A Computer Controlled Endorectal Cooling Device for Laser Thermal Therapy

by

Maged Maher Metias

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Institute of Biomaterial and Biomedical Engineering
University of Toronto

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Abstract

Interstitial laser thermal therapy is a novel local approach to treating prostate cancer. During treatment, thermal ablation may occur on the adjacent rectal wall. The aim of this thesis was therefore twofold: to study the effects of rectal cooling on lesion formation, and secondly, to engineer a computer controlled rectal cooling unit. To study the effects of the coolant temperatures and flow rate, thermal simulations were executed, followed by testing the phenomenon using agar gel phantoms which thermally mimic prostate tissue. Further simulations were run using a treatment planning software which predicted the required coolant temperatures to protect the outer rectal wall while subsequently determining the shape and size of the resulting coagulated lesion at various laser settings. Results suggest that low coolant temperatures and low flow rates cause maximum cooling rates. Furthermore, the shape and size of the coagulated region is affected by coolant temperatures at specific laser powers and positions within the prostate.
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Chapter 1
Introduction

1 Introduction

1.1 Background

The prostate is part of the male reproductive system which secretes a slightly alkaline fluid which helps protect the semen from acidic conditions found in the vaginal tract. The prostate is approximately the size of a walnut. It is located at the base of the penis and between the muscular tissues of the rectum and bladder, with the urethra running from the bladder through the prostate and to the penis [1].

![Figure 1.1 The human prostate and the surrounding tissue. Taken from [1].](image)

Prostate cancer is the most common male cancer in Canada [2]. Originating in the prostate gland, it has the potential of growing and metastasizing, albeit a few prostate cancers grow slowly.

There is no single cause of prostate cancer; however, there are several risk factors that can lead to the disease. These include old age, specifically 65 years of age or older, a family history of
prostate cancer, and having African origin. Those with high risk may be tested for prostate cancer by having a digital rectal exam for lumps and changes in the texture. A blood test screening for Prostate Specific Antigen (PSA) levels may also be used to detect prostate cancer. These are proteins that leak into the blood from the prostate. When the prostate is enlarged due to inflammation, because of either Benign Prostate Hyperplasia (BPH), or cancer, the PSA levels are much higher, and so PSA levels may or may not indicate the presence of cancer. Confirmation of cancer is made using biopsies under transrectal ultrasound imaging. This involves placing an ultrasound probe into the rectum to generate a 2-dimensional image of the prostate and to indicate biopsy trajectories. Samples are taken from the prostate tissue and analyzed to determine a Gleason score, which is a score between 2 and 10, where the higher the score the more likely the cancer is to spread. A Gleason score subsequently influences the treatment options that will be used [3].

1.2 Motivation

Prostate cancer affects 1 in every 7.1 Canadian men [4]. If left untreated, the cancer cells may metastasize which becomes fatal. Fortunately, several standard treatment options exist. These include external beam radiation therapy, brachytherapy, and radical prostatectomy, all of which assume the whole organ is diseased. However, for all of these treatments, side effects can occur, including impotence and urine leakage. Hormone therapy, another standard treatment option, blocks hormones in an attempt to bring the growth of the tumor to a stop. However, patients may experience abnormal sexual side effects, hot flashes or weakened bones [5].

More recently, new treatment options exist such as biological therapy, cryosurgery, high intensity focused ultrasound (HIFU), photodynamic therapy and interstitial laser thermal therapy. Biological therapy uses the patient’s own immune system to mount a specific attack on cancer cells. Cryosurgery has been used for whole organ treatment, which was found to cause impotence; however it needs to be further assessed for local therapy. HIFU was originally intended for whole organ treatment and also requires further assessment for local treatment. One limitation with HIFU is poor treatment of anterior regions of the prostate or undersized prostate. Furthermore urethral stenosis may also occur. Photodynamic therapy is a form of local therapy which makes use of photosensitive drugs which target the cancerous tissue and are activated by specific wavelengths of light delivered through a laser. This generates active radicals which are
toxic to the target tissue. Side effects may included urethral damage, erectile dysfunction, however, the extent of the side effects are dependent on the drug used [6]. Interstitial laser thermal therapy (ILTT) appreciates the locality of the diseases within the prostate. Thus, it is relatively accurate with minimal side effects.

Currently, we are investigating the coagulation of cancerous tissue in the prostate with laser light termed ILTT. Clinical trials have been conducted to study size and shape of the lesion based on power levels. The thermal energy from the laser propagates and may potentially damage the rectum, which is adjacent to the prostate. There has been little work done to investigate prostate temperature distributions from ILTT, however, Sherar et al. showed that localized heating in the prostate using microwave energy caused substantial heating on the rectal wall [7] and the rectal damage that would ensue does not readily heal [8]. As a result, it becomes necessary to protect the rectal tissue from ablation during ILTT. This may allow complete treatment of the prostate cancer without damage to the rectum or non-coagulated regions of prostate tissue adjacent to the rectum. Protection of the rectal tissue may be achieved by engineering a cooling device that cools rectal tissue to threshold temperatures such that only the prostate may be coagulated.

The proposed device will pump fluid through a computer-controlled heat exchanger, which uses the thermoelectric effect. The temperature regulated fluid will be controlled by calculating the necessary water temperature to keep rectal tissue below threshold temperatures based on laser fiber position and power. The coolant will be pumped into a balloon that fits onto an ultrasound probe to be placed against the patient’s rectal wall, on the side adjacent to the prostate. The ultrasound probe is necessary as it is inserted into the rectum to help visualize the location of the laser fibers in the prostate. Based on real-time temperature feedback from the rectum, the location of the laser and the rate of energy delivered, a computer will control the cooling power of the heat exchanger in order to maintain appropriate temperature gradients that protect the rectal wall while ensuring complete treatment of the prostate.

1.3 Project Goals and Overview

Cooling unwanted hotspots that develop on the rectal wall during prostate cancer treatment using will be realized through the design of a computer controlled cooling system. The following is a list of goals that are hoped to be reached through the manifestation of the final product:
1. Modify existing cylindrical balloons (Civco Medical Solutions, Kalona, Iowa) such that the cooling balloon may fit onto the ultrasound probe

2. Predict using computational software lesion size and required water temperature to protect the outer rectal wall during the duration of treatment while still providing complete treatment of the prostate.

3. Design a computer controlled cooling system that will be used to monitor water temperatures, and compute or measure prostate outer rectal wall temperatures. This will subsequently be used to determine protective water temperature.

4. Validate the computational results with physical experiments using phantoms with similar optical properties as human prostate.
2 Literature Review

2.1 Literature Review

There have been several minimally invasive treatments towards prostate cancer that involve some form of phototherapy. Furthermore, there were several attempts to protect the surrounding tissue from coagulating. This section discusses attempts to cool surrounding tissue.

Martin et. al. [9] used a transrectal antenna with a water cooled jacket to cool the antenna. The temperature of the water was manually selected. This subsequently protected the instrument during localized microwave hyperthermia treatment of BPH and avoids extreme heating of the rectum. The device consisted of a probe that housed a dipole antenna mount and a cooling system to cool the rectum (Figure 2.1).

![Figure 2.1 The Prostate Treatment Machine used during microwave hyperthermia treatment [9].]
During experimental trials with muscle tissue equivalent phantom it was found that the temperature of the cooling fluid does not change significantly and can be assumed to be constant at 10°C throughout the jacket. With power being applied in the range of 15 W and 30 W for 60 min, the rectal temperatures increased by 1.8°C and the peak temperature moves closer to the probe with maximum temperatures reaching approximately 46°C at 30 W. However, temperatures at the inner surface of the rectum did not surpass 40°C.

Nau et al. designed a multielement catheter-cooled interstitial ultrasound applicator to be used to coagulate tissue using thermal therapy (Figure 2.2) [10]. The cooling medium of degassed water provided two purposes, the first was to extract waste energy from the ultrasound transducers in order to allow the device to run efficiently thereby allowing greater power levels to be delivered. The second reason was to provide a good contact between the transducers and the tissue to allow optimal transmission of ultrasound energy.

With water at 7-40°C flowing at 20 to 60 ml min⁻¹ there was sufficient thermal lesions that had radial length of 10 mm at 5 W to 16 mm at 20 W for 4 min of heating. The probes were tested in pig thigh muscle and it was found that the shape and size of the lesion was independent of the fluid flow rate and temperature of the fluid within the catheter.

Davidson et al.[11] investigated two cooling catheters for their effective cooling in the urethra during prostate thermal therapy (Figure 2.3). He compared two catheters: Dornier Urowave catheter and a prototype from BSD Medical Corp., with a convection coefficient of 330 W m⁻²
°C\(^{-1}\) and 160 W m\(^{-2}\)°C\(^{-1}\), respectively. The two catheters were tested on a polyacrylamide muscle phantom with three thermistors aligned parallel at different distances from the antenna (Figure 2.3).

Figure 2.3 Setup simulating two cooling catheters cooling the urethra [11].

Figure 2.4 Temperature map of prostate as a result of urethral cooling [11].

It was found that the coefficient of convection was insensitive to flow rate between 40 and 120 ml min\(^{-1}\). Furthermore, his result revealed that there was a significant difference in temperature surrounding the immediate area between the two catheters; the Dornier Urowave cooled the
surrounding tissue by $10^\circ$C more than the prototype due to its higher convection coefficient and thinner wall tubing.

Sherar *et al.* used microwave thermal therapy to treat prostate cancer (Figure 2.5) [12]. He actively cooled the urethra and the rectum but also used hydrodissection which provided protection to the prostate by separating the prostate from the rectum. The urethra had a Foley catheter inserted into it, which had $10^\circ$C water flowing within a water circulation channel.

![Figure 2.5 Setup of simulation for microwave thermal therapy with urethral and rectal cooling [12].](image)

The hydrodissection technique was used by injecting 20 to 30 cc of sterile saline at room temperature to separate the rectum from prostate about 1cm. Finally, a rectal cooling unit with room temperature saline was used to cool the rectum. As a result of their cooling methods, the rectal and urethral temperature did not exceed $32^\circ$C while the hydrodissection did not exceed $35^\circ$C.

Despretz *et al.* investigated the technique of inserting rectal and urethral applicators (Figure 2.6) into their respective anatomical locations (Figure 2.7) and delivering microwave energy simultaneously to treat prostatic hyperplasia [13].
The rectal microwave antenna was incased within a Teflon tube with an external balloon while the urethral antenna was inserted into a plastic catheter which also had an external balloon. The rectal balloon was diametrically opposite to the antennas and allowed for accurate positioning along the rectal wall. Both applicators were cooled directly by flowing water at 120 ml min\(^{-1}\) through the catheter. The rectal probe was cooled with 30°C and the urethral catheter with 25°C.
It was found that the thermoregulation from the coolant maintained the urethral and rectal wall below 44°C.

2.2 Summary and proposed method

Reviewing the literature provided a thorough insight on the several types of designs that were seen to be effective in cooling hotspots in the tissue or the applicator. The approach that we desired to take was one that provided a more versatile temperature controlled system that is easy to setup, does not interfere with the ultrasound imaging and one that is able to cool down tissue at all laser power levels.

It was first proposed to use convective cooling using room-temperature air to chill the coolant. While this may work, it does not allow versatility in temperature modulation. Further, the temperature would have a lower limit of 20°C, which may not provide sufficient cooling at high laser powers. Therefore, convective cooling using air was ruled out and other solutions were investigated. On such solution is through the use of Peltier devices. These are small semiconductor units designed to pump heat from one side when a current is passed through it thereby creating a hot and cool side. The use of Peltier devices will allow for computer regulated coolant temperatures, larger gradients and relatively shorter cooling times.

Of the several cooling units that were presented within this chapter, Sherar et al. and Despertz et al. provided similar rectal cooling designs that were effective in protecting the rectum from ablation at their respective power delivery level. However, these systems would need to be adapted for interstitial laser thermal therapy. Firstly, the cooling sheath will need to be able to fit tightly onto an endorectal ultrasound probe without affecting image quality. Also, the water temperature needs to be actively regulated to changing tissue temperatures either due to rising tissue temperature or a changing laser power level and position. This dynamic approach would allow improved versatility and convenience.

During treatment, the outer rectal wall temperature will need to be sustained below a certain temperature to prevent coagulation over the treatment time. In order to accomplish this, the water temperature will be set to a calculated level necessary for protection at a given laser power and position. This will be achieved by running computational software that will simulate treatments at several laser powers over a 10 minute period (a given time interval that was used in clinical
trials), thereby elucidating the required water temperature that would keep the outer rectal wall from ablation. On the left in Figure 2.8 is a cross-sectional view of the setup within the patient, while on the right is a schematic of an expected temperature profile along the cross-section A-A.

Figure 2.8 [Left] A cross-sectional view of the rectum and cancerous prostate. [Right] A sample temperature profile along the cross-section A-A. Temperatures and profile shape were estimated.
Chapter 3
Methods

3 Methods

This section reviews the procedures and setups that were designed during the simulations and laboratory experiments. Each step throughout the project served to answer specific scientific questions, specifically the affects of physical phenomenon on the treatment of the prostate and protection of the rectum. This section is divided into three experiments or simulations. The first simulation was run in COSMOSXpress to determine how coolant temperatures and flow rates protect the outer rectal wall. The second simulation was executed in a treatment planning platform (TPP) software to study how lesion volumes and shapes were affected by the coolant temperature. The software will also help determine the coolant temperature necessary to protect the outer rectal wall at various laser powers and positions. Finally, phantom experiments were conducted to determine the cooling capacity of the device as well as to confirm the simulations from COSMOSXpress.

3.1 COSMOSXpress Simulations

In order to design a cooling system, it is important to understand the heating phenomenon of the tissues under question, specifically the prostate and rectum. Several permutations of fluid flow rate and fluid temperature will be run to determine the optimal flow rate at typical power levels used during therapy.

3.1.1 Design of the Anatomy in COSMOSXpress

Several components of the anatomy and instruments were designed in SolidWorks. This is a computer-aided design software that allows the user to construct 3D structures. The structures are then imported into COSMOSXpress and subjected to physical forces and gradients. COSMOSXpress is a finite analysis tool that divides the 3D model into several elements and performs the analysis element by element using the preprogrammed governing equations for the specific physical phenomenon being simulated.
Using SolidWorks, six structures were constructed and assembled together to form a model of the human anatomy and the equipment setup. The prostate was designed as a cube with a volume of 40.823 cm$^3$ as this is close to the size of prostates with carcinoma [14]. A slight groove at the base of the prostate was inserted to allow more surface contact between the prostate and the rectum, which was designed to be a hollow circular tube. This setup was done to mimic intra-operative conditions where the prostate experiences a slight deformation when a probe is inserted into the rectum. The deformation allows for increased tissue contact between the prostate and rectum (Figures 3.1 and Figure 3.2), and therefore reduced thermal resistance. Also, the design incorporated cylindrical invaginations for a laser probe and seven thermocouples at several lengths above the probe. It is important to note that although during clinical trials the laser position would be placed 2-5mm above the outer rectal wall, the simulations place the laser directly atop the rectal wall. The reason behind this setup was that past clinical trials did not cool the rectum. Thus, having the laser at closer proximities to the rectum was not possible; however, we wanted to determine if such close proximities were possible with the incorporation of rectal cooling (Figure 3.2). The size of the laser probe and thermocouples were similar to the size of the intra-operative tools. The laser probe was taken as 2cm long and 1.9mm diameter while the thermocouples were 2cm long and 200 microns in diameter.
Figure 33.1 Intraopratve ultrasound image of the prostate. The dark figure is the prostate which experienced a deformation when the ultrasound probe is inserted into the rectum.
To perform the simulations, it was necessary to assign physical properties to the anatomy and instruments. The prostate was given physical properties that were used were taken to be similar to water, at 1004.5 kg/m\(^3\) [7], while tissue specific heat, \(c(T)\), and thermal conductivity, \(k(T)\) of the prostate were approximated according to the dynamic changes of the thermal properties of water in the range of 20-100°C [15]:

\[
k(T) = 0.419(0.133+1.36k_w)[\frac{W}{m \cdot K}]
\]

Equation (3.1)
\[ c(T) = 4.190(0.37 + 0.63k_w w)[\frac{J}{Kg \cdot K}] \]

**Equation (3.2)**

where \( w \) is the percent mass of water in the tissue and was taken to be about 69%, \( k_k \) and \( k_c \) are the temperature dependant factors for conductivity and specific heat, respectively. These are defined as [15]:

\[ k_k = 1 + 1.78 \times 10^{-3}(T - 20^\circ C) \]

**Equation (3.3)**

\[ k_c = 1 + 1.016 \times 10^{-4}(T - 20^\circ C) \]

**Equation (3.4)**

This results in a linear relationship between the temperature and the physical properties (Figure 3.3 and Figure 3.4). However, over a temperature change of 20°C, the values of the specific heat and thermal conductivity do not change significantly.
Figure 3.3 Specific heat of the prostate as a function of temperature.

Figure 3.4 Thermal conductivity of the prostate as function of temperature.
The rectum is a multilayered tissue with a longitudinal muscle coat, a circular muscle coat, the muscularis mucosa and the mucous membrane [16]. However, the rectal structure was simplified and assumed to be a complete muscular structure. Thus, the density, specific heat and thermal conductivity where assumed to be that of muscle and taken to be 0.556 kg/m$^3$, 3800 J/(kg*K), and 0.42 W/(m*K), respectively [17]. The length of the rectum was taken to be close to physiological, with a length of 11cm, inner diameter of 3 cm [16] and a rectal wall thickness of 2.6mm (Figure 3.5) [18].

![Figure 3.5 SolidWorks model of the rectum.](image)

The cooling balloon was designed to fit within the rectum so as to maximize surface contact. It was designed as an extruded semicircle made from polyethylene. This balloon’s purpose is to house circulating coolant which serves to regulate the rectal wall temperature during thermal therapy. The balloon was made from polyethylene and its dimensions include a length of 13mm an internal diameter of 14.65mm and a wall thickness of 0.35mm (Figure 3.6). Table 3.1 summarizes the physical properties of the polyethylene.
Finally, the thermocouples were taken to be insulators. This would not have a significant affect on the heat transfer as the diameter of the thermocouples is 200 microns, and therefore significantly smaller than the prostate. However, they can be inserted into catheters which are approximately 2 mm in diameter.

### 3.1.2 Simulation: Setup and Governing Equations

Several conditions were set for the simulation. The prostate is a vascularized organ, where the blood may be assumed to serve as a physiological coolant during thermal therapy and therefore removes thermal energy away from the prostate [9]. However, it is difficult to design a prostate with a capillary network embedded within it. In order to alleviate this problem, an assumption was made in that the blood coming into the prostate experienced a 1°C increase in temperature as it exited.
To calculate the amount of energy that the prostate tissue (apart from the capillaries) experienced, it was necessary to calculate the power delivered. The energy profile of the intra-operative laser power profiles were recorded based on clinical experience (Figure 3.7); from this, a time averaged power level was calculated to be 2.3W.

![Figure 3.7 Power profile of laser during therapy.](image)

In order to input the laser power into COSMOSXpress, the effective laser power, in the absence of simulated blood perfusion, is required. The prostate volume was designed with the following parameters:

\[
V = 40cm^3 \\
\rho = 1.0045g/cm^3 \\
m = \rho V \\
m = (1.0045g/cm^3) \times 40cm^3 \\
m = 40.18g
\]

During thermal therapy, the perfusion of the prostate increases from 10 ± 8ml/100gm – min to 14 ± 2ml/100gm – min [20]. Thus, an average perfusion rate was chosen at 12/100gm – min such that

\[
\text{Volumetric perfusion rate} = \frac{0.12ml}{gm \cdot \text{min}} \cdot 40.18g
\]

\[
\text{Volumetric perfusion rate} = 4.82ml/\text{min}
\]

\[
\text{Volumetric perfusion rate} = 4.82 \times 10^{-6} m^3/\text{min}
\]
Knowing the perfusion rate, the blood specific heat \([17]\), blood density \([21]\) and assuming that the blood temperature rises from \(37\, ^\circ C\) to \(38\, ^\circ C\) then it is possible to calculate the power that the blood removes from the prostate.

\[
Q = 4.8\text{ml/min} \quad \text{T}_f = 38\, ^\circ C
\]

\[
\bullet \quad \dot{q} = m \cdot C_p \left( T_f - T_i \right)
\]

\[
\bullet \quad \dot{q} = \rho_{\text{blood}} \cdot Q \cdot C_p \left( T_f - T_i \right)
\]

\[
\bullet \quad \dot{q} = \frac{1043\, \text{kg}}{m^3} \cdot \frac{1000\, \text{g}}{\text{kg}} \cdot 4.82 \times 10^{-6} \cdot \frac{\text{m}^3}{\text{min}} \cdot \frac{1\, \text{min}}{60\, \text{s}} \cdot 3.84 \frac{\text{J}}{\text{gK}} \cdot (311 - 310)\, \text{K}
\]

\[
\bullet \quad \dot{q} = 0.32\, \text{W}
\]

Where \(\dot{q}\) is the power removed from the prostate by the blood, and \(m\) is the mass flow rate of the blood, \(C_p\) is the specific heat, \(T_f\) and \(T_i\) are the final and initial blood temperature, respectively, and \(Q\) is the volumetric flow rate of the blood.

Subtracting this rate of energy from the time averaged power of 2.3W, a total of 1.98W of energy is absorbed by the prostate. Thus, the simulated laser fiber will generate a power of 1.98W during the simulation.
In addition to this laser power, the rectum was assumed to generate its own power due to metabolic rate of 684 W/m$^3$ [22]. For the rectum that is to be simulated this is a total power of:

\[
Power = 684 \frac{W}{m^3} \cdot \pi L (r_{outer}^2 - r_{inner}^2)
\]

\[
Power = 684 \frac{W}{m^3} \cdot \pi \cdot (110 \times 10^{-3} m) \cdot [(15 \times 10^{-3} m)^2 - (17.6 \times 10^{-3} m)^2]
\]

\[
Power = 0.02 W
\]

Where L is the length of the rectum and $r_{inner}$ and $r_{outer}$ are the inner and outer radius of the rectum, respectively. No power generation for the prostate was considered as this would be insignificant to the power delivered to the rectum by the laser.

Using the laser power and material properties stated above, the following permutations of fluid temperature and flow rate were simulated (Table 3.2). As shown the flow rate and temperature of the coolant were altered during the simulations in order to evaluate the cooling effect of each on the rectal wall at close laser proximities.

Table 3.2 The scenarios of the coolant temperature and flow rate that were simulated in COSMOSXpress.

<table>
<thead>
<tr>
<th>Flow rate ([x10^{-3} \text{kg/s}])</th>
<th>Temperature of Coolant(^{\circ} \text{C})</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>10</td>
</tr>
<tr>
<td>5</td>
<td>5</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td>15</td>
<td>15</td>
</tr>
<tr>
<td>20</td>
<td>20</td>
</tr>
</tbody>
</table>
Finally, the assembly of the prostate, rectum, cooling balloon and the thermocouples were placed in such a position that could eventually be used during an clinical setup. Furthermore, it was assumed that there was no contact resistance between any of the structures. That is, all the structures were completely touching the adjacent structures (Figure 3.9). This is true especially for the inflated balloon, such that it was in complete contact with the rectal wall. Subsequently, the presence of the balloon would push the rectal wall against the prostate for good contact.

Figure 3.9 Setup of the prostate, rectum, balloon, laser fiber and thermocouples.
3.2 Treatment Planning Simulations

A second set of simulations of interstitial laser thermal therapy was executed using a treatment planning software. The program uses transverse images of the prostate, typically from transrectal ultrasound images, with image slices separated by 5 mm beginning from the apex to the base of the prostate. The prostate in these images was contoured, with these tracings used to construct the prostate geometry. The temperature distribution in both the rectum and the prostate was calculated using the finite element method to derive a mathematical equation based on heat transfer principles. The goal was to determine the required water temperature that would cool, and therefore protect the rectum from coagulation, while also predicting the size of the lesion that would result from treatments at several constant power levels.

3.2.1 Governing Equations

From previous experimental and clinical trials, laser treatment usually lasted 10 minutes. It is necessary that during this time the thermal dose is high enough to cause coagulation and form a lesion. However, lesion formation depends on both the length of treatment time as well as temperature. Thus in order to protect the rectal wall adjacent to the prostate, it is necessary to determine the threshold temperature at which coagulation will occur at the end of 10 minutes. This temperature would be used as the upper limit during future clinical trials which would be designed to protect the rectum from ablation.

The threshold temperature was determined using the concept of equivalent minutes at 43°C for calculating the thermal dose delivered to the tissue during the treatment [23]. The thermal dose can be found using the Sapareto-Dewey integration as a function of time and temperature:

\[
t_{43} = \sum_{t=0}^{t_{final}} R^{(43-T_t)} \Delta t
\]

Equation 3.5

Where R is equal to 0.5 for temperatures exceeding 43°C and 0.25 for temperatures below 43°C; choosing a constant value of R, in our case, 0.5, results in an error less than 2 percent [24]. \(T_t\) is the average tissue temperature during treatment over \(\Delta t\); and \(t_{43}\) is the equivalent time at 43°C which causes thermal damage. According to Damianou & Hynynen [23], it requires 120 min to 240 min for muscles to coagulate at 43°C. Taking a cautious approach, \(t_{43}\) was taken to be 120
min. Figure 3.11 is a graph of Equation (3.5) which reveals the temperature and respective times at which coagulation is expected to occur. From Figure 3.10, shorter times lead to higher temperatures, albeit not significantly higher for a 10 minute duration.

Figure 3.10 The respective temperatures at a given time duration at which coagulation is expected to occur. The above figure is adapted from 120 equivalent minutes at 43°C.

Thus, according to Figure 3.10, the limiting temperatures do not vary significantly as treatment duration is changed. Therefore, a conservative temperature of 46°C was chosen as the outer rectal wall temperature in order to prevent damage.

The temperature distribution was calculated using the finite element method applied against the Pennes bioheat equation, which incorporates tissue thermal properties as well as the blood perfusion rate [23]. Our purpose was to determine the appropriate temperature of water that is required to keep the outer rectal wall temperature below 46°C. This requires that an equation be derived that models heat flux through a hollow cylinder, representing heat flux through the rectum. This will allow us to calculate the required cooling temperature that will maintain the outer rectal wall at 46°C. Figure 3.11 is a simple schematic of the coordinate system used to
define the prostate geometry. The following is a derivation of the governing equation that was used:

In order to calculate the temperature profile, the cylindrical form of Fourier-Biot Equation [25] must be considered (Equation (3.6)). This considers heat flow in three directions, the radial direction, \( r \), the azimuthal direction, \( \phi \), and along the axis of the cylinder, \( z \).

\[
\frac{1}{r} \frac{\partial}{\partial r} \left( k r \frac{\partial T}{\partial r} \right) + \frac{1}{r^2} \frac{\partial}{\partial \phi} \left( k r \frac{\partial T}{\partial \phi} \right) + \frac{\partial}{\partial z} \left( k \frac{\partial T}{\partial z} \right) + g = \rho C \frac{\partial T}{\partial t}.
\]

Equation 3.6)

Figure 3.11 Geometry of Prostate used for simulations in treatment planning software.
Several assumptions were made which simplifies the equation. The first assumption made was that the tissue temperature reaches steady state conditions during treatment, such that the term \( \frac{\partial T}{\partial t} \) becomes zero. Furthermore, we assume that the thermal conductivity of the muscle is constant regardless of whether the tissue is coagulated or not. Also, heat transfer is taken to occur in one direction since the surface area of the rectum is large relative to the rectal wall thickness. Therefore, the second term from the left can be ignored. Also we assume that the heat generation due to normal physiological metabolism of the muscle is negligible, allowing us to set \( \cdot g \) to zero. Finally, since the analysis is in the radial direction, we could ignore the third term from the left. Ultimately, the equation simplifies down to

\[
\frac{1}{r} \frac{\partial}{\partial r} \left( kr \frac{\partial T}{\partial r} \right) = 0
\]

\[
\frac{\partial}{\partial r} \left( r \frac{\partial T}{\partial r} \right) = 0
\]

\[
\left( r \frac{\partial T}{\partial r} \right) = C_1
\]

\[
\frac{\partial T}{\partial r} = \frac{C_1}{r}
\]

\[
T(r) = C_1 \ln(r) + C_2
\]

Equation 3.7)

Equation (3.7) has two unknowns; however, we know that \( T_2 \leq 46^\circ C \) (Figure 3.11) which is necessary to prevent rectal hyperthermia and \( T_1 \) represents the temperature at the fiber tip. In other words, at \( r_1 \) and \( r_2 \) we get

\[
T(r_2) = C_1 \ln(r_2) + C_2 = T_2
\]

Equation 3.8)
\[ T(r_i) = C_1 \ln(r_i) + C_2 = T_i \]

**Equation 3.9)**

Rearranging Equation (3.8) for \( C_2 \) and substituting it into Equation (3.9) we get

\[ \therefore C_2 = T_2 - C_1 \ln(r_2) \]
\[ \therefore T_i = C_1 \ln(r_i) + T_2 - C_1 \ln(r_2) \]

**Equation 3.10a) (top) and (3.10b) (bottom)**

Rearranging Equation (3.10b) we get an expression for \( C_1 \)

\[ T_1 - T_2 = C_1 (\ln(r_1) - \ln(r_2)) \]
\[ T_1 - T_2 = C_1 \left( \ln \left( \frac{r_1}{r_2} \right) \right) \]
\[ C_1 = \frac{T_1 - T_2}{\ln \left( \frac{r_1}{r_2} \right)} \]

**Equation (3.11a), Equation (3.11b) and (3.11c) (top down)**

Substituting Equation (3.11c) into Equation (3.10a) we get the following expression for \( C_2 \)

\[ \therefore C_2 = T_2 - \frac{T_1 - T_2}{\ln \left( \frac{r_1}{r_2} \right)} \ln(r_2) \]

Substituting both \( C_1 \) and \( C_2 \) into Equation (3.7) and simplifying, we get the following governing equation:
The governing equation can be developed further by incorporating the distance of the laser fiber, whose temperature will be used as a boundary. If we take ‘D’ as the distance of the fiber from the rectal wall then

\[ r_1 = r_2 + D \]  

**Equation 3.13**

Substituting Equation (3.13) into Equation (3.12), we get

\[
T(r) = \frac{T_1 - T_2}{\ln \left( \frac{r_1}{r_2} \right)} \left[ \ln(r) - \ln(r_2) \right] + T_2 
\]

**Equation 3.14**

The form of the governing equation can be modified for the fiber distance, the outer rectal wall temperature, and the rectal thickness.

The simulation program was design to run calculations on interstitial laser thermal therapy in human prostate. Thus, we needed to define the optical properties to model the laser light propagation in the tissue and the prostate thermal properties which will be used to model heat flow through the tissue. Table 3.3 summarizes the properties that were incorporated into the simulations. However, it is important to note that due to the limitation of the simulation software, all tissue present in the simulation was assumed to be prostate and given the prostate’s physical properties. Given that both the rectum and prostate contain smooth muscle, this approximation is reasonable and should not cause significant changes in the results.
### Table 3.3 Physical Properties of Human Prostate Tissue and Blood

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Blood Heat Capacity</td>
<td>3.84 J/g K [7]</td>
</tr>
<tr>
<td>Prostate Thermal Conductivity</td>
<td>0.004930 W/cm K [7]</td>
</tr>
<tr>
<td>Tissue Specific Heat</td>
<td>3.90 J/g K [7]</td>
</tr>
<tr>
<td>Tissue Density</td>
<td>1.0045 g/cm³ [7]</td>
</tr>
<tr>
<td>Human prostate (native) coefficient of absorption</td>
<td>0.6 cm⁻¹ [26]</td>
</tr>
<tr>
<td>Human prostate (coagulated) coefficient of absorption</td>
<td>0.7 cm⁻¹ [26]</td>
</tr>
</tbody>
</table>

In order to run the simulations, it was necessary to define the boundaries of the prostate and the surrounding organs such as the urethra and the rectum. As show in Figure 3.12, the inner and outer rectal walls were defined as two semicircular boundaries in green, with the outer wall having a greater radius. The simulations were designed such that the water temperature was taken to be equal to the inner rectal wall. The purpose was to calculate the appropriate inner rectal wall temperature (or water temperature) such that the outer rectal wall would not exceed 46°C during treatment. Furthermore, the distance between the two semicircular boundaries, which constituted the thickness of the rectal wall was set to approximately 3mm [27]. The urethra boundary was defined as a small circular boundary in the center of the prostate.

Hyperthermia was generated in the prostate using a (diffuse) laser placed in the lateral end of the prostate as shown by the green dot in Figure 3.12. The laser tip was modeled as a 2 cm long diffusing end. Furthermore, the computer model incorporated 25 thermistors placed, as shown in
Figure 3.12, on a diagonal line that is arranged perpendicular to the rectal wall and dissects the laser fiber. Fibers were used such that a temperature map may be constructed for each scenario executed.

The treatment planning software was set to run several treatment scenarios. This involved a permutation of several power levels and laser fiber distances. Laser powers began at 2W and were increased by increments of 1 W until the cooling water temperatures required to protect the outer rectal wall dropped down to freezing temperatures. Furthermore, the distance of the laser fiber from the outer rectal wall placed at 5 - 10 mm for each scenario, in increments of 1 mm. Table 3.4 summarizes the scenarios that were run.
### Table 3.4 Scenarios That Were Simulated in the Treatment Planning Software

<table>
<thead>
<tr>
<th>Laser Fiber Distances from Outer Rectal Wall</th>
<th>Power Used</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 mm</td>
<td>2-5W</td>
</tr>
<tr>
<td>6mm</td>
<td>2-5W</td>
</tr>
<tr>
<td>7mm</td>
<td>2-6W</td>
</tr>
<tr>
<td>8mm</td>
<td>2-7W</td>
</tr>
<tr>
<td>9mm</td>
<td>2-8W</td>
</tr>
<tr>
<td>10mm</td>
<td>2-10W</td>
</tr>
</tbody>
</table>

### 3.3 Design of the Endorectal Cooling Unit

The cooling unit was designed such that it was capable of computer-controlled cooling of the water. This was accomplished through the use of a Peltier device which was connected to a programmed controller. The cold side of the Peltier device was connected to a copper heat exchanger in which water was forced through by a pumping unit. The hot side of the Peltier device was connected to a large fin that was cooled using air convection cooling. Fluid exiting the heat exchanger entered a small reservoir where the temperature was measured and fed back to the controller using a thermistor. Finally, the coolant was pumped through long tubes that inflated the endorectal balloon for subsequent cooling of the rectal wall. This section discusses in detail the design and properties of the cooling unit that was used to protect the rectal wall from damage.
3.3.1 Endorectal Balloon

In order to cool the rectal wall an endorectal balloon from Civco Medical Solutions was adapted (Figure 3.13). The balloon’s original use is to act as a condom for ultrasound probes such that the added balloon volume (with stagnant water inflating the balloon) would allow the rectum to push up against prostate to a desired shape and height to allow for enhanced image quality.

![Endorectal balloon placed on ultrasound probe. Taken from [28].](image)

The balloon measures approximately 14 cm in length with two cavities: one for the balloon and another as a sleeve which tightly fits onto the ultrasound probe. The walls of the balloon are approximately 1 mm thick. The inner diameter is approximately 20 mm at maximum inflation. The balloon is constructed from polyethylene whose properties are summarizes in table 3.1.

As shown in Figure 3.13, a single tube is fed into the endorectal balloon in order to inflate it with water. This configuration does not allow for circulation necessary for temperature regulation of the balloon’s fluid. Thus, it was necessary to modify the balloon such that an inlet tube and an outlet tube replaced the single tube. Due to the physical restriction of the balloon, the new tubing had a smaller diameter, with an inner diameter of 0.08 inches for both the input and output tubes.

3.3.2 Reservoir, Pump and Thermistors

During surgery it is required that all instrumentation that are not sterile be at least 2 feet away from the patient. This constitutes the minimum length of external tubing of the endorectal
balloon. Fluid resistance is inversely proportional to the quartic of the radius, as suggested by Equation (3.15) [29].

\[ R = \frac{8\mu L}{(\pi r^4)} \]

**Equation 3.15**

By increasing the length and decreasing radius of the tubing the resistance of tubing network increases. Due to the small inner radius of the tubing leaving and entering the balloon, as well as its potential length of at least 2 feet, the resistance of the tubing would be relatively high. As a result, a powerful pump with a large head would be necessary. The pump that was incorporated into the device was Swiftech MC655 pump with a head of 10 feet. Pump performance is found in Figure 3.14. It is important to note that the pump has the option of speed control. This will allow us to elucidate whether water flow rate affects rectal cooling.

![Figure 3.14 Performance curves at various pumping speeds. Taken from [30].](image-url)
Temperature regulation was accomplished by continuous sampling at a rate of 1 Hz of the water temperature that was in a reservoir. The clear acrylic reservoir can hold 133cc volume. A hermetically sealed thermistor (OMEGA Engineering, Stamford, Connecticut), with a resistance of 10,000 Ohms at 25°C, was placed into the reservoir through a drill hole and sealed at the top with silicon. The thermistor is capable of operating within a temperature range of -80°C to 150°C with a 3.75 second response time and an interchangeability of ±0.1°C between 0°C and 70°C.

A temperature sensor is placed into the reservoir because the coolant in the reservoir is pumped to the patient, and so the temperature is fed back to the computer controller to compare it against the set temperature. If the measured water temperature deviates from the set temperature, the computer tries to readjust the water temperature. Ideally, it would be more accurate to measure

Figure 3.15 Swiftech Pump. Requires 8V nominal voltage and 2 Amps. Taken from [30].
the water temperature within the balloon, however, due to the small diameter of the tubing which connects to the balloon; this would completely obstruct the flow into or out of the balloon.

3.3.3 Heat Exchanger

Low coolant temperatures are accomplished using thermoelectric cooling using a Peltier module. This is a simple reversible heat pump that uses electrical energy to pump thermal energy across the module. Peltier modules are arranged such that two complementary semiconductor blocks of equal size, negative type (n-type) and positive type (p-type), are arranged adjacent to one another, although physically separated. Furthermore, these blocks are connected at the ends through a copper plate to the adjacent complementary semiconductor (Figure 3.16).

![Figure 3.16 Peltier modules (N+P) connected in series. Taken from [31].](image)

Peltier modules, which consist of two legs, n-type and p-type semiconductors, work by absorbing or releasing heat when a current passes between two dissimilar conducting materials, which is usually between the semiconductor (usually Bismuth Telluride) and the copper conducting wiring. Heat moves in the direction of the charge carrier [31]; in the n-type pellet, the charge carriers are the electrons while the p-type contains holes, which are positive charge carriers. To better understand the concept of n-type and p-type pellets, consider the example of a silicon lattice structure. Silicon is a semiconductor which has 4 valence electrons. However, when the silicon lattice structure is doped with Phosphorus (with 5 valence electrons and is approximately the same size as silicon), they substitute silicon atoms within the lattice structure. However, only 4 valence electrons are needed to form bonds with the surrounding silicon atoms. As such, the extra electron in Phosphorus is weakly bound and may be easily excited and conducted through the doped silicon lattice. This is termed a charge carrier. Thus, the material becomes a negative type or n-type semiconductor.
Analogously, if the same silicon lattice structure is doped instead with group III, such as Boron which has 3 valence electrons instead of the 4 found in silicon, then bonding with the surrounding silicon atoms within the lattice is not satisfied. The vacant position in the valance orbital is termed a hole and can easily accept an electron. Thus, the material becomes a positive type or p-type semiconductor [32].

Applying a current to flow through Peltier modules as shown in Figure 3.16, causes electrons to flow up though a n-type pellet and through the p-type pellet, and then continue on to the next n-type pellet connected in series and ultimately to the whole Peltier device (Figure 3.17).

![Diagram of Peltier device](image)

**Figure 3.17** The elements of a Peltier device. When activated with a current, heat is pumped from the cold side and released at the hot side [33].

It follows that due to the current density flowing in both pellets, that a heat flux is generated from the cold side to the hot. The heat that is pumped is equal to the combined heat pumped in each leg within a Peltier module. In other words, the heat pumped is equal to [34]

\[ Q = Q_n + Q_p \]

**Equation (3.16)**
Where $Q$ is the heat pumped through the pellets that are arranged in thermally in parallel. The magnitude of heat pumped can also be found by using

$$Q = \Pi_{AB} \cdot i$$

Equation 3.17)

Where $\Pi_{AB}$ is the Peltier coefficient and is equal to the addition of the absolute Peltier coefficients for the dissimilar materials $A$ and $B$ which comprise the Peltier module:

$$Q = (\Pi_A + \Pi_B) \cdot i$$

Equation 3.18)

Thus, by increasing the current, the heat pump will theoretically increase as well. However, there are always losses involved in heat pumps. Heat absorbed at the cold junction in a Peltier device due to a current $I$ is modeled by [35]

$$Q_C = \alpha d T_C - \frac{1}{2} I^2 R - \kappa(\Delta T)$$

Equation 3.19)

Where the first term on the left is the Peltier heat absorbed, the second (middle) term is the Joule heat flowing to the cold junction, and the last term on the right end of the equation is the heat conducted back to cold junction.

It follows that since the Peltier device is an electrical heat pump, that it has a Coefficient of Performance (COP) which is a measure of the device’s efficiency. This is given by

$$COP_{\text{max}} = \frac{T_{\text{cold}}}{\Delta T} \cdot \frac{\sqrt{1 + ZT} - \frac{T_{\text{Hot}}}{T_{\text{Cold}}}}{\sqrt{1 + ZT} + 1}$$

Equation 3.20)
Where ZT is known as the figure of merit and is equal to 1 in commercial Peltier devices to [34].

The Peltier device that was used was an 8A 12V device. The hot surface of the Peltier device was in contact with a large aluminum heat sink that was air cooled with a high flow rate fan. The cool surface of the Peltier device was in contact with a copper heat exchanger which absorbed heat from the water flowing within it (Figure 3.18)

![Figure 3.18 Heat exchanger modeled in SolidWorks.](image)

The inlet and outlet of the heat exchanger have dimensions of 3/8 inches diameter. As shown in Figure 3.18, a water channel forms a groove into the copper with a depth of 11mm and a straight length of 240 mm. Channel width is approximately 4.35 mm giving a total surface area of 6324 mm². The large surface area allows for optimal heat transfer between the flowing water and the copper heat exchanger.

The copper base plate has a circular bottom, with a diameter of 4cm, while the Peltier device is square with a length of about 4cm. Thus, although the majority of the surface area of the Peltier device is...
device would be in contact with the copper base plate, the corners, however, would be exposed to the ambient air temperature. Hence to alleviate any loses due to temperature differences between ambient air and the cooled surfaces, both the copper base of the heat exchanger and exposed cold side of the Peltier device were insulated with approximately 1 cm thick Styrofoam.

The Peltier device was connected to the 5R7-001 TE module RoHS controller (Oven Industries, Mechanicsburg, Pennsylvania). This module connects to a computer through a RS232 port and is capable of delivering a voltage between 12-36VDC. More importantly, it has an output load between 0.1 Amps to 25 Amps, which is beneficial to controlling the rate of heat pumped across the Peltier device. Its ability to be programmed allows us to set control temperatures through computer algorithms to modulate the water temperature. Finally, it has the capability of reading from two thermistors, which will allows us to read both the temperature of the water and of the prostate as feedback data to help regulate power output to the Peltier device.

3.3.4 The LabVIEW Program

The Peltier controller was programmed through two means. The first was through the use of LabVIEW, which is a graphical programming software used for measurement or automation. In order to communicate with the Peltier controller, a specific driver was required that was compatible with LabVIEW. A free driver was found and downloaded from Kajei.

The LabVIEW program was divided into three steps: Initiation, continuation and termination. The initiation step confirms that the Peltier device is turned off and then initializes the Peltier device by setting the bandwidth to ten. In other words, the bandwidth is centered at the set temperature with a range of five above and below the set temperature. If the temperature that is read falls within the bandwidth, the power to the Peltier device is reduced linearly with the temperature. The second step, continuation, is activated when the user activates the program. In this step, the computer takes input measurements from the thermistor in the water reservoir. Since the water from the reservoir is fed directly to the balloon, the temperature from the reservoir was taken to represent the balloon temperature, which in turn was used to represent the inner rectal wall temperature. This is a good approximation as the results from the COSMOSXpress suggest that the water temperature is approximately equal to the inner rectal wall temperature. The program also receives input from the user regarding the position of the laser from the rectal wall. Furthermore, the temperature at the laser was taken as 100°C, which
again is an approximation as water will not exceed 100°C until all the water has evaporated. The
rectal wall temperature was taken to be at 46°C. Having this information, the LabVIEW program
uses Equation (3.14) to calculate the required inner rectal wall temperature in order to maintain
the outer rectal wall at 46°C. The output from Equation (3.14) will be the set temperature of the
coolant against which the reservoir temperature will be compared. If the temperature of the
reservoir does not match the set temperature, the Peltier device continues to either cool or heat
the water until the set temperature is reached. The final phase, termination, is activated when the
user has deactivates the program. This discontinues the power of the Peltier device and saves the
data to a file specified by the user.

3.3.5 Tubing and Energy Losses

The next step was to assemble the cooling unit. This involved connecting tubing to the various
components of the device to form a plumbing network. Due to the various inlet and outlet sizes
of each component, as well as the tubing length, there are several minor losses in the fluid’s
energy. It follows that in any piping system, several losses may be present that reduces the
energy of the flow. These losses may be due to entrance or exit into or out of pipes, respectively,
expansions or contractions within a pipe, bends, elbows, valves, and gradual expansions or
contractions [29]. Minor losses are defined through the loss coefficient which is the ratio of the
head loss \( h_m = \Delta p / (pg) \) to the velocity head \( V^2 / 2g \) as the flow passes through the pipe. Note
that the loss coefficient, \( K \), is dimensionless:

\[
K = \frac{\text{HeadLoss}}{\text{VelocityHead}}
\]

\[
K = \frac{2g \cdot \Delta p}{V^2 \cdot pg}
\]

\[
K = \frac{\Delta p}{1/2 \rho V^2}
\]

\text{Equation 3.21)}

Where \( \Delta p \) the pressure difference, \( g \) is the acceleration due to gravity, \( V \)’s the velocity of the
fluid, and \( \rho \) is the fluid density.
To find the total minor losses of a system, all the loss coefficients, $K$, are added together along the length of the pipe [29]. This section will summarize and calculate the respective tubing lengths and inner diameters while calculating the total loss of the plumbing system. Finally, a water flow rate will be determined using Bernoulli’s energy equation.

i) Calculate Reynolds Number at the pump’s outlet (1/2” diameter):

$Q = \frac{\pi}{4} (D)^2 \cdot V$

$\frac{3.33 \times 10^{-3}}{m^3/s} = \frac{\pi}{4} (0.0127m)^2 \cdot V$

$V = 2.63m/s$

**Equation 3.22**

Where $A$ is the cross-sectional area of the tube and $V$ is the velocity of the fluid.

\[
\therefore \text{Re} = \frac{\rho V D}{\mu}
\]

\[
\text{Re} = \frac{(998 \frac{kg}{m^3})(2.63m/s)(0.0127m)}{1 \times 10^{-3} \frac{kg}{m \cdot s}}
\]

\[
\text{Re} = 33,334
\]

**Equation 3.23**

Where $\rho$ is the fluid density, $\mu$ is the dynamic viscosity, $V$ is the fluid velocity and $D$ the diameter of the tubing. From Equation (3.23), the Reynolds number is greater than the threshold value of 2300 [29]. Thus, at the pump’s outlet, water flow is turbulent. The value of the Reynolds number will be used to calculate the loss coefficient in the respective portion of the pipe.

ii) Find $K$ at 90 deg. Elbow with diameter of ¼”
Where $R$ is the radius of curvature and ‘d’ is the diameter of the elbow. The elbow radius curvature is approximately 1 cm.

iii) Contraction from $\frac{1}{2}”$ (D) to $\frac{1}{4}”$ (d) diameters going from the pump outlet to the tubing

$$K = 0.388\alpha\left(\frac{R}{d}\right)^{0.84} \Re_D^{-0.17}$$

$$\alpha = 0.95 + 4.42\left(\frac{R}{d}\right)^{-1.96}$$

$$K = 0.388\left[0.95 + 4.42\left(\frac{R}{d}\right)^{-1.96}\right]\left(\frac{R}{d}\right)^{0.84} \Re_D^{-0.17}$$

$$K = 0.388\left[0.95 + 4.42\left(\frac{R}{d}\right)^{-1.96}\right]\left(1.575\right)^{0.84} \left(33.334\right)^{-0.17}$$

$$K = 0.267$$

Equation 3.24)

iv) Expansion from $\frac{1}{4}”$ to $\frac{3}{8}”$ diameters going from the tubing to the heat exchanger

$$K \approx 0.42\left(1 - \frac{d^2}{D^2}\right)$$

$$K \approx 0.42\left(1 - \left(\frac{1}{4}\right)^2\right)$$

$$K \approx 0.315$$

Equation (3.25)

iv) Expansion from $\frac{1}{4}”$ to $\frac{3}{8}”$ diameters going from the tubing to the heat exchanger

$$K = \left(1 - \frac{d^2}{D^2}\right)^2$$

$$K = \left(1 - \left(\frac{1/4}{3/8}\right)^2\right)^2$$

$$K = 0.3086$$

Equation 3.26)
v) Heat exchanger grooves: ten 90 deg. turns:

find the Reynolds number in the heat exchanger:

\[ Q_{pump} = Q_{HeatExchanger} \]
\[ A_{pump} \frac{V_{pump}}{V_{HE}} = A_{HE} V_{HE} \]

\[ \frac{\pi}{4} (0.0127m)^2 (2.63m/s) = \left(4.35 \times 10^{-3} m\right)\left(11 \times 10^{-3} m\right) V_{HE} \]
\[ V_{HE} = 6.96m/s \]

\[ \therefore \text{Re} = \frac{\rho V D_H}{\mu} \]
\[ \text{Re} = \frac{(998kg / m^3)(6.96m/s)(4 \left(11 \times 10^{-3}\right)\left(4.35 \times 10^{-3}\right)}}{(2 \times 4.35 \times 10^{-3})} \]
\[ \text{Re} = \frac{1 \times 10^{-3} kg}{m \cdot s} \]
\[ \text{Re} = 43,322 \]

Where \( D_H \) is known as the hydraulic diameter which is used for non-circular pipes. It is defined as the ratio between the cross-sectional area and the wetted perimeter (i.e. the perimeter that is in contact with water). In other words \( D_H = \frac{4A}{P} \) [29].

\[ K = 0.388 \alpha \left( \frac{R}{d} \right)^{0.84} \text{Re}_D^{-0.17} \]
\[ \alpha = 0.95 + 4.42 \left( \frac{R}{d} \right)^{-1.96} \]
\[ K = 0.388 \left[ 0.95 + 4.42 \left( 0.7195 \right)^{-1.96} \right] \left( 0.7195 \right)^{0.84} \text{Re}_D^{-0.17} \]
\[ K = 0.388 \left[ 0.95 + 4.42 \left( 1.575 \right)^{-1.96} \right] \left( 1.575 \right)^{0.84} \left( 43321.6 \right)^{-0.17} \]
\[ K = 0.45 / \text{turn} \]

Equation (3.27)
Thus, for 10 turns, the loss coefficient for the heat exchanger is equal to 4.5. It is important to note that the Reynolds number of 43,341.6 used above may not be accurate as it was approximated. However, if the flow was laminar, which results in a Reynolds number below 2300 [29], the value of the coefficient does not change significantly. This is due to the small value of the exponent on Re_D which dampens the effect of the Reynolds number on the loss coefficient.

vi) Contraction from 3/8” to 5/16” diameter: Heat Exchanger to the external tubing

\[ K = 0.42 \left( 1 - \frac{d^5}{D^2} \right) \]

\[ K = 0.42 \left( 1 - \left( \frac{5/16}{3/8} \right)^2 \right) \]

\[ K = 0.1283 \]

vii) Expansion from 5/16” diameter to 3.5”: tubing to the reservoir

For all submerged exits, K=1 due to viscous dissipation causing the fluid to lose its velocity head [29].

viii) Contraction: 3/8” to 5/16”: reservoir outlet to the tubing

\[ K = 0.42 \left( 1 - \frac{d^5}{D^2} \right) \]

\[ K = 0.42 \left( 1 - \left( \frac{5/16}{3/8} \right)^2 \right) \]

\[ K = 0.1283 \]

ix) Contraction 5/16” to ¼”: tubing to smaller tubing
\[ K = 0.42 \left( 1 - \frac{d_s}{D^2} \right) \]

\[ K = 0.42 \left( 1 - \left( \frac{1/4}{5/16} \right)^2 \right) \]

\[ K \approx 0.1512 \]

x) Contraction: \(\frac{3}{4}''\) to \(1/8''\): tubing to smaller tubing

\[ K = 0.42 \left( 1 - \frac{d_s}{D^2} \right) \]

\[ K = 0.42 \left( 1 - \left( \frac{1/8}{1/4} \right)^2 \right) \]

\[ K \approx 0.315 \]

xi) Contraction \(1/8''\) to \(0.08''\): tubing to smaller tubing

\[ K = 0.42 \left( 1 - \frac{d_s}{D^2} \right) \]

\[ K = 0.42 \left( 1 - \left( \frac{0.08}{1/8} \right)^2 \right) \]

\[ K \approx 0.2479 \]

xii) Expansion \(0.08''\) to \(0.787''\): tubing to balloon

\[ K = \left( 1 - \frac{d_s}{D^2} \right)^2 \]

\[ K = \left( 1 - \left( \frac{0.08^2}{0.787^2} \right) \right)^2 \]

\[ K \approx 0.979 \]
xiii) Contraction 0.787” to 0.08”: Balloon to tubing

\[ K \approx 0.42 \left( 1 - \frac{d_s}{D^2} \right) \]

\[ K \approx 0.42 \left( 1 - \frac{0.08}{0.787} \right)^2 \]

\[ K \approx 0.4160 \]

xiv) Expansion 0.08” to 1/8”: tubing to larger tubing

\[ K = \left( 1 - \frac{d_s}{D^2} \right)^2 \]

\[ K = \left( 1 - \frac{0.08^2}{0.125^2} \right)^2 \]

\[ K \approx 0.348 \]

xv) Expansion 1/8” to ¼”: tubing to larger tubing

\[ K = \left( 1 - \frac{d_s}{D^2} \right)^2 \]

\[ K = \left( 1 - \frac{1/8^2}{1/4^2} \right)^2 \]

\[ K \approx 0.5625 \]

xvi) Expansion from 1/4” to 3/8”: tubing to larger tubing

\[ K = \left( 1 - \frac{d_s}{D^2} \right)^2 \]

\[ K = \left( 1 - \frac{1/4^2}{(3/8)^2} \right)^2 \]

\[ K \approx 0.3086 \]
xvii) Expansion from 3/8" to ½" tubing to pump inlet

\[
K = \left( 1 - \frac{d_s}{D^2} \right)^2
\]

\[
K = \left( 1 - \left( \frac{3/8}{1/2} \right)^2 \right)^2
\]

\[K \approx 0.1914\]

Therefore, the total loss coefficient from all expansions and contractions is approximately 10.4. The importance of this lies in that it enables us to calculate the water flow rate in the balloon, which in turn will allow us to further study the heat transfer phenomenon in the heat exchanger. The flow rate in the balloon was determined using the Bernoulli equation. This equation is used to describe internal flow as it follows a water molecule traveling along a steady, non-turbulent streamline [29]. In other words, the equation models the conservation of energy within fluid flow.

\[
P + \frac{1}{2} \rho v^2 + \rho gh = C
\]

Equation 3.28)

Where the first term on the left is the pressure, the second term is the kinetic energy per unit volume and the third term is the potential energy per unit volume. Equation (3.28) suggests that along a single streamline of fluid, may also be used which takes into account energy losses. Equation (3.28) can be rearranged while incorporating loss terms to better study fluid flow within the piping system:
\[
\frac{P_1}{\rho g} + \frac{v_1^2}{2g} + h_i = \frac{P_2}{\rho g} + \frac{v_2^2}{2g} + h_2 + h_{\text{loss}}
\]

where...

\[
h_{\text{loss}} = \frac{v^2}{2g} \left( \frac{fL}{d} + \Sigma K \right)
\]

\[
\frac{P_1}{\rho g} + \frac{v_1^2}{2g} + h_i = \frac{P_2}{\rho g} + \frac{v_2^2}{2g} + h_2 + \frac{Q^2}{2g} \left( \frac{1}{A^2} \left( \frac{fL}{d} + \Sigma K \right) \right)
\]

**Equation (3.29)**

Where, \(P\) is the pressure, \(\rho\) is the fluid density, \(g\) is the acceleration due to gravity, \(h\) is the head, \(f\) is the friction factor which accounts for the roughness of the piping. Furthermore, \(L\) is the length of the pipe, \(d\) is the diameter, and \(A\) is the tube cross-sectional area. The final term on the far right accounts for the energy loss due to the minor losses. Thus, if all velocities are converted into flow rates, \(Q\), then it would be possible to calculate the flow rate in the piping system.

Several values were approximated in Bernoulli’s equations. The first was the pressure in the reservoir. The pressure was calculated by opening the reservoir to the atmospheric pressure, which allows the reservoir pressure at the water surface to be at 101325 Pascal. The reservoir was then sealed, and with a marker, the water level was marked on the reservoir walls. When the pump was activated, the water level increased by 5mm, which allows us to calculate the approximate water pressure using Equation (3.30):

\[
P = 101325 Pa + \rho gh
\]

\[
P = 101325 + (998 \text{ kg/m}^3)(9.81 \text{ m/s}^2)(-0.005 \text{m})
\]

\[
P = 101276 Pa
\]

**Equation 3.30**

This value of pressure will be taken as \(P_2\) while \(P_1\) will be assigned the value of the pump’s pressure. Thus, the flow rate may be found as follows:
\[
\frac{P_1}{\rho g} + \frac{v_1^2}{2g} + h_1 = \frac{P_2}{\rho g} + \frac{v_2^2}{2g} + h_2 + \frac{Q^2}{2g} \left( \frac{1}{A^2} \left( \frac{fL}{d} + \Sigma K \right) \right)
\]

\[
\Rightarrow \frac{3.5 \times 10^5 \text{Pa}}{(998 \text{ kg/m}^3)(9.81 \text{ m/s}^2)} + \frac{Q^2}{2(9.81 \text{ m/s}^2)} + 0m = \frac{101276 \text{ Pa}}{(998 \text{ kg/m}^3)(9.81 \text{ m/s}^2)} + \frac{0}{2g} + 0m + \frac{0}{2g} + 0m + \frac{0}{2g}
\]

\[
\left[ \left( \frac{4}{\pi(0.00635)^2} \right)^2 \left( \frac{0.0233(0.01m)}{0.00635m} + 0.267 \right) + \left( \frac{4}{\pi(0.009525)^2} \right)^2 \left( \frac{0.02(0.24m)}{0.00635m} + 0.315 \right) + \right]
\]

\[
\left( \frac{4}{\pi(0.009525)^2} \right)^2 \left( \frac{0.02(0.01m)}{0.009525m} + 0.3086 \right) + \left( \frac{1}{(11 \times 10^{-3} \text{ m})(4.35 \times 10^{-3} \text{ m})} \right)^2 \left( \frac{0.02(0.28m)}{0.00794m} + 0.1283 \right) +
\]

\[
\left( \frac{4}{\pi(0.009525)^2} \right)^2 \left( \frac{0.02(0.01m)}{0.009525m} + 0.1283 \right) + \left( \frac{4}{\pi(0.00794)^2} \right)^2 \left( \frac{0.02(0.01m)}{0.009525m} + 0.4 \right) +
\]

\[
\left[ \left( \frac{4}{\pi(0.00794)^2} \right)^2 \left( \frac{0.02(0.032m)}{0.00794m} + 0.1283 \right) + \left( \frac{4}{\pi(0.009525)^2} \right)^2 \left( \frac{0.02(0.16m)}{0.003175m} + 0.315 \right) + \right]
\]

\[
\left( \frac{4}{\pi(0.009525)^2} \right)^2 \left( \frac{0.02(0.24m)}{0.009525m} + 0.2479 \right) + \left( \frac{4}{\pi(0.002032)^2} \right)^2 \left( \frac{0.02(0.13m)}{0.02m} + 0.979 \right) +
\]

\[
\left( \frac{4}{\pi(0.002032)^2} \right)^2 \left( \frac{0.02(0.24m)}{0.002032m} + 0.416 \right) + \left( \frac{4}{\pi(0.002032)^2} \right)^2 \left( \frac{0.02(0.18m)}{0.003175m} + 0.348 \right) +
\]

\[
\left( \frac{4}{\pi(0.00635)^2} \right)^2 \left( \frac{0.02(0.07m)}{0.00635m} + 0.5625 \right) + \left( \frac{4}{\pi(0.009525)^2} \right)^2 \left( \frac{0.02(0.02m)}{0.009525m} + 0.3086 \right) +
\]

\[
\left( \frac{4}{\pi(0.0127)^2} \right)^2 \left( \frac{0.02(0.01m)}{0.0127m} + 0.1914 \right)
\]

\[
\Rightarrow 35.75 + \frac{Q^2}{3.15 \times 10^7} = 10.347 + \frac{Q^2}{2 \times 9.81 \text{ m/s}^2}
\]

\[
[3.02 \times 10^8 + 1.1 \times 10^9 + 6.4 \times 10^7 + 0.7739 + 3.4 \times 10^8 + 2.0 \times 10^8 + 8.3 \times 10^7 + 8.5 \times 10^7 + 3.1 \times 10^8 + 2.11 \times 10^{10}
\]

\[
+ 2.27 \times 10^{12} + 1.1 \times 10^7 + 2.29 \times 10^{12} + 2.36 \times 10^{10} +
\]

\[
7.8 \times 10^8 + 6.9 \times 10^7 + 1.29 \times 10^7]
\]
This low speed corresponds to a Reynolds number of 2027 which is in the laminar region. This justifies the use of the Bernoulli equation which is used for steady non-turbulent flow.

The value of the flow rate will allow us to approximate the heat transfer phenomenon that occurs in the heat exchanger. Converting the volumetric flow rate into a fluid speed within the heat exchanger, the rate is found to be 21.3 cm/s.

In order to determine whether the flow in the heat exchanger is developed or not, the thermal entry length for laminar flow shall be calculated:

\[
L_{t,Laminar} \approx 0.05 \text{Re Pr } D_{\text{hydraulic}}
\]

Equation 3.31)

Where Re is the Reynolds number, Pr is Prandtl number which is, like Reynolds number, a dimensionless ratio between dissipation and thermal conduction [29]. This constant is significant in heat convection scenarios. The final term is the hydraulic diameter which was discussed earlier. The concept of thermal entry length is a model that follows the development of a thermal boundary layer. If fluid of uniform temperature flows into a tube with heated walls, the fluid particles in contact with the walls will assume the temperature of the walls; this will cause convection heat transfer, in which a thermal boundary layer would develop gradually downstream until it reaches the center of the pipe. The length of pipe in which the thermal boundary region has just reached the center is called the thermal entry region, \(L_{t,Laminar}\). Any length of tubing found within this region, the fluid is said to be developing [29]. Thus, to determine the heat transfer in the heat exchanger, we must first find whether the heat exchanger has fully developed flow or not. From Equation (3.31) we can find the entry length:
\[ L_{t,La \text{ min } ar} = 0.05 \left( \frac{\rho V D_{\text{hydraulic}}}{\mu} \right) \left( \frac{C_p \mu}{k} \right) D_{\text{hydraulic}} \]

\[ \Rightarrow L_{t,La \text{ min } ar} = 0.05 \left( \frac{(998 \, \text{kg/m}^3)(0.213 \, \text{m/s})(0.01247 \, \text{m})}{\mu} \right) \left( \frac{4182 \, \frac{J}{\text{kg} \cdot \text{C}}}{k} \right) \left( 1 \times 10^{-3} \, \frac{\text{kg}}{\text{m} \cdot \text{s}} \right) (0.01247 \, \text{m}) \]

\[ L_{t,La \text{ min } ar} = 2.88 \, \text{m} \]

Given that the heat exchanger is less than 2.88 meters in length, then it follows that the fluid flow is developing within the heat exchanger. To determine the effectiveness of cooling within the heat exchanger, the Nusselt number will be determined. This is a dimensionless value which indicates the effectiveness of heat convection through the fluid as compared to heat conduction through the same fluid. In other words, it’s a ratio between the heat transfer due to convection to the heat transfer due to conduction [29].

Given that heat exchanger is noncircular, the Nusselt number was determined from a table of values which matches the ratio of a duct’s width to depth to a corresponding Nusselt number [29]. The graph of these ratios is found in Figure 3.19.

**Figure 3.19** The variation of Nusselt number with the duct's cross-sectional dimension. The vertical axis is Nusselt number and the horizontal axis is the ratio of the channel width to depth.
Using the equation for the line of best fit from Figure 3.19, which has a $R^2$ value of 0.977, it can be calculated that at a ratio of 4.35mm/11mm, the Nusselt number is 2.87. This demonstrates that convective cooling is more effective than conductive cooling through the fluid. To better understand the effectiveness of the heat exchanger, a dimensionless parameter termed the number of transfer units may be calculated. This requires the value of the heat transfer coefficient for convective heat transfer. That may be found as follows:

$$
Nu = \frac{q_{\text{convetion}}}{q_{\text{conduction}}} = \frac{hD_H}{k}
$$

$$
h = \frac{k \cdot Nu}{D_H}
$$

$$
h = \frac{\left(0.598 \frac{W}{m \cdot ^\circ C}\right)(2.87)}{4 \left(\frac{(11 \times 10^{-3} \text{m})(4.35 \times 10^{-3} \text{m})}{2(11 \times 10^{-3} \text{m} + 4.35 \times 10^{-3} \text{m})}\right)}
$$

$$
h = 275.28 \frac{W}{m^2 \cdot ^\circ C}
$$

**Equation (3.32)**

Where $q$ is the heat transfer rate, $h$ is the convection coefficient, $D_H$ is their hydraulic diameter and $k$ is the thermal conductivity of the liquid. Finally, the number of transfer units (NTU) can be found from:

$$
NTU = \frac{hA_{\text{surface}}}{mC_p}
$$

$$
NTU = \left(275.28 \frac{W}{m^2 \cdot ^\circ C}\right)\left(0.006324 m^2\right)
$$

$$
NTU = \left(998 \frac{kg}{m^3}\right)\left(1.019 \times 10^{-5} \frac{m^3}{s}\right)\left(4182 \frac{J}{kg \cdot ^\circ C}\right)
$$

$$
NTU = 0.04
$$

**Equation 3.33**
Having a NTU value that is greater than 5 indicates that the outlet temperature of the fluid will be equal to the surface temperature of the heat exchanger regardless of the temperature of the fluid at the inlet. Thus, the small value of NTU indicates that the heat transfer will continue to rise along the tube in the direction of the flow [29]. This phenomenon is shown in Figure 3.18 where the coolant temperature gradually increases along the internal windings of the heat exchanger. Since the channel in the heat exchanger has a combined length of 24 cm, then it will require several cycles in order for the water to reach the surface temperature of the heat exchanger. Assuming we want a maximum temperature drop of \(2\,^\circ C\), then calculating the rate at which heat is pumped out of the system, we get:

\[
q = m C_p (T_{exit} - T_{entrance})
\]

\[
q = \left(\rho \dot{Q}\right) C_p (T_{exit} - T_{entrance})
\]

\[
q = \left(998 \frac{kg}{m^3} \times 1.019 \times 10^{-5} \frac{m^3}{s}\right)(4182 \frac{J}{kg \cdot ^\circ C})(2\,^\circ C)
\]

\[
q = 85W
\]

Equation (3.34)

Thus, in order to achieve a heat transfer rate of 85W a Peltier device with a rating of 85W or greater should be used. However, due to losses, such as internal heating, it is preferable to use a greater rating. As a result, the Peltier device that was used had a total power rating of 89W.

3.3.6 Preliminary Thermal Regulation Experiments

After constructing a completed cooling device with all the components, preliminary cooling tests were made on phantoms heated to, and maintained at various temperatures.

Phantoms were made from in water mixed with 1% agar in 1 L beakers. The solution was heated to a boil at which point the beaker was removed from the heat and placed into an ice-filled bucket. As the solution cooled, an inflated endorectal balloon was inserted and held at the center of the beaker until the solution solidified. The gel was then placed into the cold room and left over night. The setup was similar to Figure 3.20.
After complete solidification, the gel-filled beakers were placed into a water bath until the core temperature of the gel equilibrated with the water bath temperature. This was measured using the thermistor connected to the cooling unit.

Two variables can potentially change the rate of cooling of the rectal wall: water flow rate and the starting temperature of the gel. The purpose of this study was to determine the rate at which the water temperature within the balloon dropped as well as to determine the lowest possible temperature within the balloon at different gel temperatures and flow rates. Each of these two variables was studied while keeping the other constant. The first scenario involved changing the water temperature of the bath while keeping the flow rate constant and the second scenario was to keep the gel temperature constant to study the effects of flow rate. A summary of the gel temperatures and respective flow rates are listed in Table 3.5.

Figure 3.20 Sample set up of phantom experiment.
Table 3.5 Temperature of gel core and water bath

<table>
<thead>
<tr>
<th>Bath/Gel Temperature</th>
<th>Pump Speed</th>
</tr>
</thead>
<tbody>
<tr>
<td>34°C</td>
<td>Max</td>
</tr>
<tr>
<td>37°C</td>
<td>Min, Medium &amp; Max</td>
</tr>
<tr>
<td>40°C</td>
<td>Max</td>
</tr>
<tr>
<td>48°C</td>
<td>Max</td>
</tr>
</tbody>
</table>

For each scenario, a single thermistor was placed at approximately 2-3mm away from the cooling balloon. The thermistors monitored the temperature at a distance of about 3 mm which represented the outer rectal wall. Tests were conducted for approximately 40 minutes. The first 2.5 minutes was simply watering flowing through the balloon without active cooling form the Peltier device (power at 0%). After that point, the Peltier device was activated and operated in PID mode. This was designed such that a target temperature of 15°C was used with a bandwidth of 10°C, that is, ±5°C about the set point temperature. The power was linearly proportional to the temperature difference from the set point within the bandwidth. However, outside the bandwidth maximum power was applied. At the 25th minute, maximum power was applied while PID was switched off. This was done for each run at different gel temperatures.
Chapter 4
Results

4 Results

There are four experiments that were run from which data was extracted. These are divided into two computer simulations, one from COSMOSXpress and the other from a treatment planning software. The other two were extracted from laboratory experiments on phantoms. This section will report on all four data sets from these areas.

4.1.1 COSMOSXpress Simulations

The simulations in COSMOSXpress were designed to determine the effect of fluid flow rate on the rate of cooling. As such several flow rates at different temperatures were simulated. The results were to monitor the rectal wall temperature throughout the treatment due to the laser fiber, which was placed into the prostate directly on top of the rectum.

In Figure 4.1, Figure 4.2, Figure 4.3 and Figure 4.4, the input temperatures of the fluid were maintained at 0°C, and the flow rate was altered. Results showed that the flow rate did not have any effect on the temperature of the rectal wall. The temperatures reached by each graph are solely affected by the temperature of the coolant. This is partly due to the fact that the convective heat transfer coefficient strongly depends on both the surface area across which the heat energy crosses, as well as the temperature difference between the fluid and the inner surface of the balloon. The temperature of the outer rectal wall experienced the greatest heat buildup as it was adjacent to the laser probe, while temperatures at the inner surface of the rectal wall experienced the lowest temperatures as it is in direct contact with the balloon. It is seen in Figure 4.2 and Figure 4.3 that the thermocouples farther away from the heat source experienced a delay. This was due to slow heat conduction from the heat source out to the thermocouples.
Figure 4.1 Temperature profile of the outer and inner rectal wall. The temperature of the coolant was kept constant at 0°C and the flow rates were altered.

Figure 4.2 Temperature of thermocouple at 2mm. The temperature of the coolant was kept constant at 0°C and the flow rates were altered.
Figure 4.3 Temperature of thermocouple at 5mm. The temperature of the coolant was kept constant at 0°C and the flow rates were altered.

Figure 4.4 Temperature of thermocouple at 10mm. The temperature of the coolant was kept constant at 0°C and the flow rates were altered.
Figure 4.5, Figure 4.6 and Figure 4.7 show the effect of fluid temperature on the surrounding tissue, at a constant flow rate of 5 g/s. From these figures, there is a clear difference in tissue temperature due to temperature of the fluid. As could be predicted, the inner rectal wall experienced the lowest temperatures compared to the rest of the rectum. This was due to the direct contact between the rectal balloon and the inner rectal wall. Rectal tissue temperatures rose between 5- 8°C with every 10°C rise at the inner rectal wall, achieving a minimum temperature of 16°C at the inner rectal wall for fluid temperatures of 0°C and a maximum temperature of 44°C for fluid temperatures of 37°C also at the inner rectal wall. Furthermore, an initial drop in inner rectal temperatures is observed. This drop is less pronounced for warmer coolant temperatures due to their similarity to physiological temperatures which leads to smaller temperature gradients. Also, these initial drops are due to the delay in heat conduction through the thickness of the rectal wall as the inner rectal wall first experiences the cold water temperatures of the balloon.

Figure 4.5 Temperature variation of the inner and outer rectal wall. The flow rate was kept constant at 5g/s and the fluid temperatures were altered.
The outer rectal wall temperature profile (Figure 4.5) reveals that over time the rectal temperature profile for each coolant temperature begins at the same initial temperature but then diverges away from each other. This results in an approximate difference of 4°C between consecutive temperature conditions. The graphs exhibit a state of constant kinetics. By extrapolating, it can be deduced that the temperature rises towards 87°C for coolant temperatures of 37°C and towards 67°C for fluid at 0°C. Furthermore, ablative temperatures of 55°C are reached within the first 15 seconds for all fluid temperatures due to the close proximity to the laser probe. As a result, it is not feasible to protect the outer rectal wall from ablation with the laser fiber sitting directly above the rectal wall. Therefore, laser positions should be maintained at the 2-5mm position range instead of directly atop the rectum.

Figure 4.6 Tissue temperature readings at 2mm and 5mm distance from laser fiber. The flow rate was held constant at 5g/s and the fluid temperatures were altered.
Figure 4.6 reveals the temperature profile for the thermocouples at distances of 2mm and 5mm from the laser source. As shown, the tissue temperature at 2 mm for each fluid temperature is identical for the first 5 seconds and then begins to diverge. This is due to the delay in heat conduction from the laser source to the 2mm thermocouple. Furthermore, the profiles diverge slowly way from each other. The divergence is also seen at 5mm away (Figure 4.6) and 10 mm away (Figure 4.7) from the laser fiber; however, it is more pronounced at thermocouples closer to the laser. The temperature at 10mm does not, however, exceed 38°C for the first minute at all fluid temperatures while the maximum temperature at the 2 mm position occurred above 62°C for water at 37°C and slightly above 57°C for water at 0°C, in which ablation is experienced at both water temperatures at 2mm.

![Thermocouple Temperature 10mm](image)

**Figure 4.7** Temperature profile for the thermocouple at 10mm. The flow rate was kept constant at 5g/s and the fluid temperatures were altered.
Figure 4.8, Figure 4.9, Figure 4.10, Figure 4.11 and Figure 4.12 show the temperature map of the cross section of the anatomy. It is seen that the coagulated region grows in an elliptical fashion where the long axis is the vertical direction. Although the rectum is seen to be near freezing point, it is unlikely to be at such a low temperature for two reasons: The first, the heat generation of the rectum and surrounding tissues as well as heat generated by the laser will cause the temperature of the rectum to be higher than that of the coolant. Secondly, COSMOSXpress may have placed several temperatures under one colour due to both the large temperature gradient generated within the anatomy and also the low colour resolution of the program.

Figure 4.8 $T_{\text{fluid}} = 0^\circ C$ and $Q = 0.005 \text{kg} / \text{s}$

Figure 4.9 $T_{\text{fluid}} = 10^\circ C$ and $Q = 0.005 \text{kg} / \text{s}$
Figure 4.10 $T_{\text{fluid}} = 20^\circ C$ and $Q = 0.005\text{kg/s}$

Figure 4.11 $T_{\text{fluid}} = 30^\circ C$ and $Q = 0.005\text{kg/s}$
Figure 4.12 $T_{\text{fluid}} = 37^\circ C$ and $Q = 0.005\text{kg/s}$

For the sake of simplicity, it shall be assumed that the temperature on the rectal inner and outer surface has reached steady state. Table 6 summarizes the temperatures of the inner and outer rectal wall for each fluid temperature regardless of flow rate, as flow rate does not influence the temperature of the tissue.
Table 4.1 Respective temperature of the inner and outer rectal wall for different fluid temperatures in the rectal balloon

<table>
<thead>
<tr>
<th></th>
<th>Temperature of fluid flowing in balloon</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0°C</td>
</tr>
<tr>
<td>Inner rectal wall</td>
<td>16</td>
</tr>
<tr>
<td>wall temperature</td>
<td>[˚C]</td>
</tr>
<tr>
<td>Outer rectal wall</td>
<td>63</td>
</tr>
<tr>
<td>wall temperature</td>
<td>[˚C]</td>
</tr>
</tbody>
</table>

Although these results would suggest that the flow rate does not affect the cooling ability of the fluid, future experiments showed that they do. The reason for this discrepancy may be due to the proximity of the laser fiber to the rectum causing excessive heating. Furthermore, these simulations were the first set of calculations done at the beginning of the project. Over time, higher power laser were used during laboratory experiments. These would cause inevitable damage to the rectal wall if the laser fiber was to be placed into the prostate, directly on top of the rectum. Furthermore, it was realized, after building the device, that reaching temperatures as low as 0˚C was difficult and required too much energy. It would be more practical to place the laser fiber higher up into the prostate away from the rectal wall, which will allow two effects. The first is that it allows for higher coolant temperatures. Secondly, it would allow for controlled ablation of the prostate tissue that lies between the rectal wall and the laser fiber. This is possible
at several previously determined water temperatures that would shape the thermal gradient such that the rectal wall remains partially unscathed. As such, these protective water temperatures would be determined using the treatment planning software.

4.1.2 Simulations from the Treatment Planning Program

The treatment planning software was utilized to predict the lesion sizes and thermal map of the prostate and surrounding tissue at various laser power levels and fiber distances from the outer rectal wall. Furthermore, it was necessary to calculate the required coolant temperature which serves to protect the outer rectal wall at 46˚C. In all simulated scenarios, the temperature of the outer rectal wall equilibrated at a single temperature of 46˚C. From Figure 4.13 it can be seen that for various laser powers and distances, the temperature at the outer rectal wall is safely maintained at or below 46˚C for the duration of the treatment.

![Temperature of Outer Rectal Wall](image)

Figure 4.13 The temperature of the outer rectal wall at several laser fiber positions and power levels. All scenarios experienced temperatures that converged towards 46˚C.
Figure 4.14 summarizes the water temperatures required to protect the outer prostate wall at specific laser powers and fiber distances. This allowed us to see which temperatures were required to maintain the outer rectal wall temperature below 46°C. This was important because the cooling device will be limited in terms of the lower limit of water temperature it will be able to reach and maintain. These scenarios will allow us to selectively choose the appropriate setting given the location of the cancer from the prostate as well as allow us to predict the size of the lesion that will be generated at a given power level and fiber distance, which is beneficial when the size of targeted cancer is known.

![Water Temperature Variation with Laser Fiber Distance from Rectal Wall and Power Level](image)

Figure 4.14 The relation between the required protective water temperature at specific laser powers and laser positions.
As shown in Figure 4.14, there is an underlying relation between the three parameters: laser proximity to the outer rectal wall, laser power and coolant temperatures. Specifically, the coolant temperature is directly proportional to the distance of the laser fiber from the outer rectal wall and indirectly proportional to the laser power used. Furthermore, the change in coolant temperatures per millimeter change in distance away from the rectal wall is summarized in table 4.2 which shows that there is an increasing trend in the slope with increasing laser power. There is one anomaly at 8W which shows a slight decrease in the slope. However, this may be due to the fact that the slope is based on two points of data, rather than the preferred three points.

Table 4.2 The relationship between wattage and change in water temperature per unit distance of the laser fiber from the outer rectal wall.

<table>
<thead>
<tr>
<th>Wattage</th>
<th>Slope = $\frac{\Delta Temperature}{\Delta distance}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>2W</td>
<td>2.1°C/mm</td>
</tr>
<tr>
<td>3W</td>
<td>3.4°C/mm</td>
</tr>
<tr>
<td>4W</td>
<td>4.4°C/mm</td>
</tr>
<tr>
<td>5W</td>
<td>5.6°C/mm</td>
</tr>
<tr>
<td>6W</td>
<td>6.3°C/mm</td>
</tr>
<tr>
<td>7W</td>
<td>6.5°C/mm</td>
</tr>
<tr>
<td>8W</td>
<td>6.0°C/mm</td>
</tr>
</tbody>
</table>

At each laser power and distance from the outer rectal wall, a temperature map was generated to monitor the cooling effects of the coolant on the shape and size of the lesions generated. Figure 4.15 is a sample temperature map generated during the simulation at 5 mm and 2W. Appendix A
contains thermal maps and thermal dose contours generated at maximum and minimum laser powers for each fiber position.

Figure 4.15 A temperature map generated from the treatment planning software. Simulation conditions were at a laser power of 2W and position of 5mm away from the rectal wall.

Figure 4.16, Figure 4.17, Figure 4.18, Figure 4.19, Figure 4.20 and Figure 4.21 shows the temperature profile of all power levels and laser fiber positions that were simulated.
Figure 4.16 Temperature profiles of the rectum and prostate at several power levels with laser position at 5mm from outer rectal wall.
Temperature Profile of Prostate with Laser Fiber at 6mm from Outer Rectal Wall

![Graph showing temperature profiles at different power levels with laser position at 6mm from outer rectal wall.]

Figure 4.17 Temperature profiles of the rectum and prostate at several power levels with laser position at 6mm from outer rectal wall.
Figure 4.18 Temperature profiles of the rectum and prostate at several power levels with laser position at 7mm from outer rectal wall.
Figure 4.19 Temperature profiles of the rectum and prostate at several power levels with laser position at 8mm from outer rectal wall.
Figure 4.20 Temperature profiles of the rectum and prostate at several power levels with laser position at 9mm from outer rectal wall.
Figure 4.21 Temperature profiles of the rectum and prostate at several power levels with laser position at 10mm from outer rectal wall.

The effectiveness of the coolant is shown by the asymmetry of each graph with increasing power and decreasing laser distances from the outer rectal wall. In other words, on either side of the laser fiber, points that are equidistance from the laser fiber are not equal in temperature along the array of thermistors (Figure 4.22). Thermistors on the upper side of the laser fiber experience higher temperatures than those closer to the cooled rectum. This is clearly shown in Figure 4.15 as the thermal contours extend farther above the laser fiber than those below. Thus, a greater volume above the laser of tissue experiences coagulation relative to tissue below the laser fiber. This serves to suggest, as shown in Figure 4.23, that the shape of the coagulated region is not exactly circular but oval in shape with the minor axis being along the array of thermistors.
Figure 4.22 Schematic of the setup used during the simulations. The white diagonal line is the array of thermistors measuring temperatures at increments of 1mm.
Furthermore, the steepness of the temperature profiles is dependent on the power and distance. The higher the power at relatively large distances from the outer rectal wall results in steep temperature profiles, with higher maximal temperatures that translate into higher coagulation rates. Furthermore, due to the asymmetrical shape of the temperature profiles, larger lesions are seen on the upper side of the laser fiber.

It was necessary to determine the effect of cooling on the volume of the coagulated lesion. The lesions were assumed to be ellipsoids, and therefore the appropriate equation for the volume of an ellipsoid was used. The volume of the coagulated region was found using three axes: the length of the coagulated region, which was taken to be the length of the diffuser at 1cm, and the major and minor axis of the coagulated region, which were measured using software.

Figure 4.23 Schematic of the coagulated region generated during the simulation. The laser power is at 2W and located 5mm from the outer rectal wall.
measurement tools. Figure 4.25 summarizes the results for each power level and laser distance from the rectal wall.

![Volume of Coagulated Ellipsoid at Various Probe Distances](image)

**Figure 4.24 Volume of coagulated lesion at specific laser powers and positions from outer rectal wall.**

complimentary evidence to that suggested in Figure 4.13, that is, that the outer rectal boundary is maintained at 46°C. However there are some discrepancies with the data. As seen in Figure 4.16, Figure 4.17, Figure 4.18, Figure 4.19, Figure 4.20 and Figure 4.21, the temperatures at the outer rectal all are not exactly at 46°C, due to several discrepancies which is inherent in the data. Firstly, the thermistors are placed at exactly 1 mm increments from one another. The first thermistor location is approximated manually as the thermistors within the simulation program are quite large, and so the first thermistor located at the rectal wall may not be exactly on the rectal wall but may be off by approximately 0.25 mm at most. This may cause a small error in
the exact temperature at the desired location. Another reason for the discrepancy is that the
temperature of the coolant required to sufficiently cool and thus protect the outer rectal wall was
found by manual trial and error. That is, a coolant temperature was guessed and the simulation
was ran, if need be, to determine whether the rectal wall temperature was at 46°C. If not, the
coolant temperature was adjusted in the next trial such that the outer rectal wall temperature
converged towards 46°C. This was an inaccurate method because the y-axis (temperature) scale
of the output data from the simulation was crude due to the large scaling that the software pre-
set. Therefore, the temperature of the outer rectal wall was usually approximated. Furthermore,
in some cases, it was required that we vary the temperature of the fluid by fractions of a
temperature until the 46°C level was approximately reached. This would prove to be too tedious
and time consuming. This was judged to be unnecessary, as the calculated temperatures are
simply approximations of physical phenomenon.

4.1.3 Phantom Experiments

The purpose of these experiments was to validate the simulations from COSMOSXpress, as well
as test out the lower limit of the cooling device. Figure 4.26, Figure 4.27, Figure 4.28, Figure
4.29, Figure 4.30, and Figure 4.31 summarize the results of the test in which the coolant
temperature and flow rate was varied. It was found that the lower temperature limit of the device,
which has a single Peltier device, was 15°C; however, this temperature was reached over several
minutes. This is due to the water’s high heat capacity, which is about 4.184 J/g·K.
Figure 4.25 Temperature profiles of the water reservoir and outer rectal wall. The thermistor was placed 3 mm away from balloon. Phantom initial temperatures were 37°C with a maximum pump flow rate. The minimum coolant temperature reached was 17°C.

Figure 4.26 Temperature profiles of the water reservoir and outer rectal wall. The thermistor was placed 3 mm away from balloon. Phantom initial temperatures were 40°C with a maximum pump flow rate. The minimum coolant temperature reached was 18°C.
Figure 4.27 Temperature profiles of the water reservoir and outer rectal wall. The thermistor was placed 3 mm away from balloon. Phantom initial temperatures were 34°C with a maximum pump flow rate. The minimum coolant temperature reached was 19°C.

Figure 4.28 Temperature profiles of the water reservoir and outer rectal wall. The thermistor was placed 3 mm away from balloon. Phantom initial temperatures were 48°C with a maximum pump flow rate. The minimum coolant temperature reached was 19°C.
Figure 4.29 Temperature profiles of the water reservoir and outer rectal wall. The thermistor was placed 3 mm away from balloon. Phantom initial temperatures were 37°C with a minimum pump flow rate. The minimum coolant temperature reached was 14°C.

Figure 4.30 Temperature profiles of the water reservoir and outer rectal wall. The thermistor was placed 3 mm away from balloon. Phantom initial temperatures were 37°C with an intermediate pump flow rate. The minimum coolant temperature reached was 15°C.
The cooling device was properly functioning. The reservoir, or balloon, water temperature was initially at 20-25˚C, which is the expected initial water temperature before cooling begins. The balloon at time zero was placed into the phantom and the cooling device power was initially set to zero for 2.5 minutes to see the behavior of the balloon water temperature when the Peltier device was not activated. As seen in Figure 4.26, Figure 4.27, Figure 4.28, Figure 4.29, Figure 4.30 and Figure 4.31, during the initial 2.5 min the water temperature in the balloon increased as it was being heated by the warm phantom. At the 2.5 min mark, the power was applied in PID mode. At this point, the temperature of the coolant began to drop in what seems like an exponential decay. This can be explained using the number of transfer units, which is the seen in Equation (3.33). Given that the phantom material adjacent to the balloon slowly cools down, we can assume that it is approximately constant. Furthermore, we can also assume that the balloon wall temperature is equal to that of the adjacent phantom material. As a result, Newton’s law of cooling, which considers the temperature difference between the surface of the balloon and the mean fluid temperature, which may increase due to the heat transfer to the fluid, is given as

\[ Q = hA_s(T_s - T_m). \]

Equation (4.1)

This can be equated to the conservation of energy equation for a differential control volume

\[ h(T_s - T_m)dA_s = mC_pdT_m \]

Equation (4.2)

Where \( dT_m \) and \( dA_s \) are the differential temperature of the fluid and balloon surface area, respectively. Expressing the differential area as \( dA_s = pdx \), where \( p \) is the perimeter of the balloon, and \( dT_m = -d(T_s - T_m) \), and given that the surface areas of the balloon is constant, we can rewrite Equation (4.2) as
\[
\frac{d(T_s - T_m)}{T_s - T_m} = -\frac{hp}{m C_p} dx
\]

**Equation (4.3)**

Integrating and rearranging from the entrance of the balloon \((x=0)\) to the exit \((x=L)\), we get

\[
\ln \frac{T_s - T_e}{T_s - T_i} = -\frac{hA_s}{m C_p}
\]

**Equation (4.4)**

Taking the exponential of each side and solving for the exit temperature at \(x=L\) we get [25]:

\[
T_e = T_s - (T_s - T_i) \exp\left(-\frac{hA_s}{m C_p}\right)
\]

**Equation (4.5)**

As such, Equation (4.5) may be used to model the cooling of the water through the heat exchange. As shown in Figure 4.26, Figure 4.27, Figure 4.28, Figure 4.29, Figure 4.30 and Figure 4.31 the PID mode equilibrates at a certain temperature at which the rate of heat transfer from the phantom to the coolant is equal to that of the heat exchanger with the Peltier device. However, during the PID mode, full cooling power was not reached as the closer the coolant temperature is to the set temperature, the lower the power output to the cooling device.

After the 25th minute, the PID mode was switched off and full power was instigated. As shown, the cooling power continues to drop. From the treatment planning software, the temperature required to keep the rectum protected from thermal damage is to be held constant. Having a PID controlled device may delay or prevent the arrival of the necessary cooling temperature. As such, it would be beneficial to either narrow the bandwidth or to apply full power throughout the treatment.
The second aspect that was tested was the affect of flow rate on cooling. From Figure 4.26, Figure 4.27, Figure 4.28, Figure 4.29, Figure 4.30 and Figure 4.31 the flow rate was altered from maximum flow rate, to an intermediate flow rate and finally to a minimum flow rate. It was found that reducing the flow rate increased the cooling rate causing the temperature of the coolant to drop relatively more quickly. Equation (4.5) can explain this phenomenon for internal forced convection. An ideal heat exchanger is one that causes the exit water temperature to be equal to the surface temperature of the heat exchanger. Practically, this may not be possible, and so, in a more realistic sense, it would be desired to have the exit water temperature to be as close as possible to the surface temperature. This requires that the term on the far right of Equation (4.5) be as small as possible. One possible method to do so is to reduce the mass flow rate, $m$. This causes the exponential term to increase, thereby reducing the term on the far right of the equation, and allows the exit temperature to approach the surface temperature of the heat exchanger. Therefore, future phantom experiments will set the flow rate in the heat exchanger to minimum flow to allow for maximum heat transfer between the water and the walls of the heat exchanger.
5 Conclusion and Future Work

It has been shown that the temperature of the coolant during ILTT plays a primary role in cooling the rectum. Secondary to this effect is that the flow rate plays an increasing role as the laser fiber distance from the cooled rectum increases; however, at extremely close fiber proximities to the rectal wall, the affect of flow rate is relatively nonexistent. Furthermore, the volume and shape of the lesion is affected by the proximity of the laser probe to the cooled rectum. Firstly, the shape of the lesion resembles an asymmetrical ellipsoid where the shape slightly flattens near the rectal wall. This follows the fact that more tissue above the laser probe is coagulated. Resultantly, non-cancerous tissue may be ablated in the process at the expense of rectal protection, which is not a concern. The volume of the lesion, which was approximated as an ellipsoid, is directly proportional to the proximity of the laser fiber to the rectal wall, the laser power, and the coolant temperature. Furthermore, the required coolant temperature which protects the rectal wall is dependent on both the laser position and power. Therefore, rectal cooling, if implemented during ILTT, should be incorporated into during treatment planning process.

The cooling unit has the ability of reducing the temperature of the water to 15°C and maintaining it at this temperature. However, due to the heat capacity of the water used, as well as the limitation of the Peltier device in terms of the rate of thermal energy pumped out of the coolant, the water temperature would only reach these temperatures after several minutes. However, in order to optimize the cooling effect, it is required that the heat be dissipated from the hot side more effectively than using air, which is an insulator. Furthermore, air-cooling using large fans may not be suitable for operating room conditions. Therefore, more effective cooling methods should be used that alleviate the noise such as using water instead of air to cool the hot side of the Peltier device. On the other hand, a commercial cooling unit may be incorporated, such as that from Advantage Engineering [36], to cool the coolant to lower temperatures.
In order to confirm the results predicted by the treatment planning software, the next step is to test the cooling device on optical phantoms that have similar optical and thermal properties as prostate tissue, followed by *in vivo* tests. Here, thermal maps as well as lesion sizes and shapes can be monitored using MR thermometry which can monitor the tissue temperature and thereby predict the lesion size in real-time. This same technology may be used during clinical trials. The results from these future experimental settings should not only compare the efficacy of the cooling balloon on the lesion shape and size on a qualitative basis, but also on a quantitatively basis by using a merit function.
References


Figure A.1 Temperature map at a laser fiber position of 5mm from the outer rectal wall, and a power of 2W.

Figure A.2 Lesion size at a laser fiber position of 5mm from the outer rectal wall, and a power of 2W.
Figure A. 3 Temperature map at a laser fiber position of 5mm from the outer rectal wall, and a power of 5W.

Figure A. 4 Lesion size at a laser fiber position of 5mm from the outer rectal wall, and a power of 5W.
Figure A. 5 Temperature map at a laser fiber position of 6mm from the outer rectal wall, and a power of 2W.

Figure A. 6 Lesion size at a laser fiber position of 6mm from the outer rectal wall, and a power of 2W.
Figure A. 7 Temperature map at a laser fiber position of 6mm from the outer rectal wall, and a power of 5W.

Figure A. 8 Lesion size at a laser fiber position of 6mm from the outer rectal wall, and a power of 5W.
Figure A. 9 Temperature map at a laser fiber position of 7mm from the outer rectal wall, and a power of 2W.

Figure A. 10 Lesion size at a laser fiber position of 7mm from the outer rectal wall, and a power of 2W.
Figure A. 11 Temperature map at a laser fiber position of 7mm from the outer rectal wall, and a power of 6W.

Figure A. 12 Lesion size at a laser fiber position of 7mm from the outer rectal wall, and a power of 6W.
Figure A. 13 Temperature map at a laser fiber position of 8mm from the outer rectal wall, and a power of 2W.

Figure A. 14 Lesion size at a laser fiber position of 8mm from the outer rectal wall, and a power of 2W.
Figure A. 15 Temperature map at a laser fiber position of 8mm from the outer rectal wall, and a power of 7W.

Figure A. 16 Lesion size at a laser fiber position of 8mm from the outer rectal wall, and a power of 7W.
Figure A. 17 Temperature map at a laser fiber position of 9mm from the outer rectal wall, and a power of 2W.

Figure A. 18 Lesion size at a laser fiber position of 9mm from the outer rectal wall, and a power of 2W.
Figure A. 19 Temperature map at a laser fiber position of 9mm from the outer rectal wall, and a power of 8W.

Figure A. 20 Lesion size at a laser fiber position of 9mm from the outer rectal wall, and a power of 8W.
Figure A. 21 Temperature map at a laser fiber position of 10mm from the outer rectal wall, and a power of 2W.

Figure A. 22 Lesion size at a laser fiber position of 10mm from the outer rectal wall, and a power of 2W.
Figure A. 23 Temperature map at a laser fiber position of 10mm from the outer rectal wall, and a power of 10W.

Figure A. 24 Lesion size at a laser fiber position of 10mm from the outer rectal wall, and a power of 10W.