s) and the transformer to avoid having to wind custom components. The inductor had two turns with an air gap of 0.5mm between the e-core and plate. The transformer had one turn on the primary and five turns on the secondary with an air gap between the e-core and plate of 0.1mm.

Modifications and Alternatives

A medical cauterisation device has been described above. As those skilled in the art will appreciate, various modifications can be made and some of these will now be described. Other modifications will be apparent to those skilled in the art.

In the above embodiment, various operating frequencies, currents, voltages etc were described. As those skilled in the art will appreciate, the exact currents, voltages, frequencies, capacitor values, inductor values etc. can all be varied depending on the application and the values described above should not be considered as limiting in any way. However, in general terms, the circuit described above has been designed to provide an RF drive signal to a medical device, where the delivered power is desired to be at least 10W and preferably between 10W and 200W; the delivered voltage is desired to be at least 20 V_{rms} and preferably between 30 V_{rms} and 120 V_{rms}; the delivered current is designed to be at least 0.5 A_{rms} and preferably between 1 A_{rms} and 2 A_{rms}; and the drive frequency is at least 50kHz.

In the above embodiment, the resonant circuit 28 was formed from capacitor-inductor-inductor elements. As those skilled in the art will appreciate, the resonant circuit 28 can be formed from various circuit designs. Figure 9 illustrates another resonant circuit design that can be used in other embodiments. In the design shown in Figure 9, the resonant circuit 28 is formed from capacitor-inductor-capacitor elements, with the load being connected across the second capacitor 78. As shown, in this design, there is no transformer and so there is no step-up in voltage. However, the operation of this embodiment would still be the same as in the embodiment described above and so a further description shall be omitted. Other resonant circuit designs with multiple capacitors and inductors in various series and parallel configurations or simpler LC resonant circuits may also be used.

Figure 1 illustrates one way in which the batteries and the control electronics can be mounted within the handle of the medical device. As those skilled in the art will appreciate, the form factor of the handle may take many different designs.

In the above embodiment, an exemplary control algorithm for performing the cauterisation of the vessel or tissue gripped by the forceps was described. As those skilled in the art will appreciate, various different procedures may be used and the reader is referred to the literature describing the operation of cauterisation devices for further details.

In the above embodiment, the RF drive signal generated by the drive circuitry was directly applied to the two forceps jaws of the medical device. In an alternative embodiment, the drive signal may be applied to one jaw, with the return or ground plane being provided through a separate connection on the tissue or vessel to be cauterised.

In the above embodiments, the forceps jaws were used as the electrodes of the medical device. In an alternative device, the electrodes may be provided separately from the jaws.

In the above embodiments, two FET switches were used to convert the DC voltage provided by the batteries into an alternating signal at the desired RF frequency. As those skilled in the art will appreciate, it is not necessary to use two switches – one switch may be used instead or multiple switches may be used connected, for example, in a bridge configuration. Additionally, although FET switches were used, other switching devices, such as bipolar switches may be used instead. However, MOSFETs are preferred due to their superior performance in terms of low losses when operating at the above described frequencies and current levels.

In the above embodiment, the resonant circuit 28 acted as a frequency dependent attenuator. The resonant circuit was designed as a substantially lossless attenuator, but this is not essential. The resonant circuit may include lossy components as well, although the resulting circuit will of course be less efficient.

In the above embodiment, the I & Q sampling circuitry 51 sampled the sensed voltage/current signal once every period and combined samples from adjacent periods. As

those skilled in the art will appreciate, this is not essential. Because of the synchronous nature of the sampling, samples may be taken more than once per period or once every n^{th} period if desired. The sampling rate used in the above embodiment was chosen to maximise the rate at which measurements were made available to the medical device control module 57 as this allows for better control of the applied power during the cauterisation process.

In the above embodiment, a 24V DC supply was provided. In other embodiments, lower DC voltage sources may be provided. In this case, a larger transformer turns ratio may be provided to increase the load voltage to a desired level or lower operating voltages may be used.

In the above embodiment, a synchronous sampling technique was used to obtain measurements of the load voltage and load current. As those skilled in the art will appreciate, this is not essential and other more conventional sampling techniques can be used instead.

In the above embodiment, the medical device was arranged to deliver a desired power to the electrodes of the end effector. In an alternative embodiment, the device may be arranged to deliver a desired current or voltage level to the electrodes of the end effector.

In the above embodiment the battery is shown integral to the medical device. In an alternative embodiment the battery may be packaged so as to clip on a belt on the surgeon or simply be placed on the Mayo stand. In this embodiment a relatively small two conductor cable would connect the battery pack to the medical device.

Claims

1. A medical device comprising:

a handle for gripping by a user;

an end effector coupled to the handle, the end effector having at least one electrical contact;

battery terminals for connecting to one or more batteries;

a radio frequency, RF, generation circuit coupled to said battery terminals and operable to generate an RF drive signal and to provide the RF drive signal to the at least one electrical contact of said end effector, the one or more batteries and the radio frequency generation circuit housed within the handle;

wherein the radio frequency generation circuit comprises:

switching circuitry for generating a cyclically varying signal from a potential difference across said battery terminals; and

a resonant drive circuit coupled to said switching circuitry and operable to filter the cyclically varying signal generated by the switching circuitry.

2. A device according to claim 1, comprising a control circuit configured to vary the frequency of the RF drive signal.

3. A device according to claim 2, wherein the control circuit is operable to receive a measurement of the RF drive signal and is operable to vary the frequency of the of the RF drive signal to control the power, voltage and/or current delivered to the at least one electrical contact of the end effector.

4. A device according to claim 3, wherein the measurement is obtained from a sampling circuit that samples a sensed voltage or current signal at a sampling frequency that varies in synchronism with the frequency and phase of the RF drive signal.

5. A device according to claim 4, wherein the frequency at which the sampling circuit is operable to sample the sensed signal is an integer fraction of the frequency of the RF drive signal.

6. A device according to any one of claims 1 to 5, wherein the control circuit is configured to vary the frequency of the RF drive signal around the resonant frequency of the resonant circuit.

7. A device according to claim 6, wherein the resonant characteristic of the resonant circuit varies with a load connected to the at least one electrical contact and wherein the control circuit is configured to vary the RF drive frequency to track changes in the resonant characteristic of the resonant circuit.

8. A device according to claim 1, wherein the RF generation circuit comprises a signal generator operable to generate a cyclically varying signal at the RF frequency; and a frequency dependent attenuator that attenuates the cyclically varying signal in dependence upon the frequency of the cyclically varying signal.

9. A device according to claim 8, wherein the frequency dependent attenuator comprises a resonant circuit having a resonant frequency at or near the RF frequency of the cyclically varying signal.

10. A device according to claim 3, further comprising:

an input for receiving a sensed signal that varies with the RF drive signal applied to the at least one electrical contact;

a sampling circuit for sampling the sensed signal received at said input; and

a measurement circuit operable to make measurements of the RF drive signal using samples obtained from the sampling circuit;

wherein said control circuit is operable to control the RF generation circuit in dependence upon the measurements made by the measurement circuit, to vary the frequency of the generated RF drive signal; and

wherein the sampling circuit is operable to sample the sensed signal at a sampling frequency that varies in synchronism with the frequency of the RF drive signal.