Communications

The Mechanism of Cutting in Electrosurgery

WILLIAM M. HONIG

Abstract—A detailed analysis shows how electrosurgical r-f power is localized in the vicinity of the cutting electrode. This localization requires a small radius of curvature for the surgical tip, relatively high r-f voltages, and rapid surgical tip motion. The erratic results sometimes encountered in electrosurgery may be due to a deficiency in the above procedures as well as a lack of care in the use of the r-f leads and ground plates. A further analysis supports the supposition that evolving steam bubbles in the tissue at the surgical tip continuously rupture the tissue and are responsible for the cutting mechanism.

I. INTRODUCTION

The purpose of this paper is to provide a detailed derivation of the significant parameters affecting the distribution of electrosurgical r-f power in tissue. This gives the reasons for some of the erratic results which some medical and dental workers have encountered in this field and may help to settle the controversy which this author has often encountered. In addition an explanation is suggested for the cutting action of the surgical tips.

The use of r-f power in the 2–10 Mhz range has found a field of useful application in electrosurgery. The usual circuit arrangement, shown in Fig. 1, consists of an r-f generator with its low side or ground lead routed to the operating table or to a ground plate placed under the patient. The high side of the r-f generator is routed to the surgical tip and it is this tool which the surgeon uses for his work.

An important medical use for electrosurgical techniques has been in cancer surgery where the use of the cutting electrodes causes a wide swath of tissue destruction on either side of the incision, which is useful for destroying cancer cells which may be migrating from the affected region. The 'cutting' action and the occurrence of the swath of destruction are due to different cutting techniques and surgical electrodes. These effects should be understood separately since both of these effects need not occur together, and one or the other may be preferred in different surgical situations. Although the mechanism for the cutting action and for the swath of destruction are thermal, the cutting action is not like electrocauterization which uses heated nichrome wires to burn through tissue.

The recent application of electrosurgery in dentistry provides a basis for understanding medical and dental electrosurgical cutting action. Oringer [1] has shown that with some simple modifications in the usual medical electrosurgical methods, extremely delicate surgical tasks can be accomplished. The main features of his techniques may be summarized:

1. The electrosurgical tools consist of straight wire tips for cutting and wire loops for scooping out portions of tissue. These wires have diameters that are quite fine; 10 mil (.25 mm) or less.
2. The tips are always wiped clean of biological material between cutting strokes so that the original wire is always in contact with the tissue during the cutting strokes.
3. The voltage settings of the r-f generator are always kept high enough during surgery so that the speed of the tip through the tissue is limited only by the rapidity with which the tissue parts (5–10 centimeters per second, typically). Both short and long incisions proceed at this high velocity although for surgical convenience the long strokes may be broken up into shorter contiguous strokes, but always with each stroke at this high speed.
4. The effect of these methods is to confine the region of tissue destruction to the walls of the incision with a penetration depth as small as a fraction of a millimeter. In practice strokes on beef tissue the incisions looked as though they had been done with a sharp scalpel and with no discoloration or 'browning' that would indicate that any heat had appreciably penetrated the walls of the incision. On 'in vivo' tissue the concurrent advantages of coagulation and hemostasis also occurs.

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The author was with the Department of Electrical Engineering, University of Western Australia, Nedlands, 6009, Australia. He is now with the Department of Electrical Engineering, Western Australian Institute of Technology, Bentley, 6102, W. Aust., Australia.

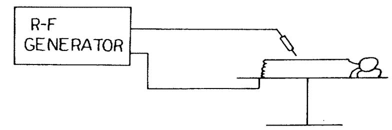


Fig. 1. Electrosurgery, schematic arrangement.

It is, of course, well known to those with electrical backgrounds that for electrodes immersed in tissue, extremely intense electric fields and energy densities exist in the immediate neighborhood of those electrode regions having small radii of curvature.

Calculations of these parameters are given here in greater than normal detail so that the mechanism for 'cutting,' its capability for accomplishing surgical tasks, and the occurence of swaths of destruction can be understood by cognizant medical and dental workers.

This appears to be necessary because of the idea that has been accepted by some, that electrosurgery is a relatively erratic technique which sometimes works well and at other times causes uncontrollable destruction. The clinical effects would appear to be exactly as expected if the correct techniques are used. Lack of attention to the treatment of the r-f leads and the grounds may also be responsible for some of the poor results. This is discussed in Section IV.

II. TISSUE PARAMETERS

The frequency range of interest in electrosurgery is 2–10 Mhz and calculations will be carried out at 5 Mhz. The dielectric constant for tissue from 2 to 10 Mhz appears to be quite a bit larger than several hundred (500–1500 for many measurements in the 2–10 Mhz region [2], [3]). These values of dielectric constant are for tissue whose organic molecular dipole structure has not been destroyed; for measurements using less than 100 millivolts, typically. For electrosurgical r-f voltages, which lie in the range of 30–350 volts R.M.S. and therefore have peak values of 45–500 volts approximately, the dielectric constant is not usually so high. It will be assumed for calculations that the dielectric constant will be approximately equal to that of water (80 approximately), although the error of this assumption will be shown to be negligible when a dielectric constant of 1000 is used.

In the case of the resistivity, ρ, Fig. 2 indicates [2] that a value of 200 ohm-centimeters may be used as an average value. Although these measurements are also made at low voltage testing levels, the resistivity is mainly that of tissue fluids and material rather than structures so that it can be assumed that the heating of tissue due to the passage of r-f currents and the destruction of tissue structure will not radically alter the tissue conductivity.

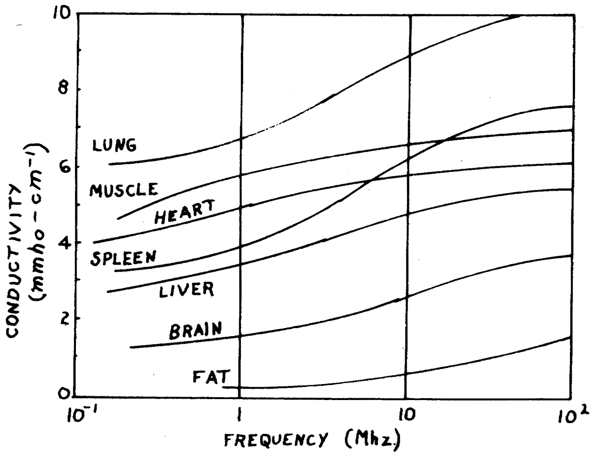


Fig. 2. Conductivity ($1/\rho$) versus frequency.

III. ELECTRICAL PARAMETERS

Suppose a length L (=1cm) of a fine wire with a diameter of .25 mm is completely immersed in a homogeneous and isotropic material with a resistivity of 200 ohm-centimeters and a dielectric constant of 80. The radial current flow (which is in the direction r) and the circular equipotential lines will have a cylindrical symmetry with respect to the axis of the wire, see Fig. 3. The departure from cylindrical symmetry at the wire ends will be considered later and will be shown to confirm the estimates of the tissue electrical parameters and r-f power distribution made here.

The power density, PD, in watts/cm^3, as a function of r must follow the general relation:

$$\mathrm{PD} = K/r^2$$

because of the cylindrical symmetry which makes the electric field lines radial so that the electric field is inversely proportional to r. This makes the power density inversely proportional to r^2. The integration of this density between the two fixed radii a and b, where $a < b$ gives the total power, P_0:

$$P_0 = \int_a^b \frac{K 2\pi r L dr}{r^2} = 2\pi K L \ln (b/a). \tag{1}$$

Substituting for K in Eq. (1) its value in Eq. (2) gives:

$$\mathrm{PD} = P_0/(2\pi r^2 L \ln (b/a)). \tag{2}$$

The power density is the main parameter which is of interest in electrosurgery. The voltage, current, and impedance (or admittance) distributions, however, throughout the tissue are also relevant because these parameters have been incompletely understood by some workers in this field. The effects of external stray capacitance or series reactance in the r-f leads are dependent on the magnitude and distribution of the above parameters. The effect of both the resistivity and the dielectric constant may also be taken into account during this analysis, which will show that in most cases the effect of a dielectric constant of 80 (and even 1000) may be neglected.

The differential resistance, dR, between equipotential surfaces is:

$$dR = \rho \frac{dl}{A} = \rho \cdot \frac{dr}{2\pi r L} \tag{3}$$

where ρ is the resistivity in ohm-cm, r is the radial distance in cm, A is the cross section area to current flow in cm^2, dl is the thickness through which the current flows, and L is the length of the wire. The resistance, R_{ab}, between two equipotential surfaces at the fixed radial distances a and b, where $a < b$, is:

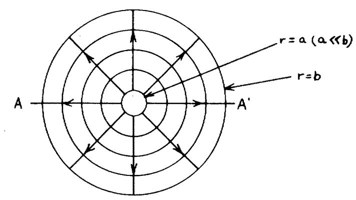


Fig. 3. Electric fields.

$$R_{ab} = \int_0^{R_{ab}} dR = \rho \int_a^b \frac{dr}{2\pi r L} = \frac{\rho}{2\pi L} \cdot \ln (b/a). \tag{4}$$

The capacitance, C_{ab}, for the same case is [4]

$$C_{ab} = .553 e_r 10^{-12} L / \ln (b/a) \tag{5}$$

where C_{ab} is in farads, and e_r is the relative dielectric constant (= 80 for water). The reactance of the capacitor, X_{ab}, in ohms, is:

$$X_{ab} = 1/j\omega C_{ab}. \tag{6}$$

An equivalent circuit for the impedance, Z_{ab}, (or the admittance, Y_{ab}) between the surfaces a and b is shown in Fig. 4(a) where:

$$Z_{ab} = 1/Y_{ab} = 1/(1/R_{ab} + j\omega C_{ab}) \tag{7}$$

where

$$Y_{ab} = (K_1 + jK_2)/\ln (b/a). \tag{8}$$

The complex parameter $K_1 + jK_2$ is:

$$K_1 + jK_2 = \frac{2\pi L}{\rho} + j\omega e_r .553 \times 10^{-12} L. \tag{9}$$

The impedance and admittance between the surface at a and the variable surface at r where $a < r < b$ may be given using the subscript ar in Eqs. (7), (8) above, see Fig. 4(b):

$$Y_{ar} = (K_1 + jK_2)/\ln (r/a) = 1/Z_{ar}. \tag{10}$$

If the total voltage applied between surfaces a and b is V_0, then:

$$P_0 = I_0 V_0 \tag{11}$$

where I_0, V_0, and P_0 are the vector current, vector voltage, and vector power, respectively. Then:

$$P_0 = V_0^2 (K_1 + jK_2)/\ln (b/a). \tag{12}$$

The power in the region between a and r where $r < b$ is:

$$P_{ar} = P_0 \ln (r/a)/\ln (b/a). \tag{13}$$

The voltage between the circular cylindrical surfaces at a and r is V_{ar}:

$$V_{ar} = I_0 Z_{ar} = (V_0/Z_{ab})/Z_{ar} = V_0 \ln (r/a)/\ln (b/a). \tag{14}$$

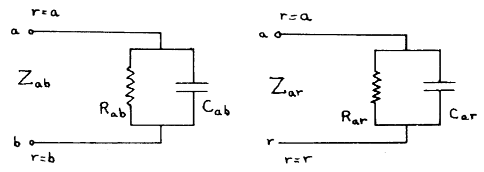
This corresponds to the voltage between the fine wire and some point deeper in the tissue. It is plotted in normalized form against the distance r in Fig. 5(b). It shows that most of the potential drop across the tissue occurs in the region immediately adjacent to a, the location of the fine wire. In Fig. 5(a) is plotted the power density PD as a function of r [this is equation (2)], which shows that the power density is greatest in the vicinity of the fine wire.

The value for $K_1 + jK_2$ is found using $L = 1$ cm, $\rho = 200$ ohm-cm, and $f = 5 \times 10^6$ Hz:

$$K_1 + jK_2 = (3.14 + j.14) \times 10^{-2} \quad \text{for } e_r = 80$$

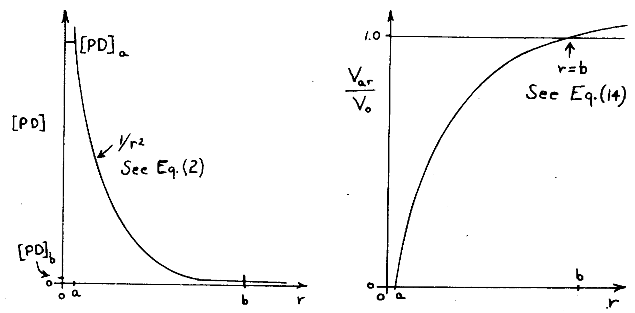
$$K_1 + jK_2 = (3.14 + j1.75) \times 10^{-2} \quad \text{for } e_r = 1000.$$

The value of Z_{ab} can be found from Eqs. (7), (8), using $\rho = 200$ ohm-cm, $a = 12.5$–10^{-3} cm, $b = 100a = 1.25$ cm, and $L = 1$ cm:



(a) (b)

Fig. 4. Equivalent circuits. (a) Z_{ab}. (b) Z_{ar}.



(a) (b)

Fig. 5. Power density (PD) and normalized voltage, V_{ar}/V_0 versus radius. (a) Power density. (b) V_{ar}/V_0.

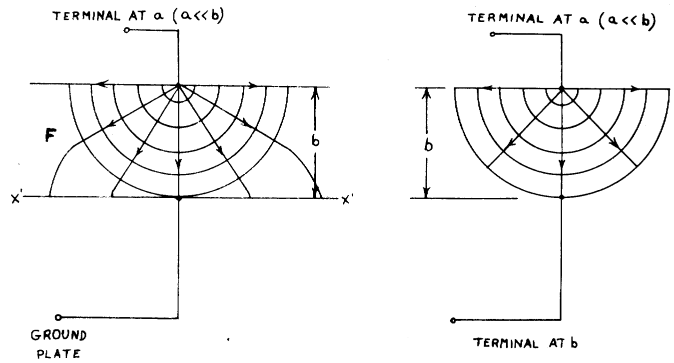
$$Z_{ab}(\text{ohms}) \cong 147 \angle 0^\circ \qquad \text{for } e_r = 80$$

$$Z_{ab}(\text{ohms}) = 128 \angle -29^\circ \quad \text{for } e_r = 1000.$$

These values of total impedance exist between a fine wire and an outer circular cylindrical surface with a radius of 1.25 cm. If one uses 12.5 cm instead as the outer radius then the values of Z_{ab} are increased in the ratio of ln 1000/ln 100 or 1.5 with respect to the values given above. Furthermore, an outer radius of 25 cm will result in values of Z_{ab} that are 1.65 (ln 2000/ln 100) times the original value above. Even a variation of the outer surface radius from 1.25 to 25 cm (which is a 1:20 ratio) will only result in a tissue impedance change in the ratio 1:1.65. When considering a single type of electrosurgical procedure, i.e., for abdomen, arm, leg, etc., the variation in the impedance will be much less than this. For a $\pm 30\%$ variation in tissue thickness, with an average thickness of 7.5 cm and 1.25 cm, the impedance variation will be $\pm 6\%$ and $\pm 7.5\%$, respectively.

In Fig. 3 it is evident that the field distribution has a horizontal plane of symmetry at A–A'. No current lines cross this plane. If the region above A–A' is replaced by another medium having an extremely high resistivity and low dielectric constant (air) then the equipotential lines do not change, although only the lower region will have current flow lines. The electrode in the upper region may now be removed, retaining the b electrode in the lower region of Fig. 3. The electric field lines in the air (upper region) must now terminate at the b electrode of the lower region. The previous argument and the plots of Fig. 5(a) and (b) show that for $r \ll b$ the PD and the major part of the voltage drop in the tissue in localized near $r = a$. This permits one to consider that the field lines in the lower region will not change appreciably if the upper region is considered to be air. Previous results, therefore, which refer to a fine wire 1 cm long completely immersed in tissue may be extended. Those results can also correspond to the case of a wire 2 cm long half immersed in tissue. The voltages, power densities, currents and impedances will be the same for this case (similar scale and immersion changes may also be carried out).

The field lines for a half immersion of this kind are shown in Fig. 6(a). The above discussion also prompts the replacement of this field configuration by that of Fig. 6(b). Here the outer cylindrical surface in Fig. 6(b) is considered to be the same as the flat electrode X' which would be the ground plate of the operating table. As has been discussed above, this amounts to neglecting a 6–7.5% variation in tissue impedance for tissue thicknesses from 7.5 to 1.25 cm, where the particular tissue thickness may vary by as much as $\pm 30\%$. The variation of this tissue impedance at these thicknesses



(a) (b)

Fig. 6. Field approximations. (a) Flat ground plate. (b) Cylindrical ground plate.

corresponds to considering that the region F adds something like $+30\%$ at most to the tissue thickness. It should increase the tissue impedance of Fig. 6(b) by 6–7.5% but it will be entirely neglected.

IV. ADDITIONAL CONSIDERATIONS

The ground plate which is placed under the patient does not contact the tissue directly. It is covered usually with a layer of cloth or plastic so that it represents a series capacitive reactance in the path between the leads to the r-f generator. Assuming a value of area of 2000 cm² (one square foot) and a dielectric thickness of 0.5 mm with a dielectric constant of 2, at 5 Mhz the capacitive reactance is 10 ohms. The actual value is usually less than this due to stray capacitances between the low side of the generator and the patient. Its magnitude will be less, therefore, than 7% of the value of Z_{ab} previously given ($=147$ ohms, $e_r = 80$).

Since frequencies of approximately 5 Mhz are being transmitted along the surgical tip leads and ground leads, the following steps can have a significant effect on the reliability of electrosurgical procedures. These comments are obvious to some, but many medical workers treat r-f leads in a similar way to d-c or 60 cycle a-c leads.

a) Lead lengths longer than several feet which coil and loop and lie against intervening furniture can radically and unreliably alter the r-f voltage at the tissue. The coils and loops introduce additional series reactance (inductive). The long lengths may introduce a capacitive bypass path between the r-f generator leads as will the placing of such excess length on metallic stands and furniture. All of these conditions will reduce the r-f voltage at the tissue but in an unreliable way since these conditions will vary each time the leads are used.

b) The placement of the r-f leads between generator and patient should at least be in an identical geometric arrangement each time they are used.

c) Shielded leads solve the above problems but they must be designed into the output circuit of the r-f generator.

d) The reliance on stray capacitance between the patient and the r-f generator as a makeshift grounding scheme introduces an unknown and unreliable reactance in series with the r-f generator.

It is believed that the above procedures form a significant portion of the operating conditions which have resulted in unsatisfactory electrosurgical effects. For those who are quite familiar with and use correct methods in the handling of r-f leads, the separation of cutting techniques from the generation of swaths of destruction is necessary. This is described in the introduction and in the following sections.

V. POWER ESTIMATES

Table I gives the results of the calculation of power density (PD) under a number of different conditions. The power density (PD) is given both at the wire, $(PD)_a$, and at the edge of the full cylindrical thickness, $(PD)_b$. This is done for three cases of total thickness;

TABLE I

	$b = 1.25$ cm $Z_{ab} = 147\ \Omega$			$b = 7.5$ cm $Z_{ab} = 202\ \Omega$			$b = 12.5$ cm $Z_{ab} = 220\ \Omega$		
Total Power	V_{ab}	$(PD)_a$	$(PD)_b$	V_{ab}	$(PD)_a$	$(PD)_b$	V_{ab}	$(PD)_a$	$(PD)_b$
15 Watts	47.	3.3×10^3	0.3	55.	2.4×10^3	.006	58.	2.2×10^3	.0002
150 Watts	144.	3.3×10^3	3.	170.	$24. \times 10^3$	.06	176.	$22. \times 10^3$	.002

Note: Although the $(PD)_a$ numbers are surprisingly large, it should be noted that the PD drops to 1/16 of those numbers at a distance of .04 mm from the wire in all cases.

TABLE II
HEAT RISE OF TISSUE ASSUMING NO BOILING (SEE TEXT)

		$b = 1.25$ cm			$b = 7.5$ cm			$b = 12.5$ cm	
Position		a	b		a	b		a	b
ΔT (°C)	15 W	830	.08	15 W	600	.001	15 W	550	.0003
ΔT (°C)	150 W	8300	.8	150 W	6000	.01	150 W	5500	.003
Time to boil	Much less than 1 second, in all cases at a; will never boil at b.								

for $b = 1.25$, 7.5, and 12.5 cm. In each of these cases two values of the total input power, P_0, are assumed: 15 and 150 watts. In each of the six combinations of these conditions V_{ab} is also presented for $e_r = 80$. As shown previously, this gives values of Z_{ab} which are sensibly real, so that the PD is sensibly real in all these cases. For the condition that $e_r = 1000$ approximately, the same value of real power and real PD exists in the tissue, but there is a reactive power and reactive PD which is present in the tissue which is approximately equal to 50% of the listed power and PD numbers.

In all cases $a = .0125$ cm and the units of PD are watts/cm³. The PD numbers in Table I, therefore, refer to dissipated power in the tissue.

It can be seen from the values shown that the power density, $(PD)_a$, at the wire is radically higher there than at the distance b where the power density is $(PD)_b$. It should be emphasized that a density of many thousands of watts per cubic centimeter only exists in the immediate vicinity of the wire; the total power dissipated in the tissue remains at the levels given in the total power column. As shown in Fig. 5(a) and (b), the greatest rate of change of PD and V_{ar} occurs in the vicinity of the wire.

VI. HEAT RISE AND THE 'CUTTING' MECHANISM

The heat rise of the tissue adjacent to the wire may now be estimated assuming that its thermal properties are similar to those of water. Since one watt = one joule/second = $\frac{1}{4}$ calorie/second, the temperature rise ΔT_r (in degrees Centigrade per cm³ per second) at a radial distance r is:

$$\Delta T_r = \tfrac{1}{4}(PD)_r.$$

For the cases shown in Table I the temperature rise, assuming no heat flow from the tissue, is given in Table II. For tissue initially at 30° Centigrade the tissue fluids will begin to boil in times of less than one second. This boiling region will be restricted to the immediate vicinity of the wire because only there is the PD high enough to supply enough heat for the process. The 'cutting' mechanism may, therefore, be due to the physical rupturing of the tissue by the expansion of the evolving steam bubbles. The formation of the steam bubbles is a microscopic effect. If one considers that the tissue is a hydrocarbon matrix with water in the interstices, then the formation of the microscopic bubbles can perform such a destructive function. Many of the bubbles may very well be absorbed in the surrounding cooler tissue after rupture has occurred so that little external liberation of gases will be seen. It should be noted that 540 calories per gram (and per cm³) of water are necessary to turn water to steam at atmospheric pressure and 100° Centigrade. This amount of energy is necessary to develop the bubbling process. The heat rise values given in Table II show that such amounts of energy are available, i.e., 540 calories will be available for 540 degrees of heat rise per cm³ per second.

The region at b has the much lower temperature rise, shown in Table II, which the tissue can easily tolerate.

If the wire is always moving so that it is always touching new tissue it will 'cut' its way through this tissue, and only the tissue adjacent to the wire will encounter the rupturing action. It is evident, therefore, that a high surgical tip velocity (5–10 cm/second), a small radius of surgical tip curvature, and a high r-f voltage (without causing sparking to the tissue) are necessary for this mechanism. See [1] for many photos of such surgical procedures.

The effect of slower cutting speeds and the use of a metallic knife edge scalpel can now be seen. The sides of the scalpel will act as plane electrode and cause a uniform power density to exist in the tissue so that much higher powers (and voltages) must be used to separate the tissue. Thermal damage or destruction of the tissue will occur at greater distances from the planar electrodes. In addition, at low tip speeds even with wire electrodes this effeect can be generated, and a slow scalpel speed will cause additional destruction (browning and burning) at the sharp edge of the blade. It is evident that the fine wire electrodes need less total power to 'cut' and that there will be no cooking of the tissue that is due to the sides of the scalpel blade. Conversely, the slower speeds with broad area blades can be useful for deliberately causing a wide swath of destruction.

VII. FINAL CONSIDERATIONS

The departure from cylindrical symmetry which occurs at the end points of the wire length L that has been considered is such that the curves given in Fig. 5(a) and (b) will be even steeper at the wire and at the end point regions. This is because the spreading of the current flow lines at the ends of the wire will be similar to the case of a spherically shaped electrode instead of the cylindrical shapes which have been considered. The equations for R and C similar to Eqs. (3), (4), (5) are:

$$dR = \rho^{dr}/2\pi r^2$$

$$R = \rho/2\pi a$$

$$C = .553 \times 10^{-12} e_r a$$

where $b \gg a$ and the power density PD will vary as $1/r^3$ instead of $1/r^2$ as in Fig. 5; thus localization of tissue destruction will be even greater in this case than for the cylindrical case.

Sometimes other electrodes known as coagulation electrodes are used. They are in the shape of small spheres or flat surfaces which are momentarily touched to a bleeding surface in order to cause coagulation (and stop bleeding). The surface 'sizzle' or heat rise causes this effect. It appears to occur best when using momentary contact and when half-wave rectified DC power is supplied to the r-f oscillator. In order to confine this effect to the surface of the tissue where it is needed, it would seem to be more reasonable to use an electrode consisting of a fine wire mesh over a plane area. In this case lower voltage and longer duration times can be used. A longer duration time can be more accurately judged than a momentary contact. Since bleeding can occur in blood vessels of different size and depth, coagulation under these different conditions can be effected more easily by varying the duration of contact.

VIII. SUMMARY

The combination of fine wire electrodes, high r-f voltage, and high cutting speeds [1] are necessary for the confinment of tissue destruction in electrosurgery. This uses the intense fields generated at the cutting electrode to maximum advantage. The parting of the tissue may be due to the generation of steam bubbles which rupture the tissue only in the immediate vicinity of the fine wire cutting electrode.

Such techniques would appear to be of great value in microsurgery since localization of the electrosurgical effects would also be accompanied by coagulation and hemostasis.

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Grounding and Safety

D. J. HATCH AND M. B. RABER

Abstract—**Several basic facts about the effects of steel conduits in A.C. power systems are reviewed to show that the geometry of a grounding path may have a greater effect on its effectiveness as a ground return path than its D.C. resistance. Data are presented on the effect of 1/2″ E.M.T. conduit and No. 10 conductors in tests simulating regular room wiring under ground fault conditions.**

It is shown that an electrical system making use of steel conduit with an internal grounding conductor tied to the conduit produces the lowest effective impedance of grounding circuit under ground-fault conditions. The voltage rise (of the order of volts) is the lowest practical limit to voltage differences in the ground circuit; additional grounding paths external to the conduit make little appreciable difference to the voltage rise due to a fault current. A separate grounding conductor external to the conduit, when used by itself, produces a much larger voltage rise than the conduit system. Because of these facts, equipotential bonding schemes must not be separate from the power-system grounding or they will be of questionable value in terms of safety, and may even be hazardous.

It is important to realize that under ground fault conditions, the voltage rises are of the order of volts under the best of conditions, and the action of the total conductive structure must be considered, not just one portion alone.

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The authors are with the Department of Biomedical Engineering, Health Sciences Centre, Winnipeg, Man., Canada.

GROUNDING—WHAT IS IT?

In this electrical age, we believe that we naturally and intuitively understand "grounding."[1] Is it not just the connection of things to the earth so that they will all be held to earth potential (and so be safe)? This simplistic view obscures our understanding of how grounding actually functions. We forget that when carrying a current a ground conductor will have a voltage rise across it, and if the current is alternating current the voltage will be proportional to the impedance not just the resistance. We assume that we understand grounding and yet we are prepared to accept unknown grounding effects such as "ground loops." And still we feel safe if grounded—particularly for patient safety.

These uncertainties have come about because we use grounding for many different purposes without specifically identifying them:

1) Lightning Protection.
2) Fault Current Protection—by providing a low impedance return path so that over-current protective devices will operate.
3) Limiting voltage rise on accessible metal surfaces during a fault.
4) Providing a common reference for measurement and transmission of signals.
5) Shielding of circuits.
6) Many, Many, others.

Insofar as patient-safety is concerned, the prime requirement is to limit voltage differences. Since voltage differences are caused by fault currents in the power system, we must consider the effectiveness of grounding methods in achieving a low impedance path that will allow easy return of the fault current, and produce the minimum voltage rise.

It is unfortunately true that those of us who usually work with milliamperes and small voltages have lost sight of some of the basics which should govern the design of power systems, while power system designers are quite unused to concerning themselves with very small voltage differences. If a grounding system is to be "effective" it must be designed to meet all of its requirements, therefore these requirements must be defined.

This communication points out some of the not generally realized facts of grounding circuits and the behavior of fault currents in A.C. systems. The first of these is that A.C. circuits, even at the low frequency of 60 Hz. cannot be treated as D.C. circuits just by consideration of resistances. In fact, in appropriate circumstances a part of an A.C. circuit can act like an impedance lower in value than its D.C. resistance!

The second fact is that we cannot consider the patient area in isolation from the rest of the structure! The basic assumption made is that potential differences in the grounding system arise because of fault-currents flowing in the system. Even induced currents—so-called ground-loops, are faults, and are evidence of defects in design or construction. A further assumption is that special requirements for patient-care areas should add to, and must not detract from, the regular protective methods.

With these assumptions, this communication will emphasize the fact that the conduit, with its associated wiring system, is the limiting factor in determining the minimum voltage rise on the grounding circuit, and that additional external grounding produces little effect

[1] Everyone "knows" that things are "safe" if "grounded" because the "electrical code" demands grounding: Yet even the electrical codes mirror our uncertainty about grounding when they state that there shall be no objectionable passage of current over the grounding conductors and they allow interruption of some ground paths in order to achieve this—i.e., NFPA-70-1971, National Electrical Code (U.S.A.) Sec. 250-21; C.S.A. C22.1-1972, Canadian Electrical Code, Part I, Section 10-200.