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Basic Radiofrequency: Physics and Safety and Application to Aesthetic Medicine

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Abstract

This chapter summarizes the basic science of radiofrequency (RF) and its application in aesthetic medicine. The main parameters of RF including RF frequency, waveform, power, pulse duration, and penetration depth are described, and its application for treatment is analyzed. Monopolar and bipolar devices are described in detail for different clinical applications. The effect of RF electrode geometry on tissue heating is shown, and tissue-specific electrical parameters are summarized. The chapter discusses which RF parameters are required to reach therapeutic temperatures for tissue ablation, coagulation, or subnecrotic heating. RF parameters used for noninvasive, minimally invasive, and fractional treatment are compared. Finally, the chapter explains the main safety concerns associated with RF treatments and details the most common causes of adverse events.

The term radiofrequency (RF) was first introduced with the invention of radio and was applied to electromagnetic radiation or current ranging from 3 kHz to 300 GHz. Since then, the field of medicine has used the relatively narrow band of this spectrum from 200 kHz to 40 MHz in many different applications. The main advantage of RF energy in medicine is a low or negligible reaction of nerves to high-frequency alternating current (AC) in comparison to lower frequencies.

William T. Bovie invented the first electrosurgical device while working at Harvard [1]. This device was used by Dr. Harvey Williams Cushing on October 1, 1926, at Peter Bent Brigham Hospital in Boston, Mass., to remove a tissue mass from a patient's head [2]. Since then, RF electrosurgical devices have become one of the most useful surgical instruments. Recently, RF has experienced a resurgence in aesthetic medicine with applications for ablative and nonablative applications. RF energy has become an irreplaceable tool in almost every field of medicine including dermatology, plastic surgery, and aesthetic medicine, the primary interest of this book. The tissue effects achievable using RF energy are based on a versatile thermal end point and are dependent on the applied energy density.

Several RF-induced thermal changes of tissue are commonly used in medicine:

(1) Ablation of tissue. This effect is generally used for cutting or removing tissue and is based on thermal evaporation of tissue. Ablation requires very high energy density, allowing conversion of tissue from a solid state to vapor with minimal thermal damage to the surrounding tissue [3]. A new use for RF ablation is for cautery of tumors.

(2) Coagulation. When applied to blood vessels, coagulation provides hemostasis for controlling bleeding during surgery. The same mechanism is effective for vascular lesion treatment [27]. Coagulation may be applied to soft tissue as well, to induce necrosis when immediate tissue removal is not required or not practical.

(3) Collagen contraction. High temperatures induce immediate transformation in the tertiary structure of proteins. When applied to collagen, heating allows tissue shape to change for medical and cosmetic purposes. Immediate, predictable collagen contraction occurs at a temperature range of 60–80°C in orthopedic procedures [4] and ophthalmology [5]. For noninvasive cosmetic procedures, this effect is produced with lower temperatures in order to avoid skin necrosis. However, due to the lower temperatures, the outcome of the procedure is often less consistent, requires multiple procedures, and takes a longer time to show results [6, 7].

(4) Tissue hyperthermia. Heating of tissue to superphysiologic temperatures is a popular method of skin treatment using subnecrotic temperatures to stimulate natural physiological processes in attempts to modify skin appearance and to reduce subcutaneous fat [8, 9]. This heating does not induce immediate effects of coagulation but can stimulate fibroblasts to synthesize collagen and may alter the metabolism of adipocytes in favor of lipolysis.

Radiofrequency Energy Characteristics

The clinical effects of RF depend on a combination of the RF parameters and on the method of its application to the tissue.

Radiofrequency Frequency

The frequency of electrical current characterizes how many times per second an electrical current changes its direction and is reported in hertz. This change in direction is associated with a change of voltage polarity. Direct current has a frequency of 0 Hz, which is typically used in battery-powered devices. Standard AC in the range of 50–

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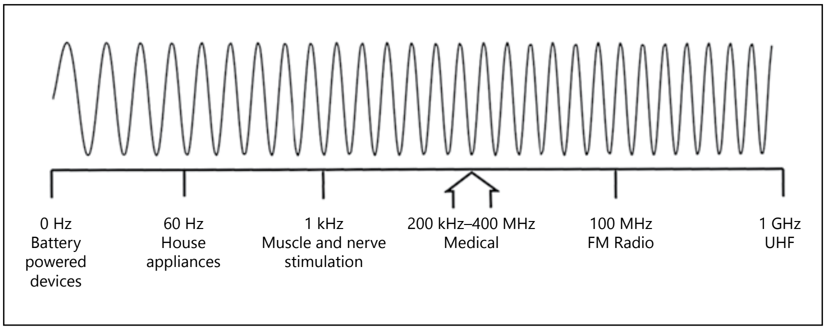

Fig. 1. Frequency spectrum of electrical current.

60 Hz is used for most home appliances. AC current causes nerve and muscle stimulation and at high powers is very dangerous. It can cause acute pain, muscle spasms, and even cardiac arrest.

At a frequency of 100 kHz and higher, the muscle- and nerve-stimulating effects decrease. In this range, higher power can be applied to the tissue safely to create the desired thermal effect (fig. 1). Although at frequencies above 100 Hz nerve reaction from electrical current is dramatically diminished, at high amplitudes skin reaction can be observed even at a frequency above 1 MHz. RF energy propagates in the tissue in the form of electrical current between applied electrodes and in the form of radiation at higher frequencies. Frequencies in the range of 200 kHz to 6 MHz are the most common in medicine, but there are devices with frequencies up to 40 MHz [10]. The higher frequency electrical oscillations are used mostly for communication.

Radiofrequency Waveform

Typically, sine RF voltage is used in medical devices. The RF energy can be delivered in continuous wave (CW) mode, burst mode and pulsed mode (fig. 2). For gradual treatment of large areas, the CW mode is most useful as it allows a slow increase in temperature in bulk tissue. This approach is applied for targeting cellulite, subcutaneous fat, and skin tightening. The burst mode delivers RF energy with repetitive pulses of RF energy. It is used in applications where peak power is important while average power should be limited. This application is used in blood vessel coagulation. Pulsed mode is optimal when the goal is to heat a small tissue volume while limiting heat conduction to the surrounding tissue, similar to the rationale of applying short pulse duration in laser treatments. Pulsed mode is effective for fractional skin ablation and is characterized by pulse durations which do not exceed the thermal relaxation time (TRT) of treated zone.

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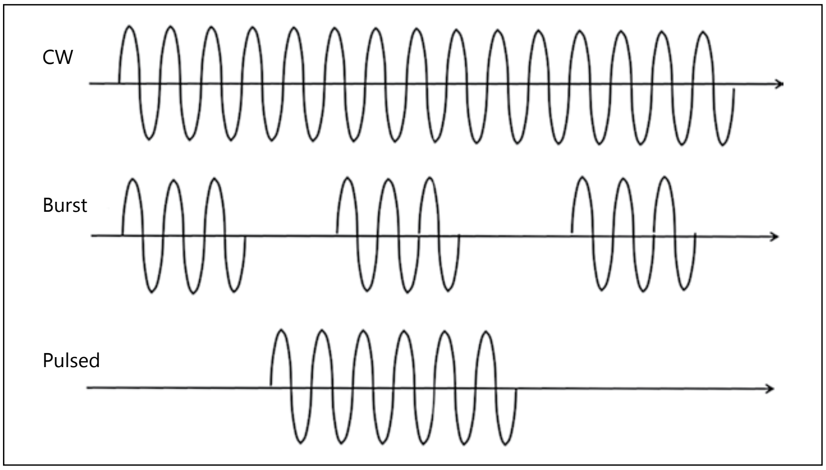

Fig. 2. Typical RF waveforms.

Radiofrequency Power

The most important characteristics of RF energy are its peak and average power. Peak power is important to estimate the thermal effect produced, while average power affects the speed at which the heating is induced. For CW operation mode, the peak and average power are the same. For pulsed or burst mode, the average power is the total power delivered divided by the time the device is applied, including the 'off' cycles.

Another important characteristic of RF is power density. High power applied to a large skin surface may create only gentle warming, but when applied through a needle electrode, the same power is applied over a small contact point, leading to high power density. At high power densities, RF may create intense tissue ablation rather than warming or coagulation.

Thermal Effect of Radiofrequency Current

The heat power *(P)* generated in a tissue volume by electrical current during a period of time *(t)* is described by Joule's law:

$$P = \frac{j^2}{\sigma}. \qquad (1)$$

The heat generated is measured in joules/cm^3. As the equation describes, power increases as a square function of the RF current density *(j)*. Conversely, heating power changes in inverse proportion to tissue conductivity *(σ)*.

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Taking into account that current density according to Ohm's law is proportional to the electric field strength and tissue conductivity (equation 2),

$$j = \sigma E \qquad (2),$$

we can rewrite the equation (1) as

$$P = \sigma E^2 \qquad (3),$$

In other words, the higher the tissue conductivity, the greater the heat that will be generated when constant RF voltage is applied between the electrodes. In addition, the amount of heat generated increases with increasing exposure to RF; stated differently, tissue will heat more with longer duration of RF current. As tissue heats, its conductivity increases (or, stated differently, impedance decreases), and the equations are therefore relevant only at a given time. This is taken into account during RF procedures: in modern devices, RF power is automatically adjusted to tissue impedance.

Penetration Depth and Radiofrequency Energy Distribution Between Electrodes

Penetration depth is a parameter broadly used in laser dermatology to mean the distance below the skin which is heated. More correctly, the depth of RF effect is characterized by attenuation of applied energy with the depth. The most common understanding of this parameter is a depth where applied energy is decreased by an exponential factor (e ~2.7). In contrast to optical energy, which is attenuated with distance of travel through tissue as a result of scattering and absorption, RF current decreases at a distance from the electrode due to the divergence of current lines. The depth of penetration can be affected by altering the topology of the skin and optimizing the electrode system. In aesthetic medicine, the most common configurations of electrode systems are monopolar, bipolar, and multipolar including fractional, where the effect is achieved by superposition of RF current paths between paired electrodes. Penetration depth also can be affected by the anatomical structure of treated area. For example, penetration depth over a bone can be limited by low conductivity of bone tissue. For this reason, treatment parameters over bone, for example the forehead and hip, often differ from the parameters applied in adjacent areas.

Monopolar Radiofrequency Systems

Monopolar RF devices utilize an active electrode in the treatment area and a return electrode, usually in the form of a grounding pad with a large contact area, which is placed outside of the treatment zone. In this electrode geometry, a high RF current

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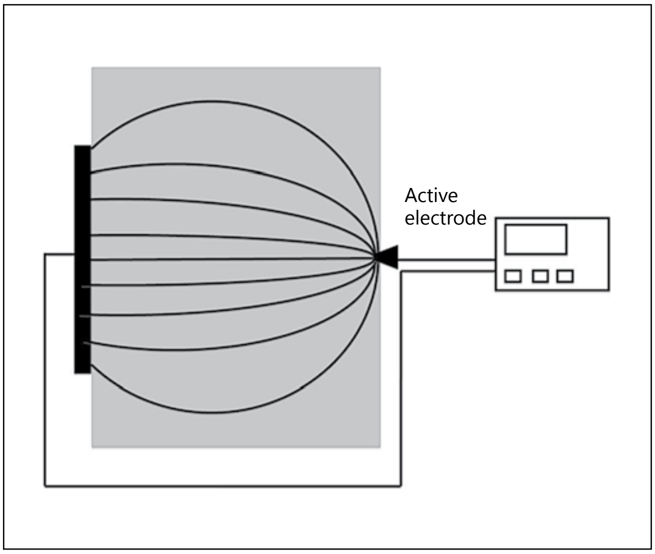

Fig. 3. Schematic of RF current distribution between electrodes for a monopolar system.

density is created near the active electrode, and the RF current diverges toward the large return electrode. Schematically, RF current behavior in the body for a monopolar system is depicted in figure 3.

The heat zone for this geometry can be estimated using an analytic spherical model for the continuity equation, stating that electrical current flows continuously from one electrode to another:

$$\nabla_r j = 0 \tag{4}$$

Taking into account Ohm's law in differential form (equation 2) and the definition of an electric field, equation 4 can be rewritten as:

$$\frac{1}{r^2}\frac{\partial}{\partial r}r^2\frac{\partial\varphi}{\partial r} = 0, \tag{5}$$

where φ is the potential of the electric field. The solution for this equation provides the RF current density distribution between electrodes:

$$j = \frac{\sigma V r_0 R}{r^2\left(R - r_0\right)}, \tag{6}$$

where σ is tissue conductivity, V is voltage between electrodes, r_0 is radius of small electrode and R is the radius of the large electrode.

For the instance when the return electrode is much larger than the active electrode, the equation can be simplified as:

$$j = \frac{\sigma V r_0}{r^2}. \tag{7}$$

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Fig. 4. Thermography of tissue in cross-section during treatment: a monopolar RF generator with a frequency of 1 MHz and 50-watt power was applied using a 1-mm electrode at the tissue surface and a large 100-cm^2 return electrode at the bottom of the tissue. The heat is concentrated near the surface of the small electrode, and the depth of thermal zone is half of the electrode size.

Correspondently, heat power according to Joule's law can be estimated as:

$$P = \frac{\sigma V^2 r_0^2}{r^4} \tag{8}$$

This simple equation leads to a few interesting conclusions:

(1) Heat generated by RF current near the active electrode does not depend on the size, shape, or position of the return electrode when the return electrode is much larger in size than the active electrode and is located at a distance which is much greater than the size of the active electrode.

(2) Heating decreases dramatically as distance increases from the electrode. At a distance equal to the electrode size, heating becomes insignificant. In other words, most of the RF energy applied in monopolar systems is converted into heat near the active electrode. Therefore, the heat zone can be estimated as a radius or half size of active electrode.

(3) RF current is concentrated on the RF electrode and rapidly diverges toward the return electrode. Figure 4 shows a thermal image in cross-section of bovine tissue treated with a monopolar electrode and demonstrates that heat generation is observed near the active electrode only.

Monopolar devices are most commonly used for tissue cutting. Schematically, the RF current flow for monopolar devices is shown in figure 5.

RF current always flows in a closed loop via the human body. As shown above, the current density far from the active electrode is negligible. However, a malfunction in which low frequency current escapes from a monopolar configuration holds high risk because the entire body is exposed to the electrical energy. Most commercially available devices have isolated output to help avoid any unexpected RF current path to the surrounding metal equipment.

Treatment effects with monopolar devices depend on the density of RF energy, which can be controlled with RF power, and the size of active electrode. In order to create tissue ablation, very high energy density is required. In cutting instruments, a needle type electrode is used to concentrate electrical current on a very small area.

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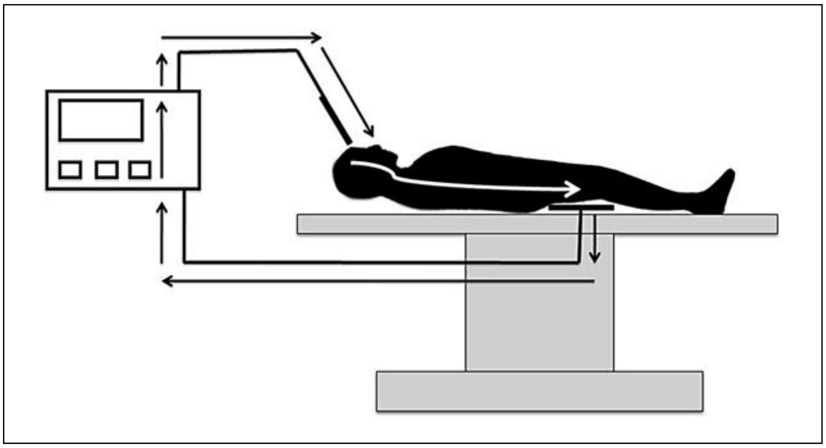

Fig. 5. Electrical current flowing through the patient and electrosurgical device.

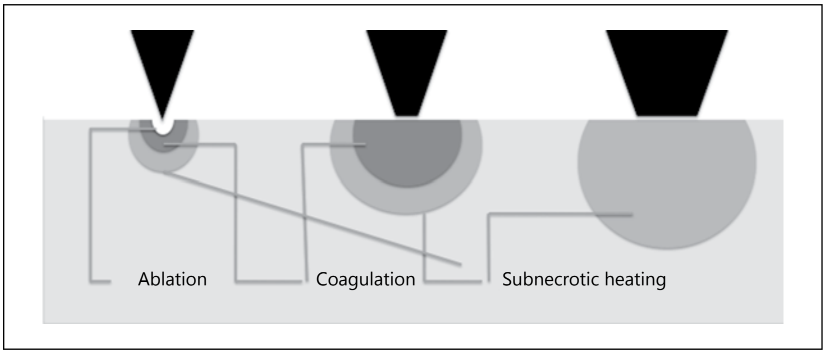

Fig. 6. Spot size effect.

Coagulation hand pieces have a larger surface area than ablative devices, usually a few square millimeters, to generate heat on a larger area, creating coagulation rather than ablation. Subnecrotic heating is usually used for treatments related to collagen remodeling, and in this case the spot size is about 1 cm^2 [7]. A schematic of the spot size effect on the treatment area is shown in figure 6.

For monopolar devices, the penetration depth is a function of the active electrode size and can be estimated as a half the electrode size.

The main features of monopolar devices are:

- Predictability of thermal effect near the active electrode
- Ability to concentrate energy on a very small area
- High nonuniformity of heat distribution, with very high heat at the surface of the active electrode and dramatic reduction at a distance exceeding the size of electrode, thereby limiting penetration depth.

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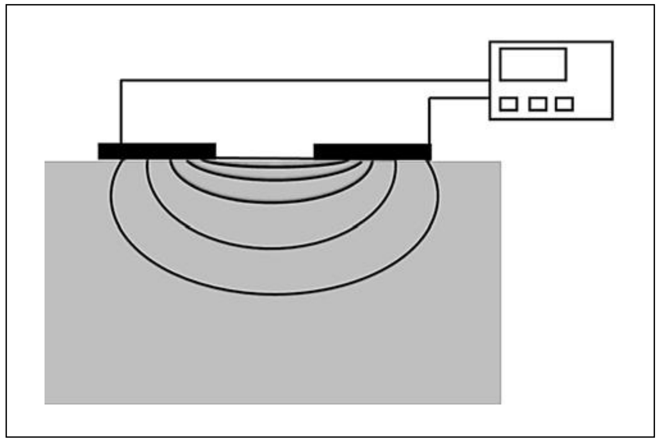

Fig. 7. Electrical current distribution for a bipolar RF system.

Bipolar Radiofrequency Systems

Bipolar configuration is characterized by the use of two electrodes which are in contact with the treated area. This geometry is better able to create uniform heating in larger volume of tissue than a monopolar system. In order to understand heat distribution between electrodes, the following three rules should be taken into the account:

(1) For any geometry, RF current density is higher along the line of shortest distance between the electrodes and reduced with distance from the electrodes.

(2) Heating is greater near the electrode surface and drops with distance because of current divergence.

(3) RF current is concentrated on the part of the electrode that has high curvature, creating hot spots.

A schematic distribution of electrical currents in uniform media for typical electrode geometries used for noninvasive treatment is shown in figure 7.

In bipolar devices, both electrodes create an equal thermal effect near each of the electrodes, and the divergence of RF current is not strong because of the small distance between the electrodes. For bipolar systems shown in figure 7, most of the heat is concentrated between electrodes.

Penetration depth of RF for bipolar devices is a function of electrode size and the distance between them. By increasing the distance between the electrodes, electrical current can go deeper, but divergence is also increased. For the case when the distance between the electrodes is much larger than the electrode size, the heating profile will be similar to two monopolar electrodes. Schematically this situation is shown in figure 8.

Thermal images of tissue cross-section for small and large distance between electrodes are shown in figure 9.

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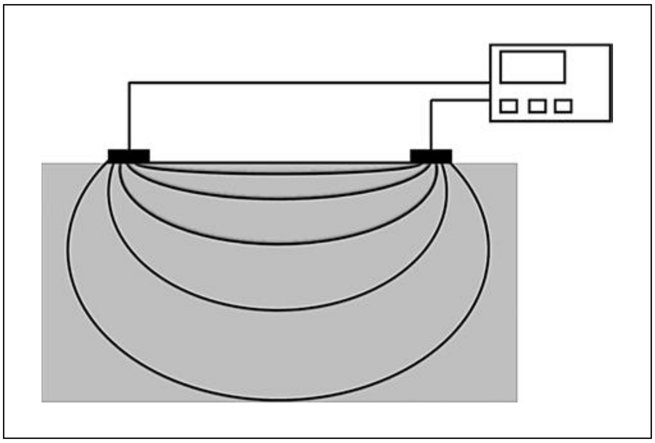

Fig. 8. Electrical current distribution for a bipolar system with a large distance between electrodes.

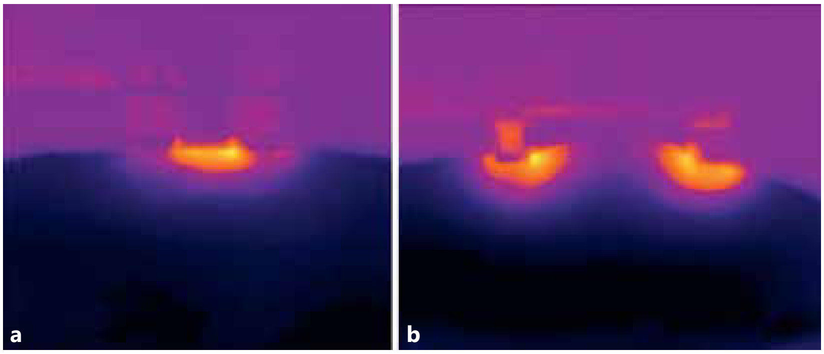

Fig. 9. Thermal images of tissue cross section treated with bipolar device at small (**a**) and large (**b**) distance between electrodes.

In figure 9a, the heat is generated between the electrodes, while the heating profiles directly under the electrodes are less pronounced. This geometry allows generation of uniform heat in a limited volume. This geometry is suitable for homogeneous heating of the skin layer with a depth of up to a few millimeters. The main application of this geometry is subnecrotic skin heating for collagen denaturation and stimulation of remodeling. In figure 9b, the heat is concentrated under the electrode, as occurs in monopolar devices. The temperature distribution is not uniform, and in practice it is evident the heating occurs with hot spots.

The most uniform distribution of RF current is obtained in planar geometry when the area of parallel electrodes is larger than the distance between them. RF current distribution for planar geometry is shown in figure 10.

RF heating between electrodes will be uniform for most of the volume with divergence of current at the periphery of the electrodes. This geometry can be reached by

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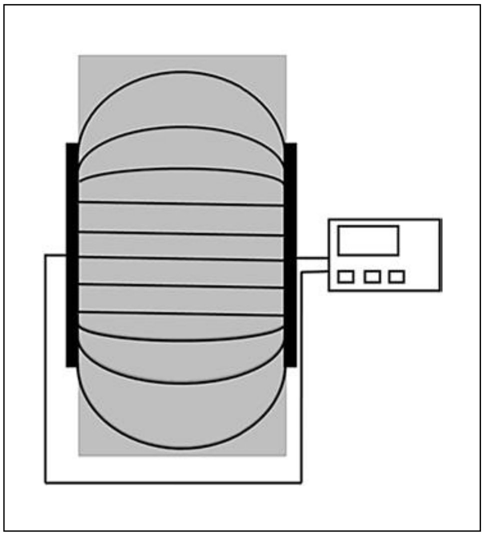

Fig. 10. Electrical current distribution for planar bipolar geometry.

folding tissue between electrodes. This is commonly done in aesthetic medicine by applying negative pressure (in the form of vacuum) to elevate and pinch the skin between two parallel electrodes. This geometry is typically used in body contouring to deliver uniform heating to depth.

Bipolar devices are usually used to create larger thermal zones in nonablative applications. The advantage of bipolar systems is the localization of electrical current in the treatment area.

The response of tissue to bipolar RF can be demonstrated by thermal experiments conducted in in vitro studies using porcine tissue. For the current example, an RF generator with a frequency of 1 MHz and 50-watt power was applied. A thermal camera (FLIR A320) was used for thermography of tissue during RF application. Figure 4, earlier in this chapter, shows the thermal response to monopolar RF where a 1-mm electrode was applied to the tissue surface and a large, 100-cm^2 return electrode was placed at the bottom of the tissue. The heat is concentrated near the surface of the small electrode, and the depth of thermal zone is about half of the electrode size. In contrast, figure 9b shows bipolar geometry where both electrodes have an equal size of 10 mm and the distance between them is 10 mm. The thermal zone is located between electrodes and has uniform distribution down to a depth of 5 mm. For bipolar geometry, where the distance between the electrodes is about electrode size or less, the penetration depth is about half of the distance between electrodes. At an increasing distance between the electrodes, the RF energy distribution becomes nonuniform, and most of the heat is concentrated near the electrode surface (fig. 9b). Folding the skin between two planar electrodes allows uniform heating of large tissue volume (fig. 11). Penetration depth is determined by electrode height and can be as large as a few centimeters.

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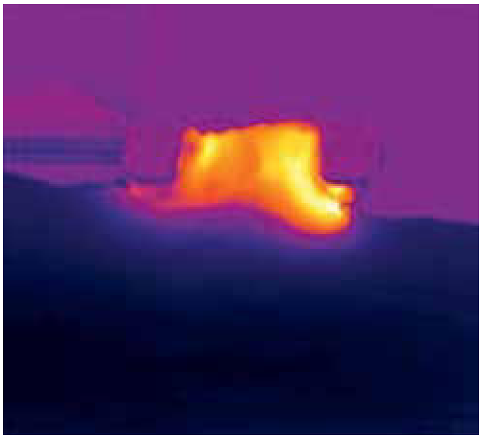

Fig. 11. Thermal image of heat distribution created in the skin folded between two parallel electrodes.

Electrical Properties of Tissue

A specific feature of RF current in biological tissue is ion conductivity. As a result, the electrical effects related to magnetism are negligible, and tissue behavior under RF current is quite well described using Maxwell theory. Considering tissue as a resistant media having some capacitive properties, this has an effect which becomes more significant at higher frequencies. In the RF range of 200 kHz to 1 MHz, the tissue resistivity significantly dominates in tissue behavior, and we can ignore capacitive properties, which are more significant for RF generator development than for medical applications. Therefore, for purposes of this discussion, the terms resistance and impedance will be considered the same.

For tissue with uniform properties, resistance *(R)* is equal to:

$$R = \rho \frac{L}{S}, \tag{9}$$

where ρ is resistivity of tissue, which is equal to resistance of a conductor with an area of 1 m^2 and length of 1 m. S is the cross-section of tissue experiencing RF current and L is the distance between electrodes. This simplified equation allows comprehension of the most basic principles of RF current behavior: tissue impedance is higher for smaller electrodes and a larger distance between them.

Often in literature, the term conductivity is used as the opposite to resistivity. Conductivity of different types of tissue may vary significantly. Electrical properties of some tissues are presented in table 1.

It is critical to understand that in vitro measurements for pure substances can be significantly different from a living patient because on a macro level there is a mix of tissues. For example, according to the table above the difference between wet skin and fat is approximately a factor of 8, while at multiple measurements conducted in vivo

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Table 1. Conductivity of different types of biological tissue at 1 MHz [12]

Tissue	Conductivity, S m^{-1}
Blood	0.7
Bone	0.02
Fat	0.03
Dry skin	0.03
Wet skin	0.25

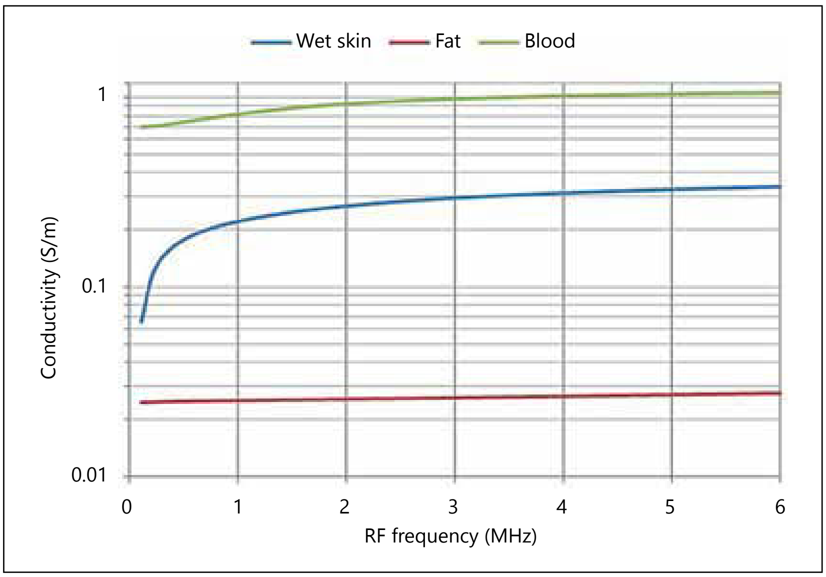

Fig. 12. Tissue conductivity as a function of RF frequency [12].

the difference is approximately a factor of 3. This can be explained by the presence of a vascular network, connective tissue matrix, and intercellular liquids in the adipose layer. It can also explain the significant variance in data reported in different studies [11]. Basically, tissue with higher water and blood content has high electrical conductivity. Tumescent anesthesia may significantly increase tissue conductivity by increasing water and salt content.

Tissue conductivity can be a strong function of RF frequency. Figure 12 shows conductivity of fat and skin calculated according to the parametric model [12]. Skin conductivity is strong function of frequency in the range of 100 KHz to 1 MHz and has a weak change at higher frequencies. Fat conductivity is flat in all the ranges of frequencies used in medicine.

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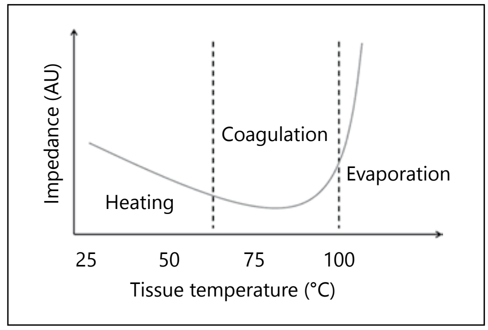

Fig. 13. Tissue conductivity as a function of temperature.

Tissue electrical conductivity is a function of temperature. Qualitative behavior of tissue impedance as the function of temperature is shown in figure 13.

Warming of tissue reduces its impedance with a rate of about 1.5–2% per degree centigrade up to the point of coagulation [11]. This change is related to reduction of tissue viscosity which is reduced with temperature increase. Coagulation of the tissue causes a chemical change in tissue structure, and the trend of impedance behavior is changed. When tissue is heated to 90–100°C, the evaporation of liquids starts, which increases tissue impedance substantially. Further heating of tissue leads to its carbonization. The dependence of tissue conductivity on temperature is utilized by some medical devices. For example, a technology known as electro-optical synergy applies light in particular wavelengths for preferential heating of certain tissue targets; the preheating of the target tissue then creates a preferable path for RF current [13, 14]. This can provide treatment advantages for some applications.

Radiofrequency Thermal Effect on Tissue

The thermal effect of RF on tissue is not different from laser or any other heating method. Multiple studies [15, 16] discuss the temperature effect on tissue. Treatment effect is not a function of temperature only, but also of the length of time when this temperature is applied. Therefore, exposure to a temperature of 70–90°C for milliseconds can cause coagulation, while temperature applied for a few seconds at a lower temperature of 45°C causes irreversible damage.

The typical sequence of tissue response to temperature increase is as follows. 37–44°C: acceleration of metabolism and other natural processes. 44–45°C: conformational changes in proteins, including collagen; hyperthermic cell death. 60–70°C: denaturation of proteins; coagulation of collagen, membranes, hemoglobin; shrinkage of collagen fibers. 90–100°C: formation of extracellular vacuoles; evaporation of liquids. >100°C: thermal ablation; carbonization.

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Pulse Duration Effect

Pulse duration is one of the most critical parameters when utilizing RF energy in order to achieve a clinical response. It affects treatment results because timing influences the thermochemical process in tissue. The other effect of pulse duration is energy dissipation away from the treatment zone due to heat conductivity from the exposed area to the surrounding tissue.

There is extensive data on the correlation between tissue temperature, pulse duration, and treatment effect. Moritz and Henriques [17] demonstrated that the skin thermal damage threshold is a function of temperature and time. Later, it was demonstrated that skin damage function can be described by the Arrhenius equation, where time is a preexponential factor and temperature is an exponential factor [16]:

$$D = A\,t\,\exp\left(\frac{-\Delta E}{R\,T}\right).$$

In other words, the degree of damage *(D)* is a linear function of pulse duration *(t)* and an exponential factor of tissue temperature *(T)*. Practically speaking, then, tissue temperature is more influential on the degree of damage than pulse duration. Nonetheless, prolonged low-grade temperature elevation impacts tissues [27].

It is well known that sustained hyperthermia at 42°C for tens of minutes causes death of most sensitive cells [18]. Once elevated, tissue temperature can only be reduced by dissipation of heat. Temperature dissipation is characterized by the TRT of the targeted tissue. When treatment is intended to heat a structure without heating the peripheral tissue, it must be elevated to that temperature before dissipation begins by heat transfer. Therefore, to localize treatment, the pulse duration should be less than the TRT.

The TRT is a function of tissue thermal properties, as well as the shape and size of the heated volume. Soft tissue has thermal properties close to water.

For a planar object, the TRT can be estimated as [19]:

$$TRT = \frac{d^2}{4\,a},$$

where d is the thickness of the layer and a is tissue diffusivity. Diffusivity is equal to tissue conductivity divided by the heat capacitance and is measured in $cm^2\ s^{-1}$.

For a cylindrical object, such as a blood vessel or hair, a similar equation can be used with different geometrical factors:

$$TRT = \frac{d^2}{16\,a},$$

where d is object diameter. The equation makes evident that cooling time is a square function of the size of the heated target [26].

Lapidoth M, Halachmi S (eds): Radiofrequency in Cosmetic Dermatology.
Aesthet Dermatol. Basel, Karger, 2015, vol 2, pp 1–22 (DOI: 10.1159/000362747)

Radiofrequency Applications

In aesthetic medicine, the RF applications can be divided into three main groups:

- Noninvasive tissue heating with RF, which is used in a range of clinical applications including wrinkle reduction, skin tightening, cellulite, and circumference reduction
- Fractional coagulation and ablation for skin resurfacing
- Minimally invasive treatment for volumetric collagen shrinkage and fat melting.

Noninvasive Radiofrequency

Noninvasive RF treatment is based on the application of RF electrodes externally to the skin of the treatment area. The applied RF energy penetrates into the tissue up to a few millimeters. In order to reach collagenous tissue in the dermis and subcutaneous fat, the RF current must pass through the epidermis. There are some limitations to the amount of RF energy that can be applied noninvasively because the epidermal layer should remain undamaged. The limited heating results in a relatively conservative thermal effect, and usually multiple treatments are required to provide visible improvement. The RF energy can be applied using monopolar [7, 8] electrode geometry or bipolar systems. The RF energy can be delivered in pulsed mode, where a predetermined amount of energy is delivered to each spot, or in CW mode, in which electrodes move over the skin surface continuously for gradual, incremental heating. Typically, the temperature of the tissue should not exceed 40–43°C to avoid epidermal damage. Because skin damage is an exponential function of the temperature, it is challenging to get to the maximal point of the temperature range without the risk of a burn. It is much easier – and safer – to obtain optimal results by extending the treatment time and maintaining a safe temperature longer. The treatment effect is based mostly on collagen remodeling and local metabolism acceleration. Skin tightening, which is often desired in noninvasive treatments, requires heating of the reticular dermis and subdermal structures. The required heating depth for these indications is 3–6 mm, a range that light energy does not reach well; therefore, RF is currently the main tool for these kind of treatments [26]. For the indications of temporary improvement in the appearance of cellulite or circumference reduction, heating must be deeper. Vacuum can be used to assist in folding skin between electrodes and thereby to increase the penetration depth [20, 21].

Fractional Treatment

Fractional skin treatment was introduced in aesthetic medicine about a decade ago and has become one of the most popular modalities for the improvement of skin qual-

Lapidoth M, Halachmi S (eds): Radiofrequency in Cosmetic Dermatology.
Aesthet Dermatol. Basel, Karger, 2015, vol 2, pp 1–22 (DOI: 10.1159/000362747)

ity. This procedure is based on heating or ablation of multiple small foci with a spot size of 100–400 μm. This allows the procedure to be very tolerable and with relatively short downtime.

In contrast to lasers where the thermal effect is limited to the periphery of the ablation crater, RF energy flows through the whole dermis, adding volumetric heating to fractional treatment. This volumetric heating adds a skin-tightening effect. RF fractional technologies can be administered from the surface, using a grid of electrodes, or intradermally, using a grid of microneedles which deliver the RF energy within the dermis. The surface electrodes provide a more superficial effect improving texture and fine lines [19] while longer needles penetrate deeper, providing deeper dermal remodeling [22]. These approaches are described further in other chapters.

Minimally Invasive Radiofrequency Treatment

Minimally invasive RF treatment recently has gained popularity based on the patient's desire to obtain a more dramatic treatment result after a single treatment. Microneedle RF treats the skin in a minimally invasive manner. Dielectric coated needles have become popular in delivering aggressive heating to the reticular dermis without thermal damage to the skin's surface [23]. By heating deep dermal collagen at a higher temperature than could be safely used at the epidermal level, a much stronger collagen contraction effect can be achieved in order to improve deep wrinkles and enhance skin tightening. The combination of deep dermal treatment with superficial fractional treatment has a high potential for complete skin improvement while avoiding skin excision.

By introducing larger needle electrodes into the deep dermis, for example in RF-assisted liposuction, RF can be used to address tightening of the fibroseptal network of the adipose layer with subsequent accommodation of the overlying skin during local fat removal. When energy is applied under the skin, the dermis and epidermis are relatively protected. More aggressive heating up to 60–70°C can be applied during treatment, creating immediate and more pronounced collagen contraction. In some clinical studies [24, 25] up to 42% area skin contraction was achieved after RF-assisted lipolysis.

Safety Features of Radiofrequency Technology

RF treatment is based on a thermal effect created in a treatment zone, and therefore the typical side effects associated with RF energy have thermal character. Most are related to overtreatment and nonuniformity of the thermal effect. Hot spots are an inherent problem of RF technology. Density of RF current is always higher on the

Lapidoth M, Halachmi S (eds): Radiofrequency in Cosmetic Dermatology.
Aesthet Dermatol. Basel, Karger, 2015, vol 2, pp 1–22 (DOI: 10.1159/000362747)

surface of electrode and diverges in the interelectrode space. In addition, high curvature of electrode edges can lead to the concentration of RF energy. Proper design of electrodes can make this problem negligible. In addition, poor contact of RF electrodes with the tissue may cause high RF current density at the points of contact, leading to thermal skin damage.

To minimize the risk of side effects, RF devices incorporate a number of safety features:

- Monitoring of RF energy
- Monitoring of tissue impedance
- Monitoring of skin surface temperature

Monitoring electrical parameters of RF energy is an easy task because the RF electrodes are in contact with the tissue. It is possible to capture data about the tissue temperature, as the temperature alters the impedance. Hence, by monitoring the measured output voltage and current, the device can detect changes in tissue temperature in real time. Most devices will detect and indicate bad coupling between electrodes and the skin and are able to adjust the RF output according to measurements. This real-time monitoring is not possible with laser treatments, as there is no closed-loop feedback mechanism with light-based systems.

Because the highest risk of overheating is in the vicinity of the electrode surface, a basis for safe temperature monitoring is provided by embedding a temperature sensor into the RF electrodes.

There are specific side effects related to RF treatments, which are common for all technologies which utilize heat-mediated modalities. There is a difference between side effects which are classified as expected sequelae, and complications, which are unexpected. Generally, noninvasive devices can cause temporary edema, bruising, arcing injuries due to incomplete skin contact with the electrodes, or focal depressions. The most common complaint from patients treated with these devices is less than expected improvement in their original condition, due to the limitations of noninvasive RF devices. While the same concerns may appear following treatment with minimally invasive devices, both expected sequelae and complications are more common, due to the higher temperatures used to achieve an effect, the necessary access punctures, and the accompanying lipoaspiration, when performed. Postinflammatory hyperpigmentation or hypopigmentation can be seen with either device type when melanocytes are stimulated by inflammation in patients with darker skin types. Many complications can be prevented by correct patient selection and optimization of treatment parameters. Choosing a safe optimal temperature and dividing treatment sessions, rather than using an overaggressive approach, can also help to optimize safety. In general, RF treatments are quite safe and predictable, which has led to the growing popularity of RF in aesthetic medicine.

Lapidoth M, Halachmi S (eds): Radiofrequency in Cosmetic Dermatology.
Aesthet Dermatol. Basel, Karger, 2015, vol 2, pp 1–22 (DOI: 10.1159/000362747)

Side Effects and Treatment Safety

Side Effects

As the main impact of RF is thermal, the major side effect associated with RF treatment in aesthetic medicine is thermal in nature. The overheating of the tissue is usually connected to two main events:

- Overdose of RF energy
- Hot spots created due to the nonuniform application of RF energy.

The skin reaction on overheating appears in different ways:

- Erythema
- Edema
- Blistering
- Full-thickness skin burn
- Charring with eschar

These skin reactions are differentiated by the level of thermal damage applied to the skin. These side effects are sequential in their appearance, and physicians with treatment experience can often prevent higher-level burns through increased attention to the skin reaction.

Erythema and edema are short-term skin reactions, and for many treatments they are the end points that the operator wishes to see. Ideally, after treatment the skin should exhibit uniform redness and slight swelling. Usually, the edema and erythema dissipate after 30 min, but they can linger up to 24 h following treatment.

A superficial burn may develop, at times in the form of a blister, and can also develop a crust. Superficial burns typically resolve within a week. Blistering indicates a greater degree of edema, which leads to separation of the epidermis. Blistering is nearly always preindicated by the development of strong erythema. Skin burns of greater severity may lead to the development of longer-lasting changes, including postinflammatory hyperpigmentation, hypopigmentation, scarring, skin depressions, or textural irregularities.

In minimally invasive treatments, where RF is applied internally, control of the skin reaction is more difficult since the temperature is applied from inside and external visual changes may occur late. In this case, the amount of energy delivered to the tissue should be controlled more carefully.

There are multiple reasons that RF can cause side effects.

(1) Poor technique, especially by an operator who is learning how to use the device, may lead to improper contact or movement of the electrodes. Poor contact can lead to hot spots; therefore, the operator should always ensure firm and proper coupling of the handpiece with the skin surface during RF treatment. In addition, to maximize treatment results, the operator should plan a movement pattern that applies energy uniformly to avoid over- or undertreating.

(2) Improper parameter selection can also lead to the misuse of RF. As with laser and light, RF has no universal set of parameters that work for all patients and all areas.

Lapidoth M, Halachmi S (eds): Radiofrequency in Cosmetic Dermatology.
Aesthet Dermatol. Basel, Karger, 2015, vol 2, pp 1–22 (DOI: 10.1159/000362747)

However, a guiding principal is that thinner tissue should be treated with lower power. In addition, lower maximum temperature is mandated when treating thin skin and soft tissue, such as the neck and face.

(3) Patient sensitivity varies significantly. Some patients are more sensitive to treatment than others, and we cannot always recognize which patients are more sensitive prior to treatment. Applying test pulses and adjusting based on patient preference can assist in determining the ideal setting for a given patient.

General Safety Approach Using Radiofrequency Technology

There are a number of methods to minimize the risk of adverse effects without compromising treatment efficacy. The following are the main methods that are applicable to almost all RF treatments.

(1) Use test spots in less visible areas to determine how the skin will react to treatment.

(2) Begin with lower settings and gradually increase energy to optimal/advanced parameters

(3) Use lower settings on:

(a) Small zones

(b) Bone prominences

(c) Areas with high curvature

(4) Always observe the immediate skin reaction

(5) Stop energy and treatment when there is any indication for concern and reassess continuation of treatment

(6) Do not rush treatment

The use of test pulses is a common technique in laser and RF medicine to test treatment parameters in a less visible area in order to identify optimal settings for the full RF treatment. It is important to observe the skin reaction after each test pulse and adjust parameters if required. Adverse events may not appear immediately; therefore, it may take a few minutes or even a day after pulsing for the full response to be visible. Even for patients treated previously with higher parameters, each new session should start with slightly lower settings, as skin reaction may be different due to seasonal skin dryness or recent exposure to sun.

Parameters should be adjusted according to the treatment area. When treating small zones, the applicator overlaps the same spot more often and the average RF energy applied is higher. In order to compensate for this effect, lower RF power settings are recommended. When treating over bony areas such as the forehead, RF energy application to the thin layer of tissue results in stronger heating. Reduction of RF power improves comfort and provides a greater level of safety for the patient. In addition, it is more difficult to keep electrodes in full contact with the tissue over bony and highly curved areas. Poor contact results in high RF energy density in the areas of

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Aesthet Dermatol. Basel, Karger, 2015, vol 2, pp 1–22 (DOI: 10.1159/000362747)

contact, which generates hot spots and can cause patient discomfort and burns. For such areas, it is always recommended that the operator reduce RF power and use more gel or other coupling liquid. In addition, RF should be stopped when there is a change in the patient position, while pausing to observe skin reaction, while adding more gel, and so on. For safety, it is more important to learn how to stop the device than how to activate it.

As discussed, before an adverse skin reaction appears, there are warning signs that can be a signal when problems are minor. If left unattended to or ignored, these can result in more significant issues. By closely observing the skin as well as safety feedback data from the equipment, an operator can predict the skin's response to treatment and prevent or curtail thermal injury.

In general, all these recommendations can be summarized to one basic preface: the best device is highly dependent on the operator. Nothing is more supreme than one's own educated observation. The manufacturer's treatment recommendations reflect the average treatment pattern, but each patient is unique. It will take time to get comfortable with the technology, so it is important not to rush during the procedure. The time lost with a slower treatment can never be compared with time spent on the treatment of adverse effects and patient dissatisfaction.

Conclusions

RF-assisted medical devices have evolved dramatically within the last two decades. What used to be a simple array of fairly basic tools has now become an extremely sophisticated and sometimes confusing collection of options. There is quite a bit of value in understanding the way RF energy works. The information in this chapter can help a potential buyer of new equipment make a rational choice, based on goals of treatment and physics of the RF device in question. Even more importantly, the physician's understanding of his or her devices can maximize treatment outcomes and can minimize unwanted adverse events and complications.

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Aesthet Dermatol. Basel, Karger, 2015, vol 2, pp 1–22 (DOI: 10.1159/000362747)