Heat Transfer Assessment for Tumor Focal Therapy with Rod-Type Probe

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Abstract

Cancer tumor treatment using thermal therapy is discussed from an engineering design perspective. Although there are many ways to treat cancer, focal therapy of tumors using an energy source is a promising and extensively studied technology. Bladder, prostate and melanoma cancer types were discussed in details for their clinical needs and current thermal approaches. A probe using Radiofrequency (RF) heat source for prostate cancer was analyzed for its temperature distribution, diffusion time and length. The tumor of 1cm was modeled as a sphere and the probe tip generating constant heat source.

The purpose of this article was to assess a design capability of a probe (0.5cm long by 1mm diameter) in thermal ablation a 1cm tumor in a 4cm prostate gland. As a result, the critical warming temperature was 52.2°C, where a temperature of 50°C for 5 seconds was achieved for the thermal ablation based on the selected thermal properties. In addition, the diffusion length was up to halfway to approximately 0.5cm.

In conclusion, an improved probe design is suggested to provide a significant improvement in the thermal ablation of the whole tumor as opposed to parts of it.
1 Introduction

Cancer is one of the deadliest diseases worldwide with an estimate of 19.3 million new cancer cases per year projected for 2025 [30]. The current treatments for cancer includes surgery, radiotherapy, chemotherapy, immunotherapy, targeted gene therapy and hormonal therapy [30]. Thermal Therapy have not been widely used, however, it is a promising therapy that can be used as a primary or adjuvant to chemotherapy and radiotherapy [29]. Bredlau et al. describes two approaches to thermal therapy that are hyperthermia (42-45C) and thermal ablation (>50C). Hyperthermia tend to have an immunogenic effects and enhance drug delivery which is suitable to be used as a adjuvant therapy. The thermal ablation is used more for focal tumor destruction by either using radiofrequency (RF), microwaves, laser or high-intensity focused ultrasound (HIFU) [29].

Focal therapy are mainly used in (but not limited to) brain and prostate tumors by inserting a probe directly into the tumor and heating the cancerous cells. The most common type of thermal energy is RF because of its adaptability for many focal tumors by modifying RF electrode design, small size, portable and inexpensiveness [31]. In order to achieve an effective focal therapy with minimum to no damage to the normal tissue, the design of the probe would be critical. Uniformly distributing of the RF energy by using for instance multi-pronged probes for a high impact to the tumor but minimal injury to the normal cells [31]. However, this newer design of probe would require an involved mathematical analysis and modeling. On this article, we will be using the simple rod-type probe design to analyze the heat transfer from the probe to the tumor.

1.1 Bladder Carcinoma

Bladder cancer is the 5th most common cancer with an estimated 80,470 new cases in 2019 [21], which is a cancer that develops in the bladder linings. Smoking and chemical exposures of some kind are the two leading risk factors for bladder cancer [21]. There are three types of bladder cancers with higher percentage of Urothelial carcinoma or Transitional Cell Carcinoma followed by squamous cell carcinoma and adenocarcinoma [22]. Focusing on Urothelial carcinoma, the bladder cancer starts with the cells that lines the urinary bladder. The hypothesis behind the bladder cancer development is when the urothelial cells have some type of genetic mutation such as deletions of chromosom 9, mutation of several genes such as FGFR3, RB1, HRAS, TP53 and TSCI [25]. Once the cancer develops in the bladder lining, it progresses into the muscle tissue of the bladder. There are different disease states (stages) of bladder cancer as shown in Figure 1 below.
Figure 1: Bladder Cancer Stages.

The common types of bladder cancer therapies are radiation, chemo, immuno and targeted (gene) as well as Transurethral Resection (TUR) and Cystectomy described in detail on [27]. Figure 2 below shows the clinical outcome of the major treatment modalities and the superiority of TUR alone. Although the most advanced stages of the bladder cancer needs a combination of treatments, for average cases, removal of the cancer tumor and treatment of the remaining cancer cells is the standard of care.
Compared to other types of cancers, thermal therapy in bladder cancer is not a prevalent current use but there are numerous clinical trials undergoing [26]. Longo et all. showed these clinical trials offers an improved efficacy at tumor eradication and reduced recurrences. The heating mechanisms used in bladder cancer are discussed in [24] specifically for MIBC are intravesical microwave antenna heating, capacitively coupled RF current heating, radiofrequency (RF) phased array deep regional heating of the pelvis. The use of gold nanoparticles (GNPs) with laser or iron oxide nanoparticles (IONPs) with radiofrequency is a promising future technologies in exploratory and clinical trails stages.

1.2 Prostatic Carcinoma

Prostate cancer is a very common form of cancer among men in the United States with around 160,000 diagnoses in 2017 [2]. This makes it the most common form of cancer among men other than skin cancers and the third leading cause of death [2]. Treating prostate is not always the most practical, or even helpful, course of action for patients with the disease. It has previously been shown that the 10-year progression-free survival rate can be as high as 53.1 percent [3] for early stage prostate cancer that is left untreated. According to this same study, in this 10 year window, only 8.5 percent of patients died from prostate cancer [3]. This leaves an important question for physicians and patients alike. Should we treat all prostate cancer, or only the most severe forms? Due to side effects from the most common treatment options of radiation and surgery, intervention is reserved largely for those with the most severe forms. Palliative care is a common course of action for most men with less severe forms of the disease [3].

Expectant management, surgery, and radiation are the three treatment options currently available to men with prostate cancer [2]. Expectant management includes palliative care
and/or active surveillance, which includes regular examinations [2]. Active surveillance remains the standard of care as 5-year survival rates between patients under active monitoring and those that have received surgery or radiation do not differ significantly [2]. Benefits from radiation and surgery are typically confined to those individuals with a more advanced disease state. One study found that the 5-year mortality rates after surgery improved form 12.8 percent to 5.6 percent [2]. Surgery can wither be done with a small incision to remove the prostate (radical prostectomy) or with robot guidance. However, the risks of incontinence and erectile dysfunction make this an unappealing option for many men [4]. Likewise, radiation, whether it is external or internal (brachytherapy) has similar side effects that limit its use for most men with managed symptoms of the disease [4]. Less used therapies include hormone therapies and chemotherapy. These are typically reserved for metastatic cases [4]. Once again, it is frequently the case that the side effects outweigh the benefits of the treatment. Due to past complications using cryotherapy, thermal treatment options are limited to patients that have exhausted all other options [5]. This goes to show that there is a need for a minimally invasive focal therapy technique to treat prostate cancer that avoids some of the side effects that can be an even larger burden than the disease itself.

Currently, there is interest and ongoing development of thermal therapies for prostate cancer. The first we will consider is high intensity focused ultrasound (HIFU). A HIFU transducer is inserted rectally near the vicinity of the prostate. The entire prostate can be treated by scanning the focal point throughout the organ. A 2006 study showed that 86 percent of treatments resulted in negative biopsies, which is comparable to other treatment measures [6]. It also had the benefit of reduced side effects like incontinence. However, as discussed in class, it is a very expensive technology, which makes it a less-than-ideal therapeutic option. Lasers are also being developed for focal thermal treatment of prostate cancer. The laser is brought in proximity to the prostate transrectally or through the perineum. MRI compatibility allows the laser to have a high degree of accuracy (1.1 ± 0.7 mm) in ablating cancerous tissue without damaging the surrounding tissue [7]. Combining lasers with gold nanoparticles resulted in tumor resolution in 93 percent of mice compared with the laser alone being uneffectual [8]. The addition of gold nanoparticles generates more heating of the tissue, resulting in more tissue death. Uneven heating has often doomed attempts to use microwave heating of tumor tissues. The ability to rapidly heat tissue deep in the body using microwaves is still an attractive therapeutic option that may be more realistic as technologies improve for its implementation [9]. Thermal warming using radiofrequency and iron oxide nanoparticles is another treatment option under investigation. The feasibility of this technique has been shown at field strengths well under those that can be tolerated by humans. This means that more heating could be applied for further ablation [10]. Localization and uniformity of nanoparticle dispersion must be optimized for this treatment to be used clinically. It also does not have the capability of being MRI compatible, which would be a huge advantage for spatial resolution. Radiofrequency alone has also proven effective as a fast, minimally invasive technique that is well tolerated. Removing the use of nanoparticles makes this a quicker and less complicated procedure. Studies have also shown that the side effects are relatively mild [11]. Finally, as discussed previously, cryosurgery is already in use clinically as a last resort for individuals that have not had success with other therapies. Though a minimally invasive option to surgery, cryotherapy has many of the same side effects, such as erectile dysfunction and incontinence.
1.3 Melanoma

Melanoma, a malignant tumor associated with skin cancer, is now considered a major public health issue in particular, for those countries whose population is light skinned and sun radiation is abundant. Many studies suggest that the cause of incidence for melanoma is a balance between the individual predisposition and environmental factors. The incidence rate for melanoma has been increasing in developing countries. Arguably, this might be due to a higher sensitivity in classification, together with better diagnoses and earlier detection. In any case, incidence rate increase has been reported to be 3% to 7% for light skinned populations. Estimates in darker skinned countries is not consistent, and thus incidence rates are difficult to determine. [12] As incidence rates increase in developing countries, awareness is raised. Melanoma is now becoming the fourth most common cancer in Australia and New Zealand, the seventh most common in the USA and Canada, the tenth most common in Scandinavia and eighteenth most common in England, Wales and Scotland. In 2000, the cumulative lifetime risk of developing melanoma in the USA was estimated to be around 1:75. [13]

Just as it is the case for incidence rates, melanoma related mortality rates have increased as well, in particular for developing countries with light skinned populations. Again, one should be careful when interpreting these data. Arguably, technological advances and post mortem diagnoses may be increasing the population for with melanoma is considered the cause of death, as opposed to previous decades where this was not the case. The annual mortality rates for countries in the northern hemisphere is estimated to be between 1 to 3 peoples for every 100,000 deaths [14]. Although the mortality rate for the overall population is steadily increasing, studies aimed at younger subgroups see a stable, and in some cases decreasing, rate of mortality. Such is the case in the USA, Scotland, and Australasa. [14]

In the past decades incidence rates have been shown to be particular to different body areas. On average, an steady rate of incidence has been shown for the neck. However, higher incidence rates are common in the abdominal area, in particular, among males. In the USA, and other developing countries in Europe, abdominal incidence of melanoma has been increasing among females [15]. Among other trends, thickness of melanoma masses has steadily decreased in all populations (\(1.5\)mm in thickness). This is attributed to removal of tumors that may have never have progressed beyond early stages of development [14].

The major individual-dependent factor associated with melanoma is skin tone. There is an inverse relation between increasing risk of incidence and increasing natural skin tone [16]. Also related, is the presence of many naevi (birthmarks or moles). At the same time the number of naevi has been shown to correlate with exposure to sunlight, particularly in childhood [17]. A final major individual-dependant factor is age. Melanoma is rare in children, and incidence begins to increase around the age of 20 and rapidly increases until the age of 40 [14]. The evidence suggests that exposure to sunlight in childhood correlates well with incidence and risk factors in adulthood [14].

The major environmental risk factor for melanoma is sunlight. However, the exposures necessary to induce melanoma is a function of total accumulated exposure, rather than the number of intermittent incidences. Noteworthy is the fact that the data is only relevant to the superficial spreading or translocation of masses. Moreover, for Hutchinson’s Melanotic Freckle, the relation is reversed [18]. Other environmental factors have been investigated in-
including diet, smoking, hair dyes, fluorescent lightning, hormone therapy and stress. However, none of these was clearly associated with risk of melanoma (circa 1995) [19].

2 Focal therapy of Prostate Cancer

Here, we are focusing on radiofrequency (RF) ablation of tumors in the prostate. This method has previously been shown to be effective in the targeting of tumors within the prostate [11]. It is advantageous in that it is fast, relatively inexpensive to develop, does not require administration of nanoparticles, and it can be guided using several imaging modalities [31]. This makes it an ideal prospect for the treatment of prostate cancer because it could be performed in a quick procedure and is well tolerated by patients [11]. With many men choosing to monitor their disease progression rather than undergo surgery, an effective, minimally invasive focal therapy option could mean that more men get treatment.

The governing equation (Equation 1) for this model system can be described using a modified form of the Pennes bioheat equation that accounts for the electrical heat generation from the probe tip [20].

\[ \nabla (k \nabla T) + \sigma | \nabla V |^2 - \rho_b C_b \omega (T - T_{amb}) + Q = \rho_c \frac{dT}{dt} \tag{1} \]

The heat generation of the probe is accounted for in this equation based on the \( \sigma | \nabla V |^2 \) term, where sigma is the electrical conductivity of the tissue. This is on the order of 0.148 S/m [20], and scales the gradient of the electrical potential (V). Here, the perfusion term is required to account for cooling that the circulating blood contributes to the system. Finally, there is a small heat generation term (Q) due to metabolism.

Based on this equation, the heating capabilities of the system will attenuate exponentially as you move further away from the probe. This can be seen in Equation 1 form the \( \sigma | \nabla V |^2 \) term. The heat generation from the probe depends on the square of the electrical potential of the tissue. Thus, heating will occur with diminishing effect as one moves further from the probe. This heating also depends on the electrical conductivity of the tissue. A higher conductivity would result in more disperse heating within the tissue according to Equation 1. Heating is also influenced by the thermal properties of the tissue. Particularly, tissues with higher thermal conductivities would be able to warm more evenly. Blood perfusion also plays an important role in the attenuation of the effects of focal warming, especially in the case of tumors. The unorganized vasculature found in tumors may assist in heating by decreasing the perfusion of cooler blood into the target tissue. The reduced blood flow would also help by decreasing the specific heat of the system as a whole, as blood has a higher specific heat capacity than the tissue. Therefore, a lower specific heat would cause the tissue to see a larger increase in temperature for a given amount of energy put into the system.

For the case of Joule heating as we are looking at here, injury to the tissue can be described by the Arrhenius model that was discussed in class and given in Equation 2, which
has a solution of the form found in Equation 3

\[ \Omega = \int_0^\tau A e^{-\delta E/RT} \, dt \]  
(2)

\[ \ln(\tau) = \Delta E/RT - \ln(A) \]  
(3)

\[ T_{\text{crit}} = \frac{E}{R \ln A} \]  
(4)

\[ \ln(k) = \ln(A) - \frac{\Delta E}{RT} \]  
(5)

\[ S = e^{-kt} \]  
(6)

Equation 4 can be obtained by solving the Arrhenius equation for the rate of damage \( \frac{d\Omega}{dt} \). When set equal to one, you can find the critical warming temperature beyond which the rate of warming proceeds very rapidly. Using values given in class for \( E \) and \( A \) (2.46x10^5 J/mol and 3.3x10^39 s^-1 respectively [32]), we obtain a critical warming temperature of 52.2°C. Our focal therapy should therefore be below this temperature for controlled tissue ablation.

Applying Equation 5, we can find a rate constant for thermal injury at a temperature of 50°C, which will serve as the target temperature for ablation in this case as it will provide fast warming while still being well-controlled. Figure (3) shows the plot for fractional survival rate over time under realistic tissue temperatures given a critical temperature of 52.2°C using the results of Equation 6. For the values used here, the fractional cell survival at these temperatures quickly goes to zero. For temperatures over 45°C, zero percent survival could be achieved in under a minute based on these findings. However, based on class discussions, this may be a little extreme. In class, we discussed that around 42°C is where you begin to see substantial cell survival even under prolonged hyperthermia. Our model seems to overestimate cell death. Also, our time estimates seem to predict zero percent cell survival to be achieved ten times faster than what was discussed in class at these temperatures. This is likely due to differences in the values for \( A \) and \( E \). Although cell death can occur quickly at temperatures around 50°C, the nearly five second time period is probably unrealistic.

Figure 3: Cell survival fraction at different temperatures over time.
3 Mathematical model

In \( \mathbb{R}^3 \), using standard Euclidean coordinates, we have a heat conduction equation of the form.

\[
\rho c \frac{\partial T}{\partial t} = \frac{\partial}{\partial x} \left( k \frac{\partial T}{\partial x} \right) + \frac{\partial}{\partial y} \left( k \frac{\partial T}{\partial y} \right) + \frac{\partial}{\partial z} \left( k \frac{\partial T}{\partial z} \right) + f_{gen} \tag{7}
\]

where \( T \) is temperature, \( \rho \) is our tissue density, \( k \) is the heat conductivity, \( c \) is the specific heat capacity, \( f_{gen} \) is a heat source and \( x, y, z, t \) are the usual coordinate and time variables. In this report, we will make the assumption that a tumor has a spherical geometry thus, we introduce spherical coordinates as shown in Figure 4.

![Figure 4: Spherical coordinate system.](image)

Applying the usual transformation

\[
x = r \sin(\theta) \cos(\varphi), \\
y = r \sin(\theta) \sin(\varphi), \\
z = r \cos(\theta),
\]

we get an equivalent description of heat dissipation as given by equation (7); namely:

\[
\rho c \frac{\partial T}{\partial t} = \frac{1}{r^2} \frac{\partial}{\partial r} \left( r^2 \cdot k \frac{\partial T}{\partial r} \right) + \frac{1}{r^2 \cdot \sin^2(\theta)} \frac{\partial}{\partial \varphi} \left( k \frac{\partial T}{\partial \varphi} \right) + \frac{1}{r^2 \cdot \sin(\theta)} \frac{\partial}{\partial \theta} \left( \sin(\theta) \cdot k \frac{\partial T}{\partial \theta} \right) + f_{gen} \tag{8}
\]

3.1 One-dimensional, steady state analysis

For simplicity we begin by assuming that heat generation is only radially dependent. This is justified if the probe to model emits radiation depending only on the distance from the source. Then to further exploit analytical result we will assume that temperature is uniform in the direction of \( \theta \) and \( \varphi \). In other words, we let

\[
\frac{\partial T}{\partial \theta} = 0, \text{ and } \frac{\partial T}{\partial \varphi} = 0.
\]

Lastly, we impose the steady state assumption and arrive at the following equation
\[ 0 = \frac{1}{r^2} \frac{\partial}{\partial r} \left( r^2 \cdot k \frac{\partial T}{\partial r} \right) + f_{\text{gen}}. \] (9)

Note that this equation is consistent with equation 1. Here \( f_{\text{gen}} \) takes the form

\[ f_{\text{gen}} = \sigma |\nabla V|^2 - \rho_b c_b \omega (T - T_{\text{amb}}) + Q. \] (10)

For further analysis we will assume that the term \( \sigma |\nabla V|^2 \) is of exponential form with a half-life decay every .5 mm. We base our choice in the results from [20]. Explicitly:

\[ \sigma |\nabla V|^2 = 75000 \exp(-1400 (r - R_{\text{inn}})) \]

Moreover we assume that radiation is emitted from a spherical domain at the tip of the probe (c.f Figure 5). The boundary conditions then, are fixed by the properties of the probe. We fix a temperature at the surface \( T_{\text{inn}} \) and we impose zero flux boundary conditions at the surface of the tissue \( (r = R_{\text{out}}) \); namely:

\[ T(R_{\text{inn}}) = T_{\text{inn}}, \quad \text{and} \quad \frac{\partial}{\partial r} T \bigg|_{r=R_{\text{out}}} = 0. \] (11)

A list of the parameters used is added on Table 1. We begin by inspecting the behavior of the steady state solution for varying temperature profiles of the probe. That is, by varying the boundary condition \( T_{\text{inn}} \). In Figure 6 we show the steady state solution for a probe whose surface attains steady state at 80°C.
Figure 6: Steady state solution for $T_{inn} = 80^\circ$C.

<table>
<thead>
<tr>
<th>Label</th>
<th>Description</th>
<th>Value</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>$k$</td>
<td>Thermal conductivity</td>
<td>0.51 W/m°C</td>
<td>[28]</td>
</tr>
<tr>
<td>$c$</td>
<td>Heat capacity</td>
<td>3760 J/kg°C</td>
<td>[28]</td>
</tr>
<tr>
<td>$\omega$</td>
<td>Perfusion coefficient</td>
<td>$6.4 \times 10^{-3}$ s$^{-1}$</td>
<td>[20]</td>
</tr>
<tr>
<td>$\rho_b$</td>
<td>Tissue density</td>
<td>1045 kg/m$^3$</td>
<td>[28]</td>
</tr>
<tr>
<td>$T_{amb}$</td>
<td>Surrounding Temperature</td>
<td>37°C</td>
<td>[28]</td>
</tr>
<tr>
<td>$T_{inn}$</td>
<td>Temperature at the probe surface</td>
<td>40-100°C</td>
<td>[28]</td>
</tr>
<tr>
<td>$R_{inn}$</td>
<td>Probe thickness</td>
<td>0.001 m</td>
<td>[20]</td>
</tr>
</tbody>
</table>

Table 1: Parameters used in the model

It is reasonable to assume that tissue damage occurs near 50°C [33]. In this manuscript we won’t go into detail about the amount of damage done to the tissue. However, we will use 50°C as a threshold to inspect desirable properties of the probe. In particular, we inspect how the heating at the tip of the probe affects the damage in the tissue. To begin, even though equation 10 can be solved explicitly for arbitrary boundary conditions using theory of elementary differential equations, the solution isn’t very useful to understand. For
exposition, the solution of 10 with $T[R_{inn}] = a$ is:

$$
T(r) = \frac{(0.0000370278a - 0.001363)e^{222.053r}}{r} + \frac{(0.00119091a - 0.0438376)e^{-222.053r}}{r} + e^{-1400.9r} \left( -\frac{0.000457383}{r} - 0.312114 \right) + 37. 
$$

(12)

An array of plots for different choices of initial conditions is shown in Figure 7 together with their respective heat map.

Figure 7: (a) Steady state solutions for different values of $T_{inn}$. (b) Representative heat map for steady state profiles (color label in units of °C).

Since we are interested in the extend to which the probe is effective we are looking for values of $r$ such that $T(r) = 50$. Figure 8 shows the relation of the extend of tissue damaged as a relation of the temperature at the boundary of the probe. Numerical values are shown in Table 2.

Figure 8: Relation between probe temperature and extend of damaged tissue
<table>
<thead>
<tr>
<th>Probe Temperature (°C)</th>
<th>Extend of damage</th>
</tr>
</thead>
<tbody>
<tr>
<td>68.54</td>
<td>2mm</td>
</tr>
<tr>
<td>126.48</td>
<td>4mm</td>
</tr>
<tr>
<td>208.40</td>
<td>6mm</td>
</tr>
<tr>
<td>284.23</td>
<td>8mm</td>
</tr>
<tr>
<td>313.18</td>
<td>1cm</td>
</tr>
</tbody>
</table>

Table 2: Numerical values for tissue damaged

We conclude this section by making some remarks of the model. First, a weakness to address is the correct behavior of the SAR term. For simplicity, we opted to decouple SAR from the temperature. This is supported by evidence in [20]. However, our curve fitting is not ideal. Chang et al [20] propose a solution in the form of a wave equation dependent on particulars of the rod behavior. Future versions of this model may take this into account as a coupled systems of ordinary differential equations. Moreover we remark that metabolic rates were ignored, inspecting the impact of such might be of interest. All being said, the model does predict thresholds for which a probe might be deemed unsafe. We elaborate on the later in our concluding section.

4 3D Model

We used Comsol Multiphysics to create a 3D model of our system. It consists of a thin copper probe inserted into a tumor within the prostate. This model is based on concepts of RF tumor ablation outlined in Goldberg’s text [31] and the modeling approach was based off of previous work by Chang [20]. We used Equation 1 as the governing equation for our model with some slight modifications. We chose not to model the heat generation of the tissue as it is negligible compared to the heat generated by the probe. We also assumed that the perfusion properties were the same in the tumor as well as the healthy tissue, though this is not likely to be the case in vivo. We assumed that the blood was 37°C and had the same properties used by Chang in a similar model [20]. Therefore the properties of the blood were as follows: density ($\rho = 1000kg/m^3$) [20], specific heat ($c_{blood} = 4180J/kg - K$) [20], and perfusion coefficient ($\omega = 6.4x10^{-3}s^{-1}$) [20].

For the modeling of the probe, only the bare copper tip was included. The wire used to insert and power the probe was assumed to be thermally insulated so that it did not contribute to the heating of the surrounding tissue. The size of the probe was similar to typical RF probes described in the literature [20,31]. It is 1mm in diameter and 0.5cm tall. It is made of copper, which has the following thermal and electrical properties: electrical conductivity ($k_e = 5.998x10^7S/m$), specific heat ($c_{copper} = 385J/kg - K$), relative permittivity ($\epsilon = 1$), density ($\rho = 8960kg/m^3$), and thermal conductivity ($k_{th} = 400W/m - K$). The tumor and the surrounding prostate were both given the same thermal and electrical properties. They are as follows: electrical conductivity ($k_e = 0.148S/m$) [20], specific heat ($c_{tissue} = 3600J/kg - K$) [20], relative permittivity ($\epsilon = 10000$) [34], density ($\rho = 1060kg/m^3$) [20], and thermal conductivity ($k_{th} = 0.5020W/m - K$) [20]. The tu
The tumor was modeled as a sphere with a diameter of 1 cm. The prostate was also modeled as a sphere with a diameter of 4 cm. Figure 9 shows the control volume that was used for this analysis in more detail.

In order to achieve temperatures near the critical temperature of 52.2°C that was determined previously, we chose a probe voltage of 12.5 V with the exterior surface of the prostate sphere serving as the ground. Figure 10 shows the electrical potential through the center cross-section of the model. We did not account for the probe frequency in our model, instead treating it like an electrostatics problem.

Figure 9: Control volume for 3D model consisting of a cylindrical probe and a small tumor within a larger prostate.
Figure 10: Electrical potential through a cross-section of the model geometry.

Figure 11: Temperature distribution through a cross-section of the model at 200 seconds.
Figure 11 shows the steady state temperature distribution through the tumor and surrounding tissue. Figure 12 shows that, at a voltage of 12.5V, the probe temperature peaks at just over 52°C. This fits nicely with the value for the critical temperature that was previously calculated. Based on class discussions, temperatures above 45°C should be sufficiently large to see near total cell death after 10-15 minutes of exposure. However, as can be seen in Figures 11 and 12, the temperature at the boundary of the tumor is not heated enough to cause cell death in a timely manner, if at all. To combat this, multi-pronged probes have previously been suggested and developed [31]. Having 4-6 prongs spread within the tumor allows simultaneous heating of different regions, resulting in much faster ablation of the entire tumor. A multi-pronged probe would therefore allow operation times to be held at a minimum while also achieving better tumor ablation results.

\[
\frac{1}{r} \frac{d}{dr} \left( r \frac{dT}{dt} \right) = 0 \tag{13}
\]

\[
T_1 - \Delta T \left[ \frac{1 - (r_1/r)}{1 - (r_1/r_2)} \right] \tag{14}
\]
We benchmarked our 3D analysis using a 1-D steady state spherical solution (Equation 13). Figure 13 plots the temperature distribution through a sphere as described by Equation 14. There was good agreement between the 3D and 1D spherical solutions. This indicates that the 3D model is appropriately modeling the temperature distribution through the tissue. We also compared the results of the 3D analysis to the 1D analytical approach. The profile of the temperature distribution was similar for both cases. Tissue temperature quickly decreased as you move further from the probe. In both cases, only the tissue that is a few millimeters away from the probe reaches temperatures sufficient to cause cell death. The agreement between these two approaches is further validation of the model.
5 Conclusions

Bladder, prostate and melanoma cancer types were discussed in details for their clinical needs and current thermal approaches. Overall the main clinical need is effectively ablating the tumor with low toxicity and damage to the surrounding normal cells/tissues.

A rod-type probe using Radiofrequency (RF) heat source for prostate cancer was analyzed for its temperature distribution, diffusion time and length. The tumor of 1cm was modeled as a sphere and the probe tip generating constant heat source. As a result, the critical warming temperature was $52.2^\circ C$. Cell survival models predicted that nearly complete cell death could be achieved in a matter of seconds at temperatures near the critical temperature. Though temperatures above $45^\circ C$ are known to cause cell death, this model seems to overpredict to what extent it occurs. Moreover, our steady state analysis gives thresholds for rod performance in terms of overheating. Insulation of the rod should be able to keep its surface temperature below $313^\circ C$. We also note that the overheating of the tip might extend the length at which tissue is damaged (see Table 2). In addition, the diffusion length was up to half way to approximately 0.5cm. The basic rod-type probe design produced sub-optimal results in ablating all the tumor cells.

Future outlook considering the clinical need to effective ablation system; instead of assessing the heating mechanisms (RF, microwave, Ultrasound or laser), assessing the design of the delivery mechanism (probe) might result in an improved result. In addition, a direct injection of magnetic nanoparticles into the tumor, instead of letting it be naturally be up take by the tumor, might ensure the presence of the nanoparticles in the tumor and the RF heating can be enhanced.
References


[22] https://www.cancer.net/cancer-types/bladder-cancer/introduction


