The cutting mechanism of the electrosurgical scalpel

This content has been downloaded from IOPscience. Please scroll down to see the full text.
(http://iopscience.iop.org/0022-3727/50/2/025401)

View the table of contents for this issue, or go to the journal homepage for more

Download details:

IP Address: 161.253.116.185
This content was downloaded on 05/12/2016 at 16:56

Please note that terms and conditions apply.
The cutting mechanism of the electrosurgical scalpel

Eda Gjika¹, Mikhail Pekker¹, Alexey Shashurin², Mikhail Shneider³, Taisen Zhuang⁴, Jerome Canady⁵ and Michael Keidar¹

¹ Department of Mechanical and Aerospace Engineering, School of Engineering and Applied Science, George Washington University, Washington, DC 20052, USA
² School of Aeronautics and Astronautics, Purdue University, West Lafayette, IN 47907, USA
³ Department of Mechanical and Aerospace Engineering, Princeton University, Princeton, NJ 08544, USA
⁴ US Medical Innovations (USMI), LLC Corporate Headquarters, Takoma Park, MD 20912, USA
⁵ Jerome Canady Research Institute for Advanced Biological and Technological Sciences, USPI Inc., Takoma Park, MD 20912, USA

E-mail: keidar@gwu.edu

Received 21 June 2016, revised 5 October 2016
Accepted for publication 4 November 2016
Published 5 December 2016

Abstract
Electrosurgical cutting is a well-known technique for creating incisions often used for the removal of benign and malignant tumors. The proposed mathematical model suggests that incisions are created due to the localized heating of the tissue. The model estimates a volume of tissue heating in the order of $2 \cdot 10^{-4}$ mm$^3$. This relatively small predicted volume explains why the heat generated from the very tip of the scalpel is unable to cause extensive damage to the tissue adjacent to the incision site. The scalpel exposes the target region to an RF field in 60 ms pulses until a temperature of around 100 °C is reached. This process leads to desiccation where the tissue is characterized by a significantly low electrical conductivity, which prevents further heating and charring. Subsequently, the incision is created from the mechanical scraping process that follows.

Keywords: electrosurgical cutting, electrosurgery, cutting mechanism, tissue conductivity

(Some figures may appear in colour only in the online journal)

1. Introduction

The scalpel is the most commonly used cutting instrument in the operating room. Historically, the cold scalpel technique has been the preferred method for creating surgical incisions. However, its inability to control blood loss has made electrosurgical probes a necessity. An innovative probe known as the hybrid scalpel has simultaneous capabilities of cutting and coagulating tissue by means of argon plasma with the benefit of reducing blood loss [1–7]. Alternative probes, such as the argon plasma probes, have been used for the treatment of early gastric cancer, superficial esophageal cancer and several types of skin cancers, but these probes are confined to coagulation [7–14]. Despite the widespread application of this technology, the understanding of the underlying mechanism of electrosurgical cutting remains limited.

One of the key parameters of the cutting mechanism is current density, which determines the end result at the treatment site. The cutting mechanism is associated with continuous delivery of currents into the tissue governed by Joule’s law [6–8]. Previous studies have investigated the mechanism of explosive vaporization for creating incisions [15–17]. These studies attribute the rapid tissue temperature increase to the length of the high frequency (HF) pulses delivered within a few microseconds. Furthermore, investigators report that if tissue heating occurs slowly, the cellular liquid evaporates, resulting in tissue desiccation [6, 9, 15–19].
Here, we propose a novel cutting mechanism based on the slow temperature increase approach where, desiccation is achieved prior to tissue slicing. The mechanism of electrosurgical cutting is evaluated using a quantitative approach by combining experimental and theoretical results. In this paper, we (1) outline a physical and mathematical model to predict parameters of tissue heating in the area that comes in direct contact with the electrosurgical probe and (2) validate the model with experimental results.

2. Physical and mathematical model

2.1. The electric field near the tip of the electrode

Figure 1(a) shows a simplified model of an electrosurgical scalpel during tissue cutting. A more detailed explanation of this process and a description of the electrosurgical scalpel are presented in section 3. Figure 1(b) shows the measured current (I) and voltage (U) produced by an electrosurgical system during cutting mode. The total load connected to the electrosurgical system is comprised of two parts connected in series; namely the tissue load and the load of the micro gaps, which may be present between the tissue and the probe. Since (I) and (U) are in phase, the active part of the load’s impedance is dominant, while the load’s reactance is negligible. Note, the absence of a phase shift between (I) and (U) suggests no presence of plasma. If plasma were to be present a phase shift would be observed.

The active part of the load’s impedance is associated with the electrical conductivity of the tissue. The typical electrical conductivity (σ) for various tissues has been reported in the range of 0.01–1 (S m⁻¹) [20–27]. In contrast, the load’s reactance is associated with the micro gaps between the tissue and the probe and/or with the near-electrode sheaths (if the breakdown of these gaps occurs). Therefore, when in cutting mode the load perceived by the electrosurgical system is almost the same as the active resistance of the tissue. In the simulations, the load connected to the electrosurgical system is simply modeled by the active tissue resistance.

Figure 1. (a) A schematic of the tissue slicing process with the tip displayed upward mirroring the simulation graphs presented in section 3. (b) The measured normalized current and voltage produced by the tip of the scalpel during cutting mode when the tip comes in contact with the biological tissue.

Figure 2. Boundary conditions for equation (7). Line 1 corresponds to the boundary of the electrosurgical scalpel, and line 2 corresponds to the remote boundary at infinity.

The tip of the electrosurgical electrode is assumed to be ellipsoid (figure 5(b)). In the simulations, the tip time-varying potential is set as a boundary condition (figure 2). Since the ESU is operated in monopolar cutting mode the second electrode, also known as the patient port, is assumed to be at infinity.

The biological tissue is envisioned as an insulator with a jelly-like texture that contains a mixture of water and ions such as: potassium, sodium, bicarbonate, amino acids etc. The sample is considered to have an electrical conductivity of \( \sigma = 0.01–1 \) (S m⁻¹) [20–27] and a dielectric constant of \( \varepsilon = 3–5 \), representative of the lipid bilayer formulated with DPPC (1,2-dipalmitoyl-sn-glycero-3-phosphocholine). DPPC is one of the most widely used phosphatidyl cholines in membrane models and one of the main components of eukaryotic cell membranes [24]. For calculation purposes it is assumed that \( \sigma = 1 \) (S m⁻¹) and \( \varepsilon = 3 \). The relaxation time of space charge (\( \tau_M \)) for the electrical field/charges to reach a steady state is estimated as follow:

\[
\tau_M = \frac{\varepsilon \varepsilon_0}{\sigma} = 1.44 \cdot 10^{-10} \text{s.}
\]

Here \( \varepsilon_0 = 8.8542 \cdot 10^{-12} \) (F m⁻¹) is the dielectric constant of vacuum.
The calculated \( \tau_M \) value is much smaller than the typical time variation of the potential on the electrode, which is \( \tau_\Phi \approx 2 \) (\( \mu s \)) (see figure 1(b)).

Based on these observations, the biological tissue can be considered quasi-neutral and the current density can be expressed as:

\[
\vec{j} = \sigma \vec{E}.
\]  
(2)

Under the considered condition, current continuity takes the following form:

\[
\text{div}\vec{j} = \Delta \Phi = 0.
\]  
(3)

Equation (3) can be written in the poloidal variable coordinate system (\( \eta \) and \( \mu \)):

\[
\begin{align*}
\eta &= r_0 \cdot \text{sh}(\eta) \cdot \sin(\mu) \\
\mu &= r_0 \cdot \text{ch}(\eta) \cdot \cos(\mu)
\end{align*}
\]
(4)

which leads to the expressions [28]:

\[
\frac{1}{H^2} \left( \frac{1}{\text{sh}(\eta)} \frac{\partial}{\partial \eta} \left( \sinh(\eta) \cdot \frac{\partial \Phi}{\partial \eta} \right) + \frac{1}{\sin(\mu)} \frac{\partial}{\partial \mu} \left( \sin(\mu) \frac{\partial \Phi}{\partial \mu} \right) \right) = 0
\]
(6)

\[
H = r_0 \cdot (\sinh^2(\eta) + \sin^2(\mu))^{\frac{1}{2}}
\]
(7)

\[
r_0 = \sqrt{A^2 - B^2}
\]
(8)

where \( A \) and \( B \) are the semi axis of ellipsoid.

Since the electrode potential is fixed (i.e., independent from the variable \( \mu \)), equation (4) modifies to:

\[
\Delta \Phi = \frac{1}{H^2 \sinh(\eta)} \frac{\partial}{\partial \eta} \left( \sinh(\eta) \cdot \frac{\partial \Phi}{\partial \eta} \right) = 0
\]
(9)

where the boundary conditions (as shown in figure 2) are:

\[
\Phi(\eta_0) = \Phi_0 \\
\Phi(\Phi_1) = 0
\]
(10)

The volumetric heat source associated with the current is obtained from the simulation results for the spatial distribution of the potential (\( \Phi \)):

\[
Q = \frac{1}{2} \sigma E^2.
\]
(11)

As detailed in section 2.3., this volumetric heat affects the amount of current that penetrates the tissue, which leads to a temperature increase.

2.2. Simulation results

Simulations of potential distribution were performed for the following conditions: \( A = 1 \) (mm), \( B = 0.25 \) (mm), \( \Phi_0 = 1 \) (V), \( \sigma = 1 \) (S m\(^{-1}\)). The radius at the tip of the electrode is estimated by \( r_0 = \frac{B^2}{A} = 0.0625 \) (mm) where \( A \) and \( B \) are the semi axis of the ellipsoid tip.

Figures 3 and 4 show the potential, the electric field and the heat distribution at the tip of the white scalpel. The region of radiofrequency (RF) field where the largest source of heat is observed, it is localized at the tip of the scalpel with radius \( r_0 \). This region creates a tissue heating volume of about \( 2 \cdot 10^{-4} \) mm\(^3\).

2.3. The estimation of tissue temperature in contact with the tip of the electrosurgical scalpel

The temperature of the biological sample during the cutting process can be estimated by referring to the simulations results presented above. Since localized heating of the tissue is carried out by heat conduction, the equation takes the following form:

\[
C \cdot \frac{\partial T}{\partial t} = \kappa \Delta T + Q
\]
(12)
where an approximate estimation of the temperature heating range is obtained from the following the thermo-physical constants of water: coefficient of heat capacity $C = 4200 \text{ (J Kg}^{-1} \cdot \text{K}^{-1})$, coefficient of heat conduction $\kappa = 0.5 \text{ (W m}^{-1} \cdot \text{K}^{-1})$, tissue density $\rho = 1000 \text{ (Kg m}^{-3})$.

The typical exposure time ($\delta_{ex}$) of a specific tissue location to the RF-field produced by the electrosurgical scalpel, is dependent on the velocity of the scalpel’s motion ($v$). This velocity is usually in the order of $v \approx 1 \text{ (cm s)}$. Thus, the typical exposure time is estimated as: $\delta_{ex} = \frac{r_0}{v} \approx 60 \text{ (ms)}$. This length heating of pulse is associated with a slow temperature increase and it is much longer than the microsecond range reported for the model of explosive vaporization.

Pulse duration and the size of the electrode also influenced the extent of thermal spread. Based on these observations, the depth of thermal diffusion for any given region during the exposure time reported above can be estimated as

$$L_T = \sqrt{\frac{\kappa}{C \rho} \cdot \delta_{ex}} < 10^{-2} \text{ (mm)}.$$  

This value is significantly smaller in comparison to the size of the region heated by the field of the scalpel’s motion ($\delta_{ex}$). Thus, the heating region is confined in the area where an RF-field exists, while heat diffusion is considered negligible for the estimated time scale of tissue exposure ($\delta_{ex}$).

The temperature increase rate of the tissue ($\dot{T} = 2 \cdot 10^6 [\text{C} S^{-1}]$) can be estimated from equation (12) by neglecting the thermal conductivity term:

$$T = \frac{\langle Q \rangle}{C \rho} = \frac{1}{2} \left( \frac{\sigma (E^2)}{C \rho} \right) = \frac{1}{2r_0} \left( \frac{\sigma (\Phi^2)}{C \rho} \right) = \frac{1}{2r_0} \left( \frac{\sigma V_{\text{RMS}}^2}{C \rho} \right)$$  

where $\langle Q \rangle$ is the heat flux averaged over period of the RF-field oscillation ($\sigma = 0.4 \text{ (S m}^{-1})$ is the electric conductivity of the tissue (according to the measurements presented in the next section), $E^2$ and $\Phi^2$ are the RMS electric field and voltage applied to the electrosurgical probe ($V_{\text{RMS}} = \sqrt{\Phi^2} = 400 \text{ (V)}$ according to the measurements presented in the next section). Based on these estimations and due to RF-field heating, the temperature at the tip of the probe increases to around 100 °C. Consequently the target site, which is in direct contact with the scalpel, experiences localized heating of up to that same temperature. These observations indicate that the size of the heated region is in a similar order of magnitude with $r_0$, the probe tip radius.

It is important to note that according to equation (13) the increase in tissue temperature is proportional to $\sigma V_{\text{RMS}}^2$. Therefore, in order to achieve a particular temperature, a highly conductive tissue would require less voltage than a less conductive tissue.

3. Experimental validation

3.1 Materials and methods

The experiments were carried out with the hybrid electrosurgical system (ESU) SS-200E/Argon2 from US Medical Innovations (figure 5(a)). The ESU can function in cutting and coagulation mode. However, for the purpose of this testing it was operated in pure cutting mode with an argon flow rate of 3 lpm. Energy was transmitted by the ESU in monopolar mode where the electrical current passing through the treatment site, exited the tissue through a dispersive neutral electrode known as the patient port. Five thawed biological samples for cow, chicken and pork were tested. The samples were rinsed twice with abundant deionized water for up to one minute to remove as much of the electrolytes present in the tissue. The minimum amount of power required for creating a smooth and clear incision was recorded before and after each rinse treatment along with sample conductivity.

3.1.1 Power produced by the ESU. The power produced by the ESU system against the known load resistance ($R_L$) was measured by using a set of non-inductive thin film shunt resistors. The resistors were connected directly to the high voltage and the patient port of the ESU. The voltage produced by the ESU was measured with a high voltage probe. The 0.2, 0.5, 1, 2, 3, 5, 10 and 20 kΩ resistors were tested in the power setting of 40, 60 and 80 W. The average power
over the cycle of applied voltage was calculated for each condition using Ohm’s law.

3.1.2. Equivalent tissue resistance. The range of equivalent tissue resistances ($R_t$) was determined using Ohm’s law while using the minimum amount of power needed for creating an incision. The voltage was measured with a high voltage probe, whereas current was determined through a 10 Ω non-inductive thin film shunt resistor connected to the circuit in series.

3.1.3. Electrical conductivity of biological tissue. Tissue conductivity was measured by utilizing the 4-point method shown in figure 5(c). The current in the circuit was determined with a 300 Ω shunt resistor. While the resistance of the conductive tissue was calculated by utilizing the measured voltage between the two most inner electrodes imbedded in the tissue.

3.2. Results

Tissue’s electrical resistance and thus conductivity primarily depends on the degree of vascularization and water content. Once the majority of the electrolytes are removed by rinsing with water, the sample experiences an increase in resistivity, and it becomes less conductive. Figure 6 presents the average measured conductivity for three different types of samples before and after they were rinsed with deionized water. The results revealed a decrease in electrical conductivity based on the number of water rinses with slight variation among samples. The conductivity of the biological tissues was thought to decrease due to the loss of electrolytes associated with tissue vascularization. Our measured sample conductivity of $\sigma_i = 0.17 \sim 0.7$ (S m$^{-1}$), which encompassed the pre and post rinse conditions, correlated well with the range reported in the literature for both human and animal tissue [20–27].

When the ESU system is operating in cutting mode, the equivalent tissue resistances determined from direct measurements were in the range of $R_t = 2$–$10$ (kΩ). Note, the resistances ($R_t$) were also calculated utilizing the tissue conductivity measurements presented in figure 6. Indeed, tissue impedance was modeled as the resistance of the half-space filled with conducting material ($\sigma_i$) between the hemisphere (with radius $r_0$) located at the origin and at infinity: $R_t = 1/2\pi r_0$. The measured tissue conductivity $\sigma_i = 0.17$–0.7 (S m$^{-1}$) and the radius at the tip of the probe $r_0 = 6.25 \cdot 10^{-5}$ (m), assumed to be in the same order of magnitude as the size of the heated tissue region, resulted in a tissue resistance of $R_t = 3$–$15$ (kΩ). This value was in agreement with the directly measured range of $R_t = 2$–$10$ (kΩ).

The actual power output characteristics of the ESU operating in cutting mode as a function of resistance are shown in figure 7(a). It is important to note the difference between the selected power setting and the actual power delivered by the ESU. In fact, the graph indicates that the actual power delivered to the tissue was 35–78% smaller in comparison to the power setting.
Figure 7(b) shows the RMS voltage ($V_{RMS}$) produced by the ESU in the range of possible tissue resistances $R_t = 2–10$ (kΩ). These results showed the ESU system performing comparably to a voltage source with 20% accuracy among different voltage readouts. As the power setting increased from 40–80 W, the RMS voltage ($V_{RMS}$) followed the same pattern by steadily increasing from 400–530 V (figure 7(b)). These observations are also inferred by figure 8, where again the RMS voltage increases based on the ESU power setting. Figures 7–8 confirm the ESU system behaving as a voltage source.

The experimental results of the tissue cutting process are discussed below. The electrosurgical unit was able to produce a scar 0.5 mm deep with a lateral spread of about 1–2 mm. The minimum amount of power needed to produce an incision was defined as the instance when 1–2 cutting strokes resulted in a smooth and visible separation of tissue. Figures 9(a) and (b) show incisions created before and after the water rinsing. As anticipated, the appearance of the sample changed after it was rinsed with deionized water. The change in appearance was due to tissue dehydration.

Figure 10 shows the average minimum value of the power setting required for creating an incision. The power setting increased with the number of water rinses and varied slightly among samples. The minimum amount of power required for producing an incision in the untreated thawed samples was 30–40 W. This range increased to 55–75 W after two consecutive water rinses, due to an increase in tissue resistance. The results presented in figure 10 confirm the findings of figures 6 and 8. Furthermore, they verified that the minimum RMS voltage required for creating an incision increased as the samples experienced a drop in electrical conductivity.

4. Discussions

In this section we proposed a cutting mechanism based on the measurements presented above. The presented simulations indicate that any given tissue location is exposed to the RF-field produced by tip of the scalpel via 60 ms pulses while the scalpel is moved by the operator with a typical velocity of about $v_s \sim 1$ (cm s$^{-1}$). This exposure time was relatively long in comparison to microsecond pulses reported from other studies. An exposure time in the millisecond range is indicative of a slow tissue temperature increase [15–17].

The presented experimental results revealed that the RMS voltages produced from the scalpel are in the 400–600 V range (see figure 8). The proposed theoretical model indicated that when this range of RMS voltage is applied in 60 ms pulses, it leads to localized heating of the tissue to around 100 °C. Thus, it can be concluded that the RF heat produced by the electrosurgical scalpel is sufficient to reach boiling temperature. The boiling of intracellular liquid is associated with the creation of bubbles. These bubbles cause the cell membrane to rupture, release, and vaporization of the cellular liquid resulting in tissue dehydration and desiccation. The desiccated tissue becomes weakly conductive and structurally less solid than the untreated tissue. Therefore, the simultaneous mechanical scraping can create separation of tissue along the blade of the scalpel by resulting in what is defined as cut. The tissue walls surrounding the incision site are in a coagulated state.

It is important to note that tissue desiccation prevents further RF heating, which inhibits extensive thermal damage. Thus, the conversion of electrically conductive tissue ($\sigma_t = 0.17–0.7$ (S m$^{-1}$)) into a virtual electrical insulator explains how electrosurgical cutting prevents extensive tissue damage and carbonization.
The proposed model also explains why samples with lower conductivity require an increase in power setting for tissue slicing to occur (figure 10). Based on the theoretical predictions (equation (13)), the localized heating of the biological sample is proportional to $V_{\text{RMS}}^2 \sigma$. Therefore, in order to achieve the desired heating effect, tissues with lower electrical conductivity (figure 8) require a larger amount of applied RMS voltage and thus a higher ESU power setting (figures 6 and 10).

5. Concluding remarks

The cutting mechanism of a hybrid electrosurgical unit with capabilities to coagulate and create incisions was studied. The presented model suggested that cutting occurs as a result of the localized heating of the tissue by Joule’s mechanism. The experimental data showed how the amount of power required for creating an incision was dependent on the electrical conductivity of the biological sample. The decrease in sample conductivity was associated with the change in the sample’s water content due to the removal of electrolytes and the slow increase in temperature. The heating region is confined in the area where an electrical field exists, while heat diffusion is negligible leading to minimum tissue damage.

Acknowledgments

This research was supported by US Medical Innovations LLC. The authors would like to thank Anna Tang, Daniel Shea and George Teel for their assistance in editing the manuscript.

References

[18] Zenker M 2008 Argon plasma coagulation. GMS Krankenhyyg. Interdisz. 3 Doc.15