COMPARISON OF FLUENCE RATE DISTRIBUTIONS MADE BY SIDE-FIRING FIBERS IN AN OPTICAL PHANTOM

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ABSTRACT
Side-firing fibers are used to provide coagulative therapy to the urologic tract. These fibers use different optical technologies to deflect the beam transverse to the fibers’ optical axis. This produces emitted beams which differ in both beam direction and divergence angles. The relative optical performance of 13 fibers was studied in an optical phantom suspension. The fluence rate distribution created by each side-firing fiber was determined. The fluence rate distribution accounts for both the direct and spurious beams emitted from side-firing fibers as well as the light scattering produced by the target tissue. Based upon limited clinical dosimetry studies, the relative fluence rate distribution appears to indicate general exposure conditions for the evaluated fibers. © 1999 Society of Photo-Optical Instrumentation Engineers. [S1083-3668(99)00103-3]

Keywords fluence rate distributions; optical phantom; side-firing fiber; exposure levels.

1 INTRODUCTION
The side-firing fiber has been used in urologic surgery to treat benign prostatic hyperplasia (BPH).1–9 This type of optical fiber emits laser energy transverse to the fibers’ optical axis. This is accomplished with reflective and/or refractive technologies. Some designs use a mirror to deflect the beam transverse to the fibers’ axis, other designs use total internal reflection (TIR) and cleave the distal end of the fiber at its critical angle.10 The fiber is advanced through the working channel of a cystoscope to the enlarged prostate where coagulative therapy can be applied under visual control. This procedure has been referred to as visual laser ablation of the prostate (VLAP). The Nd:Yttrium–Aluminum–Garnet (YAG) laser operating at 1064 nm is usually the preferred source of therapeutic light, although other near infrared wavelengths are being investigated.11 Most of the side-firing fibers are indicated for ablation and hemostasis of urologic conditions. All are required to be used with a Nd:YAG laser that has been approved for urologic use. One manufacturer recommends laser launch conditions, most do not. There are a number of side-firing fibers that are available to the clinician, each one differing in nominal emitted beam characteristics, such as, the emitted beam angle (38°–105°) and its divergence (10.5°–50°). Some manufacturers provide this information in the device labeling, others do not. Attempts to determine how these characteristics may influence the coagulation process have been performed by evaluating these nominal optical characteristics.12,13 In addition, analytic optical analysis of side-firing fibers which deflect the beam by refraction (TIR) has also been performed.10,14

Efforts to understand the clinical dosimetry of the coagulation procedure have resulted in studies that use analytic and/or laboratory techniques. Analytic modeling which accounts for the dynamic physical changes to optical and thermal coefficients as well as heat dissipation due to blood flow has been performed with nonlinear finite element analysis.15 Numerous laboratory and clinical studies have determined the volume of coagulated tissue in canine and human prostates.16–23 A tissue and gel phantom was used to evaluate the thermal response of side-firing fibers.24 These studies were usually performed with one brand of side-firing fiber. Also, further laboratory work has shown that the optical transmission of side-firing fibers under simulated clinical conditions deteriorates with use.25

Historically, phantoms have been used to evaluate and monitor the performance of medical devices. As an example, phantoms are used in the performance evaluation of hyperthermia equipment.26–28 The specific absorption rate (SAR) of radio frequency (rf) and microwave hyperthermia equipment is measured using a phantom which is tissue equivalent at the frequency of the applied source. The SAR, along with data from other tests, such as, radiation leakage measure-

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ments, temperature scale calibrations, and visual inspections of the equipment, is used to ensure proper operation of the equipment, and provide quality control procedures which improve the safety, effectiveness, and reproducibility of hyperthermia treatments. Phantoms have also been used to evaluate surgical laser performance. A number of these phantoms can be found in the local supermarket, and include such items as egg white, beef steak, liver, chicken breast, gelatin, skim milk, and potatoes. In the laboratory, agar, Intralipid, polyimide, polystyrene microspheres, porcine myocardium, and bovine corneas have been used. The optical properties of some of these phantom materials have been determined. With the proper selection of absorbing and scattering materials, liquid optical phantoms have the potential for simulating the light distributions found in specific tissues.

Optical evaluations in liquid phantoms allow for the easy placement and movement of fluence detecting fibers. Solid phantoms are usually subjected to laser exposures and the resulting damage is examined microscopically. Temperature distributions resulting from laser exposures in solid phantoms have also been measured. In phantoms, optical or microwave, heat dissipation via blood flow is notably absent. These phantoms allow for the comparison of device performance under simulated clinical conditions of use, however, they do not predict, per se, actual clinical performance.

In an effort to understand how the physical characteristics and design of various side-firing fibers may affect the coagulation process, a limited study was initiated to compare 13 side-firing fibers of varying optical designs. A scattering phantom was used to simulate the fluence distribution that would occur in a homogeneous tissue. The peak fluence rate emitted into the phantom was determined for each side-firing fiber. A volume was determined for each fiber which contained fluence rate levels bounded by the fibers’ peak fluence rate level and 50% of the smallest measured peak fluence rate level. This volume was called the fluence rate distribution. Thus, this fluence rate distribution represents a relative volume of phantom material where the fluence rate exceeds 50% of the lowest recorded peak fluence rate for all of the tested side-firing fibers.

The concept of determining a fluence rate distribution in a liquid phantom for a side-firing fiber is similar to the measurement of SAR for microwave applicators. The SAR is the fundamental dosimetry parameter for the rf exposure safety standard. Tissue-equivalent materials that simulate the dielectric properties of different tissues at specific microwave frequencies have been developed. The SAR \( (W/kg) \) for a microwave phantom or a biological system can be calculated from a knowledge of the induced E field, the tissues’ conductivity and density. An E field within a tissue-equivalent phantom can be measured at various points with an implantable electric field probe or the phantom’s SAR distribution can be determined directly from the phantom’s temperature distribution. An equivalent optical SAR \( (W/kg) \) may be calculated from the fluence rate \( (W/cm^2) \) times the absorption coefficient \( (\mu_a, \text{mm}^{-1}) \), and divided by the density \( (\text{kg/mm}^3) \), but this optical SAR plays no role in laser exposure standards.

2 METHODS

A reservoir made of Plexiglas, and in the form of a top hat, was constructed. It is shown schematically and in cross section in Figure 1. This reservoir contained the scattering phantom which was a suspension of Intralipid (10%), Kablavitrum, Inc. (Clayton, NC) and distilled water. The amount of Intralipid-10% that was mixed with the water produced a 7.5% concentration by volume. The optical properties for this suspension were determined from the results obtained in Refs. 32 and 36. The absorption coefficient \( (\mu_a) \) was estimated to be 0.036 mm\(^{-1}\), the scattering coefficient \( (\mu_s) \) was 0.997 mm\(^{-1}\), and the anisotropy \( (g) \) was 0.48. The absorption coefficient for water at the Nd:YAG wavelength of 1064 nm is 0.018 mm\(^{-1}\). These values produce an effective attenuation coefficient \( (\mu_{eff}) \) of 0.304 mm\(^{-1}\) for the phantom suspension. The penetration depth \( (\delta) \) was then \( 1/\mu_{eff} \) or 3.3 mm. This phantom attempts to simulate the light intensity distribution in pros-
tate tissue when it is exposed by side-firing fibers. These side-firing fibers may emit multiple beams, and the main beam may be focused in one dimension. The phantom integrates all beams and distributes the light by light-particle scattering.

The side-firing fiber was positioned in the center of the top hat reservoir, and remained stationary. The isotropic detecting fiber was model 2808-A03 from PDT Systems (Santa Barbara, CA). It had a 0.8 mm diam spherical detecting tip which was attached to a 200 μm diam optical fiber. It was mounted on the rim of the top hat. The reservoir was mounted upon a rotational stage with its center coinciding with that of the reservoir. The detecting fiber scanned the emitted fluence rate distribution in cylindrical coordinates (θ, r, z). It moved in a circle around the side-firing fiber (θ), in the direction parallel to the side-firing fiber’s axis (z), and in the radial direction (r). The radial movement began at 5 mm from the side-firing fiber and then continued in increments of 2.5 ± 0.001 mm to a maximum radial position of 17.5 mm. Starting the detecting fiber at the 5 mm radial position protected it from the potentially damaging effects of the side-firing fiber’s maximum irradiance. This also requires that data extrapolation be performed to obtain fluence values from 5 mm to the origin (possible data extrapolation errors are mentioned in the discussion section). The detecting fiber rotated 180° in the θ direction in steps of 10° ± 0.05°, with the midway rotation point being at the position of maximum beam fluence. This midway position was determined with the fiber emitting into water. After this midway position was determined, the Intralipid-10% was added, and stirred. After each rotational scan (θ) the detecting fiber was moved 2 mm in the vertical or z direction. This produced total z axis scan lengths of 18, 22, or 26 mm, depending on the fluence distribution emitted from the side-firing fiber.

The isotropic detecting fiber was calibrated for fluence by using the optical calibration scheme shown in Figure 2. The laser is a continuous wave Nd:YAG laser from U.S. Laser Corp. (Wyckoff, NJ) operating at the 1064 nm wavelength. It was operated in the single transverse mode with a beam diameter of 2.0 mm (1/e²). This laser beam was launched by a 25.4 mm focal length plano-convex lens into a 600 μm diam multimode silica fiber that was approximately 3 m long. The distal end of this fiber had been melted to produce a convex surface whose radius was approximately 300 μm. The light output from the fiber was then collimated with a 101 mm focal length lens. This collimated output was reduced in diameter by a 9 mm diam aperture. This produced a beam of uniform irradiance that illuminated both the isotropic detecting fiber in a cuvette of water and the reference irradiance detector, a model 66XLA radiometer from Photodyne (Newbury Park, CA). The uniformity of this beam was checked with a model LBA-100A laser beam analyzer and model TN2250 video camera from Spiricon (Logan, UT). The uniformity of the collimated beam was compared to a uniform beam emitted from a 4 in. diam integrating sphere from UDT Sensors (Hawthorne, CA). The integrating sphere was also illuminated with the same Nd:YAG laser. The uniformity data recorded by the beam analyzer was the same for each source. Thus, the fiber fluence calibration system produced a beam uniformity which was equivalent to that emitted from an integrating sphere.

By knowing the amount of light reflected and transmitted from the beamsplitter, the irradiance

![Fig. 2 The optical schematic that describes the fluence calibration of the isotropic detecting fiber at the Nd:YAG laser wavelength of 1064 nm.](https://www.spiedigitallibrary.org/journals/Journal-of-Biomedical-Optics/article-pdf/4/3/354/2035896/354.pdf)
incident on the cuvette could be determined. The irradiance incident upon the isotropic detecting fiber was determined by correcting for the Fresnel reflections of the cuvette and the absorption of water. The detecting fiber was calibrated in the vertical position which is the same orientation that was used for the side-firing fiber measurements. The complete optical fluence mapping schematic is shown in Figure 3. The same laser used in the isotropic fiber calibration was also used in the mapping. The power launched into the side-firing fiber was 1 W ± 5%. The laser beam (2.0 mm diameter) was launched into the side-firing fiber with a plano-convex lens whose focal length was 25.4 mm. This produced launch conditions which under filled (N.A. = 0.04) the numerical aperture of the side-firing fibers (N.A. = 0.1–0.4, estimated). This also produced a small image (17 μm diam estimate) on the input fiber's face, well below its diameter (600–1000 μm range). The position of the proximal end of the fiber could be adjusted with micrometer translators in the direction of the optical axis as well as in the two axes perpendicular to it. Each fiber's distal end was positioned to the launch beam for maximum power transmission. Since most fibers had subminiature type A (SMA) connectors on the proximal end, little adjustment was necessary after the first fiber was positioned. The laser power monitoring radiometer was a Laser Probe (Utica, NY) model RK-5100 pyroelectric detector. The detecting fiber was positioned with Newport Research (Irvine, CA) model 850B linear translators (r and z), and a Newport Research (Irvine, CA) model 495 rotational translator (θ). The translators were controlled by a Newport Research (Irvine, CA) model PMC 400 motion controller. The light from the detecting fiber was converted to photocurrent with a Newport Research (Irvine, CA) model 1835-C multi-function optical meter. The motion controller and optical meter were controlled by a PC using LabVIEW software by National Instruments (Austin, TX).

The data from the PC were recorded into a spreadsheet file format and converted to the hierarchical data format (HDF) with the software program, DATA UTILITY, by Fortner Research LLC (Sterling, VA). The HDF files were then analyzed by the following software programs, SLICER, DICER, and TRANSFORM by Fortner Research LLC (Sterling, VA), and KALEIDAGRAPH by Synergy Software (Reading, PA). The SLICER, DICER, and TRANSFORM programs provided for visualizing the volumetric data contained in the HDF files. These analytic tools provided for the graphical representation [two-dimensional (2D) and three-dimensional (3D)] of the fluence distribution from each side-firing fiber as well as the calculation of the volume of phantom material that contains fluence ranging from its peak value to a value that is 50% of the lowest recorded peak value. Additionally, two exponential curves were fitted to the fluence levels over radial distances of 5–17.5 mm. One fit was made to the fluence data in the direction of the emitted beam and the other to data transverse to the emitted beam.

Thirteen models of side-firing fibers were examined, and are listed in Table 1. Most fibers were 3 m in length, the Prolase was 3.5 m, the Ultraline fibers were 3.8 m, and the angle delivery device (ADD) was 4 m in length. All fibers are intended to be used with the Nd:YAG wavelength of 1064 nm. Shown in the table is the manufacturer, the fiber model, the nominal fiber diameter, and the technology for directing the beam transverse to the optical axis. All models project a beam except for the LightStic 180 from Rare Earth Medical. This fiber has a diffusing tip which uses a reflective surface which directs the diffusing radiation transverse to the optical axis of the fiber. The power launched into this side-firing
fiber was approximately 5 W. This was done to produce peak fluence levels that were comparable to those produced by the other fibers.

3 RESULTS

Table 2 shows the calculated fluence rate distributions. The peak fluence rate values are also shown along with some beam divergence values determined in water that were taken from Ref. 12.

Figures 4 and 5 show typical plots of the fluence rate distribution in grayscale for a side-firing fiber in the 7.5% Intralipid-10% suspension. The location of the side-firing fiber is shown, and located on the z axis at the 0.0 mm x-axis and y-axis position. The fiber emits into the approximate center of the displayed volume. The fluence values shown in the figures are the detector signal levels and have not been corrected to actual fluence rate values.

Figure 6 shows a plot of the relative fluence rate as a function of increasing radius (distance from the side-firing fiber aperture). These data are taken from a two-dimensional slice of the data presented in Figure 4. This slice contains data defined by the y and z axis, and is in the direction of the emitted beam. When an exponential curve of the form $\exp(-\mu_{\text{eff}})$ was fitted to the radial fluence rate data (5–17.5 mm) from each side-firing fiber in the direction of the emitted beam and transverse to it, the following effective attenuation coefficients were obtained. In the direction of the beam, the average effective attenuation coefficient for all fibers was 0.30 mm$^{-1}$ with a standard deviation of 0.03. In the

<table>
<thead>
<tr>
<th>Side-firing fiber manufacturer</th>
<th>Model</th>
<th>Beam deflection technology</th>
<th>Fiber diameter (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rare Earth Medical</td>
<td>LightStic 180</td>
<td>Diffuser half cylinder</td>
<td>600</td>
</tr>
<tr>
<td>Sharplan</td>
<td>SideTrak</td>
<td>Refractive</td>
<td>600</td>
</tr>
<tr>
<td>Trimedyne</td>
<td>Urogold</td>
<td>Refractive</td>
<td>600</td>
</tr>
<tr>
<td>Xintec</td>
<td>6160B</td>
<td>Refractive/reflective</td>
<td>600</td>
</tr>
<tr>
<td>Xintec</td>
<td>6110A</td>
<td>Reflective</td>
<td>1000</td>
</tr>
<tr>
<td>Xintec</td>
<td>5160B</td>
<td>Reflective</td>
<td>600</td>
</tr>
<tr>
<td>LaserSonic (Laserscope)</td>
<td>Ultraline III</td>
<td>Reflective</td>
<td>600</td>
</tr>
<tr>
<td>LaserSonic (Laserscope)</td>
<td>Ultraline I</td>
<td>Reflective</td>
<td>600</td>
</tr>
<tr>
<td>Myriadase</td>
<td>DFL-1000 SideFire</td>
<td>Reflective</td>
<td>600</td>
</tr>
<tr>
<td>Myriadase</td>
<td>DFL-1100</td>
<td>Reflective</td>
<td>600</td>
</tr>
<tr>
<td>Cytocare</td>
<td>Prolase II</td>
<td>Refractive/reflective</td>
<td>1000</td>
</tr>
<tr>
<td>Laserscope</td>
<td>ADD</td>
<td>Reflective</td>
<td>400</td>
</tr>
<tr>
<td>Bard (Trimedyne)</td>
<td>Urolase 350 000</td>
<td>Reflective</td>
<td>600</td>
</tr>
</tbody>
</table>

Table 2 The fluence rate distributions that were determined for 13 different side-firing fibers. The emitted beam divergence values in water were taken from Ref. 12. Since most of the beams emitted from side-firing fibers are asymmetric, the divergence data show beam angles for two orthogonal directions. The data used to determine the fluence rate distributions were obtained with a nominal launch power of 1 W (5 W for the LightStic 180) into the side-firing fiber.

<table>
<thead>
<tr>
<th>Fiber model</th>
<th>Fluence rate distribution ±10% (cm$^2$)</th>
<th>Peak fluence rate ±10% (mW/cm$^2$)</th>
<th>Divergence angle from Ref. 12</th>
</tr>
</thead>
<tbody>
<tr>
<td>LightStic 180 (5 W input)</td>
<td>2.48</td>
<td>0.24</td>
<td>...</td>
</tr>
<tr>
<td>SideTrack</td>
<td>0.60</td>
<td>0.3</td>
<td>...</td>
</tr>
<tr>
<td>Urogold</td>
<td>0.91</td>
<td>0.17</td>
<td>...</td>
</tr>
<tr>
<td>6160B</td>
<td>4.16</td>
<td>1.21</td>
<td>...</td>
</tr>
<tr>
<td>6110A</td>
<td>0.71</td>
<td>0.18</td>
<td>...</td>
</tr>
<tr>
<td>5160B</td>
<td>3.61</td>
<td>1.41</td>
<td>...</td>
</tr>
<tr>
<td>Ultraline III</td>
<td>5.16</td>
<td>4.97</td>
<td>...</td>
</tr>
<tr>
<td>Ultraline I</td>
<td>4.88</td>
<td>1.41</td>
<td>12×15</td>
</tr>
<tr>
<td>DFL-1000 SideFire</td>
<td>4.72</td>
<td>1.13</td>
<td>10×10</td>
</tr>
<tr>
<td>DFL-1100</td>
<td>4.65</td>
<td>1.51</td>
<td>...</td>
</tr>
<tr>
<td>Prolase II</td>
<td>0.21</td>
<td>0.08</td>
<td>20×26</td>
</tr>
<tr>
<td>ADD</td>
<td>1.05</td>
<td>0.58</td>
<td>8×16</td>
</tr>
<tr>
<td>Urolase-350 000</td>
<td>3.23</td>
<td>0.5</td>
<td>20×40</td>
</tr>
</tbody>
</table>
direction transverse to the emitted beam, the average effective attenuation coefficient for all fibers was 0.29 mm⁻¹ with a standard deviation of 0.09.

4 DISCUSSION

Determining the volume of prostate tissue that is coagulated during VLAP is a major goal of clinical dosimetry studies.¹⁷–²³ In these studies the coagulated tissue volume is reported along with the delivered laser power and exposure duration. Also, the maximum relative irradiance of side-firing fibers¹² has been used as a criteria for determining their appropriate use (coagulation or vaporization). However, side-firing fibers are also known to emit secondary beams caused by spurious reflections.

Fig. 4 The fluence rate distribution of the ADD side-firing fiber in an optical phantom. The side-firing fiber’s position is as shown. The intensity distribution is shown in grayscale. The fluence rate values shown are signal values and have not been corrected to actual fluence rate units of W/cm². The arrows illustrate the general emitted beam direction.

Fig. 5 The fluence rate distribution determined for the LightStic side-firing fiber. The intensity distribution is shown in grayscale. The fluence rate values shown are signal values and have not been corrected to actual fluence rate units of W/cm². The position of the side-firing fiber is shown. The LightStic emits diffuse laser light along 35 mm of its distal end.

Fig. 6 The uncorrected detecting fiber signal as a log function of radius for the ADD side-firing fiber. This distribution is for data along the x axis shown in Figure 4. The dashed curve is for the fitted exponential function shown.
of the main beam, and some designs also cause beam focusing. The effects of these spurious reflections and beam focusing properties are accounted for in the optical phantom. The side-firing fibers’ beam divergence and exit angle do not account for tissue scattering nor do they indicate what the maximum fluence rate levels are and how they are distributed. While the actual coagulated volume and the defined fluence rate distribution are different, the fluence rate distribution will mimic the actual light distribution resulting from tissue scattering depending upon the composition of the phantom.

The fluence rate distributions and maximum fluence rate values that are presented in Table 2 are relative values, since the fluence rate distributions minimum fluence level was dictated by the low fluence rate levels of the Prolase II side-firing fiber. If the fluence rate level of the LightStic 180 had been determined at the 1 W level, then it would have replaced the Prolase II and all of the other fluence rate distributions would have increased. This was not done because of the significant difference in device technology that exists between the LightStic 180 and the other side-firing fibers. See Figures 4 and 5.

Using the 50% level of the maximum fluence rate criteria as the lower bound of the fluence rate distribution is somewhat arbitrary but follows the definition of optical pulse width measurements (the pulses full width, measured at half maximum intensity) defined in the American National Standards Institute (ANSI) laser exposure standard. It may be possible to obtain a fluence rate distribution that is more representative of the coagulation volume by performing a numerical analysis of the coagulation process. Using a computationally intensive model done with Monte Carlo simulation and the bio-heat transfer equation, and using a 360° emitting diffusing applicator in human prostate, with exposure conditions of 5 W and 10 min, it would appear that the fluence rate distribution bounded by a value approaching 10% of the maximum fluence rate is similar to the coagulated volume. A similar analysis has not been performed for side-firing fibers. Further, clinical studies which measure both the extent of the fluence rate and the resulting coagulation haven’t been performed, as clinical studies which examine light distribution in human prostate have been concerned with photo-dynamic therapy, not coagulative therapy.

When the optical properties of the phantom used in this study ($\mu_a = 0.054 \text{ mm}^{-1}$, $\mu_s = 0.997 \text{ mm}^{-1}$, $g = 0.48$, and $\mu_{\text{eff}} = 0.304 \text{ mm}^{-1}$ or $\delta = 3.3 \text{ mm}$) are compared to native canine prostate tissue ($\mu_a = 0.04 \text{ mm}^{-1}$, $\mu_s = 11 \text{ mm}^{-1}$, $g = 0.96$, and $\mu_{\text{eff}} = 0.24 \text{ mm}^{-1}$ or $\delta = 4.17 \text{ mm}$), and native human prostate tissue ($\mu_a = 0.03 \text{ mm}^{-1}$, $\mu_s = 8 \text{ mm}^{-1}$, $g = 0.95$, and $\mu_{\text{eff}} = 0.197 \text{ mm}^{-1}$ or $\delta = 5.1 \text{ mm}$) there are obvious differences. The scattered light from a Nd:YAG beam launched into canine or human prostate tissue is almost totally scattered in the forward direction ($\cos^{-1}0.96 = 16°$) while the beam is scattered over a much wider area ($\cos^{-1}0.48 = 61°$) in a suspension of Intralipid and water. Also, there is a difference of about a factor of 10 in the scattering coefficients. It may be possible to construct a phantom with polystyrene microspheres which produces more forward scattering than an Intralipid based phantom; however, keeping the sphere distribution homogeneous for long fluence mapping times could be a problem.

The fluence rate distributions that were presented in Table 2 and shown in Figures 4 and 5 are pseudo volumes. These volumes were determined from data in a 180° field, when in fact the total fluence rate distribution surrounds the entire fiber. Total fluence rate distributions would require taking data in a full circular arc (360°). Data were taken for the Sharplan SideTrak side-firing fiber over 360° in order to estimate this additional scanned volume. The total fluence rate distribution was found to be 0.76 cm³ or a 27% increase over the value listed in Table 2.

The calculated fluence rate distributions shown in Table 2 may also suffer from fluence rate extrapolation errors for radial values from 0.0 to 5.0 mm. One possible way to investigate this error is to use Monte Carlo simulation for a number of side-firing fibers submerged in the Intralipid suspension. The results from this modeling could then be compared to the fluence rate distributions obtained in this study. In the absence of such a study, the modeled fluence rate emitted from a flat fiber (a symmetrical beam) can be used for comparison. For reflective side-firing fibers, there is a good comparison with this modeled fiber, as the fluence is maximum at the origin and then decreases with radial distance. This is shown in Figure 7(a) for the Urogold reflective side-firing fiber. The relative fluence rate shown in the figure does not decrease exponentially from zero as the modeled fiber, however, the reflective surface of the Urogold side-firing fiber is concave, and this distorts the symmetrical beam distribution emitted from its fiber. For those fibers that refract the beam with TIR and produce an asymmetrical focused beam, the maximum fluence level appears between the 0 and the 5 mm radial positions. An example of this is shown in Figure 7(b) for the ADD side-firing fiber. Here the maximum relative fluence rate occurs at a radial distance of 3.5 mm.

The good agreement between the average fitted and computed effective attenuation coefficients (Figure 6) indicates that the fluence rate outside a radius of 5 mm will decrease by exp[$-r\mu_{\text{eff}}$] in both the direction of the beam as well as transverse to it. This would not apply to distances less than 5 mm from the fiber, as the side-firing fibers near field asymmetrical beams produce fluence rate distribu-
tions which differ from those obtained from simple models. This would also apply only to distributions in this study.

For a single exposure (60 W, 60 s) to canine tissue with a side-firing fiber, a coagulation volume of 3.92 cm³ was created. This value is within the range of fluence rate distribution values found in Table 2. This is fortuitous as the phantom is optically simple (a homogeneous suspension) when compared to an actual prostate exposure in a water filled urethra. Here the heterogeneous optical properties are changing with laser exposure, and blood perfusion is limiting the coagulation volume by removing heat. The fluence rate distributions determined in this comparison are also limited to the laser and launch conditions used in this study, as differing launch conditions will affect the fiber’s fluence distribution.

In spite of the limitations of the fluence rate distributions obtained in this study (no direct fluence rate measurements in a radial direction less than 5 mm from a fiber, an angular data field of 180°, homogeneous optical properties that do not change and are not identical to either native canine or human prostate tissue, launch optics and Nd:YAG laser differences, and a 50% minimum fluence rate coagulation level), it is worthwhile investigating how well the relative fluence rate distributions of two side-firing fibers, the Urolase and the Prolase II, predict the dosimetry conditions for coagulating prostate volumes during clinical dosimetry studies of canine and human prostates.

Unfortunately, none of the canine prostate exposures were performed with the same laser power and exposure times, and the effects of some of the exposures overlapped. However, if all the coagulated volume exposure data (J/cm³) are averaged for each side-firing fiber and compared, then the Prolase II requires approximately 386 more Joules than the Urolase fiber to coagulate 1 cm³ of canine prostate tissue. When looking at the fluence rate distributions for these two fibers, it is not surprising that the Prolase II requires more energy to coagulate canine tissue. Its fluence rate distribution (FRV) and peak fluence level are much smaller than that of the Urolase fiber (0.21 vs 3.23 cm³, FRV, and 0.08 vs 0.50 mW/cm², fluence rate). In order to increase the fluence rate distribution of the Prolase II fiber to that of the Urolase, the power (fluence levels) would have to be increased.

Two excellent human prostate dosimetry studies have been performed. These studies have identical single exposure data for the two fibers. For a 40 W, 90 s exposure, the Prolase II fiber produced a coagulated volume of 1.98 cm³ while the Urolase produced a volume of 3.68 cm³. For a 60 W, 60 s exposure, the Prolase II produced a coagulated volume of 3.07 cm³ while the Urolase produced a volume of 1.27 cm³. Again, the Prolase II is more efficient at the higher power level, since its small fluence rate distribution would expand to a larger volume at the higher power levels. The loss of coagulated volume with the Urolase at the 60 W power level was probably due to nonlinear tissue effects caused by the excessive fluence. This would prevent the light from diffusing to the depths that occurred at 40 W. The phantom used in this study will not predict this nonlinear behavior; however, one would certainly expect that the power level would have to be increased for the Prolase II to coagulate a volume similar to that of the Urolase fiber.

Based upon the fluence rate distribution data in Table 2, the Xintecl 5160B side-firing fiber (3.6 cm³ FRV) should perform in a similar manner to the Urolase fiber (3.2 cm³ FRV), and side-firing fibers with fluence rate distributions greater than 4 cm³ should require less power to produce an equivalent coagulation volume. However, this conclusion would require confirming canine or human dosimetry studies. Successful clinical VLAP studies which used the Ultraline (5.16 and 4.88 cm³ FRV) and Myriaxlase (4.72 and 4.65 cm³ FRV) side-firing fibers do not directly support this premise, since the exposures overlapped and coagulated volumes were not directly measured. Satisfactory VLAPs were performed with the Ultraline side-firing fiber.
using a 60 W power setting,44–46 and with the Myriadase side-firing fiber using a 40 W power setting.47 Typically the Ultraline side-firing fiber is used in the contact mode in an effort to remove tissue by vaporization. It has been shown48 that this fiber deteriorates with use (loss of fiber transmission), so the actual power (total energy) delivered for vaporization and coagulation using an Ultraline side-firing fiber for a VLAP procedure is probably unknown. The fluence rate distribution would have little value in dosimetry evaluation when vaporization of tissue is a desired therapy.

5 CONCLUSIONS

A 7.5% water suspension of Intralipid-10% makes a reproducible optical phantom for mapping the fluence rate emitted from side-firing fibers. The fluence rate distribution accounts for the simultaneous effects resulting from a side-firing fiber’s beam direction (divergence angle, beam exit angle), plus any additional spurious beams, and beam focusing. The fluence rate distribution also accounts for tissue induced light scattering. The fluence rate distribution represents a comparative measure for evaluating the scattered light volume emitted from different side-firing fibers. Based upon limited clinical dosimetry studies, the relative fluence rate distribution appears to indicate general exposure criteria for the evaluated fibers. However, it is a relative performance comparison and is limited to the side-firing fibers and fiber launch conditions used in this study. Since there are a number of Nd:YAG laser sources, as well as a number of side-firing fibers, the determination of their fluence rate distributions could prove problematic.

It should be possible to define a standard side-firing fiber (either physical or virtual) which would serve as the comparative fiber. Such a fiber would make this relative fluence rate distribution test independent of the fibers used in this study. Numerous technological details, such as standard launch conditions, fiber size, and numerical aperture, would have to be determined. Alternatively, a coagulating phantom which scattered light in the same manner as human prostate may provide an easier, fiber independent, means of performance comparison. It would also provide for performance evaluation at typical clinical exposure levels where the side-firing fiber’s performance has been shown to deteriorate.25,48

A preliminary investigation has been made into creating a liquid phantom whose optical properties matched those of native human prostate and has the ability to coagulate at the higher clinical power levels. The optical properties of a liquid phantom containing different concentrations of Intralipid and egg albumin were determined at the 1064 nm wavelength. At the highest concentration of Intralipid (12.6%), the effective attenuation coefficient was 0.196 mm⁻¹ which compares very favorably to the native human prostate value of 0.197 mm⁻¹. This albumin-Intralipid based phantom appears to have both the capability for scattering the Nd:YAG laser light emitted from a submerged optical fiber in the same manner as a homogenous human prostate, and for coagulating a volume of phantom material at treatment exposure levels. This coagulated volume could then be used as a relatively easy method for comparing the performance of different laser fibers with different clinical lasers sources.

NOTES

The mention of commercial products or their use in connection with material reported here is not to be construed as either an actual or implied endorsement of such products by the U.S. Food and Drug Administration.

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