Optimizing light delivery for a photoacoustic surgical system

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ABSTRACT

This work explores light delivery optimization for a photoacoustic surgical system previously proposed to provide real-time, intraoperative visualization of the internal carotid arteries hidden by bone during minimally invasive neurosurgeries. Monte Carlo simulations were employed to study 3D light propagation in tissue. For a 2.4 mm diameter drill shaft and 2.9 mm spherical drill tip, the optimal fiber distance from the drill shaft was 2 mm, determined from the maximum normalized fluence seen by the artery. A single fiber was insufficient to deliver light to arteries separated by a minimum of 8 mm. Using similar drill geometry and the optimal 2 mm fiber-to-drill shaft distance, Zemax ray tracing simulations were employed to propagate a 950 nm wavelength Gaussian beam through one or more 600 μm core diameter optical fibers, and the resulting optical beam profile was detected on the representative bone surface. For equally spaced fibers, a single merged optical profile formed with 7 or more fibers, determined by thresholding the resulting light profile images at 1/e times the maximum intensity. The corresponding spot size was larger than that of a single fiber transmitting the same input energy, thus reducing the fluence delivered to the sphenoid bone and enabling higher energies within safety limits. A prototype was designed and built based on these optimization parameters. The methodology we used to optimize our light delivery system to surround surgical tools is generalizable to multiple interventional photoacoustic applications.

1. INTRODUCTION

Photoacoustic imaging has been proposed as a method to guide minimally invasive neurosurgeries, specifically surgeries to remove pituitary tumors using the endonasal transsphenoidal approach.[1] In this approach, the light delivery system would be attached to the surgical tool and inserted in the nose to transmit light across the sphenoid bone while this bone is being removed to access the pituitary tumor. The internal carotid arteries hidden behind the bone would absorb the light, undergo thermal expansion, and generate an acoustic response to be detected by an external transcranial ultrasound probe placed on the patient's temple.[1]

The minimum energy required to visualize real blood ranged from 1.2-6 mJ when the thickness of the cranial bone ranged from 0 mm to 2 mm, which corresponds to a fluence range of 4-21 mJ/cm^2 for the 6 mm diameter fiber bundle used to deliver the light.[2] These results demonstrated the feasibility of visualizing real blood in the presence of bone within the 26.4 mJ/cm^2 safety limit for 760 nm wavelength of light.[3] In addition, a mock tool tip (consisting of a paper clip glued to a metal ball) was placed to provide satisfactory preliminary evidence that surgical tool tips can be visualized simultaneously with blood vessels using a single 6-mm diameter fiber bundle.[2]

Although these previous results are encouraging, the light delivery design is not practical for minimally invasive surgeries because a 6-mm diameter fused fiber bundle is too bulky to be attached to surgical tools and in most cases it would be larger than the surgical tool itself. This paper describes the use of multiple fibers surrounding the tool tip to achieve the light delivery required for visualization of real blood within safety limits. A prototype was designed and built based on simulation-derived optimization parameters. An expanded version of this report is available as a journal article.[4]

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2. METHODS

2.1 Monte Carlo Light Propagation Simulations

Monte Carlo Simulations[5] were implemented to understand how the fluence seen by the arteries changes with respect to bone thickness (bt), distance between the artery and the bone (d_v), distance between the light source and the drill shaft (d_f), and distance between two arteries (d_b). These simulations trace the optical path from the light source in 3-dimensional space, voxel by voxel, also taking the optical properties for blood, bone, and brain matter into account, as well as those of the tool. This information provides insight into potential artery visibility in a photoacoustic image.

These simulations were split into one and two vessel scenarios, as illustrated in Fig. 1. For both scenarios, the drill was modeled with a spherical drill tip of diameter 2.9 mm connected to a cylindrical drill shaft of diameter 2.4 mm. The metallic drill contacted the bone surface, and a single light source was placed next to the drill shaft, set 4.95 mm back from the drill tip. For the one vessel simulations, the artery was simulated with a diameter of 4 mm and a length of 9 mm, and it was positioned directly below the drill. The bone thickness was varied from 0 mm to 8 mm, the distance between vessel and bone was varied from 0 mm to 5 mm, and the distance between the source fiber and the drill shaft was varied from 0 mm to 5 mm. Only one parameter was varied at a time, otherwise, the bone thickness, fiber distance, and vessel distance (bt, d_f, and d_v, respectively in Fig. 1) were held constant at 2.5 mm, 1.25 mm, and 1 mm, respectively. For the two vessel simulations, the arteries had the same dimensions as the single vessel simulation, and they were positioned parallel to each other and equidistant from the drill. For these simulations the distance between two internal carotid arteries was varied from 0 mm to 8 mm. The bone thickness, fiber distance, and vessel distance (bt, d_f, and d_v, respectively in Fig. 1) were held constant at 2.5 mm, 1.25 mm, and 1 mm, respectively. One optical fiber was used for these simulations. The output of these simulations was an image that displayed the normalized fluence in units of $\log_{10}(\text{cm}^{-2})$.

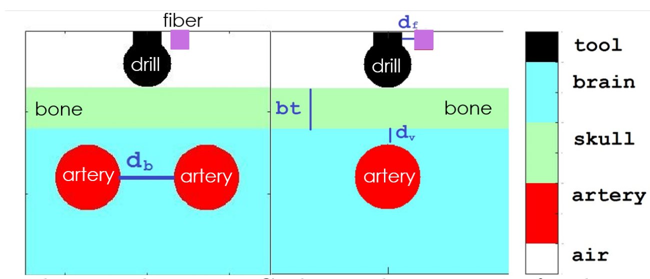

Figure 1: Schematic diagrams showing the Monte Carlo simulation setups for the two (left) and one (right) vessel simulations. The variable in each simulation setup are indicated on each diagram: distance between two arteries (db), bone thickness (bt), distance between artery and bone (d_v), and distance between the fiber and the drill shaft (d_f).

2.2 Zemax Ray-Tracing Simulations

Zemax simulations (Zemax LLC., Kirkland, WA USA) were employed to model a metal drill acting as an absorber that blocked light from reaching the bone surface. The drill had a spherical drill tip of diameter 2.9 mm and a drill shaft diameter of 2.37 mm. The fibers were modeled as glass core and cladding, and they were set back at a distance of 5.6 mm from the drill tip, as illustrated in Fig. 2. The core and cladding had the same index of refraction as the commercially available fibers we used for the prototype described in Sec. 2.3.

The goal of these simulations was to determine the number of fibers required for our light delivery system. Thus, the number of fibers was varied from 1 to 10, and we identified the threshold where the multiple beams incident upon the bone overlapped enough to make one individual beam rather than form multiple hot spots. Smoothing was applied to the beam profile. In order to qualitatively determine whether or not a spot was uniform, the images were exported to MATLAB (MathWorks, Natick, Massachusetts) and thresholded. The threshold was set at 1/e times the peak intensity. If the pixels that are within 1/e of the peak intensity of the image form a complete torus, then we considered this to indicate uniformity at the detector surface. The 1/e beam profile was used for thresholding because the ANSI safety limits are based on this measurement.[3]

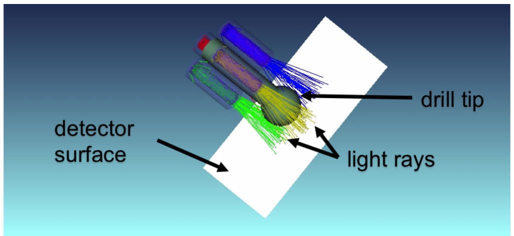

Figure 2: Schematic diagrams showing the Zemax simulation setup.

2.3 Light Delivery System Design Requirements

We built a light delivery prototype based on design requirements that were determined from the simulation results (Sec. 3). The first design requirement is that 7 or more fibers are necessary to achieve the desired beam profile. Second, the fibers should be equally spaced and held 2 mm away from the drill shaft. A commercially available 1-to-7 splitter was utilized to meet these requirements. The fiber was modified by cleaving the SMA connectors from the 7-fiber fan-out end, and exposing 2 cm of the fiber jacket and 1 cm of the fiber cladding. The fibers were then polished for a flat cleaved finish. The fibers were held 2 mm away from the drill, and equally spaced using a custom 3D printed part.

3. RESULTS

3.1 Monte Carlo Simulation Results

When the distance between the source and the drill shaft was increased, the amount of light seen by the underlying vessel increased the decreased, as shown in Fig. 3. The optimal distance was found to be 2 mm. This result was incorporated into the Zemax physical optics propagation simulations. In addition, fluence decreased as bone thickness and vessel distance increased (not shown).

The two-vessel simulation showed that there is a significant difference in fluence between two vessels if only one source fiber is used as illustrated in Fig. 4. The fluence seen by the vessel farthest from the fiber is approximately zero. This result shows that it is unreasonable to use one fiber in our design, because it would be difficult to visualize two arteries simultaneously and because the asymmetry would not provide accurate information about vessel proximity if approaching an artery from the fiberless side of the tool.

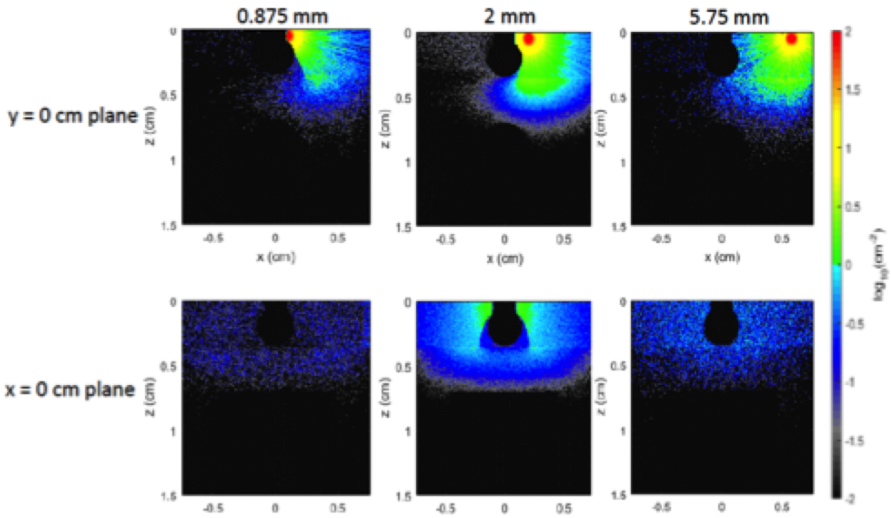

Figure 3: The distance of the fiber from the drill shaft alters the normalized fluence distribution. Images display the normalized fluence when the fiber is located at distances of 0.875 mm, 2 mm, and 5.75 mm from the drill shaft (as indicated above each image).

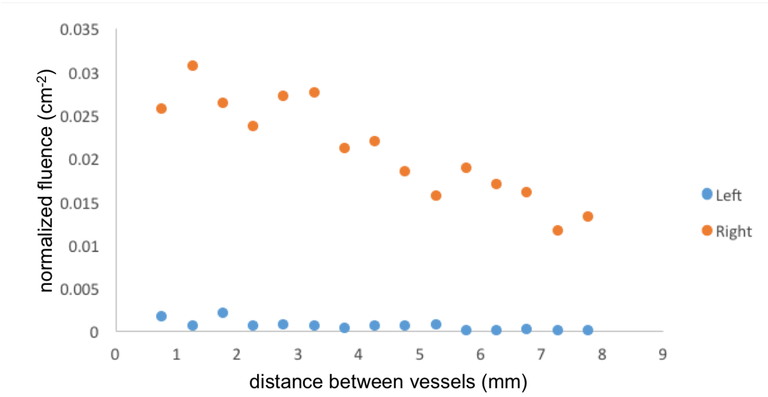

Figure 4: Normalized fluence as a function of the distance between two arteries.

3.2 Zemax Simulation Results

With the Zemax simulations, the incident laser spot size increased as the number of fibers increased and the number of spots eventually transformed from creating multiple hot spots to creating a single beam, as demonstrated in Fig. 5. A single uniform beam was formed with 7 or more fibers for a numerical aperture of 0.39 and a core diameter of 600 μm. One measurement for the increase in spot size is beam diameter. Note that as the number of fibers increased, the beam's outer diameter increased while the inner diameter decreased.

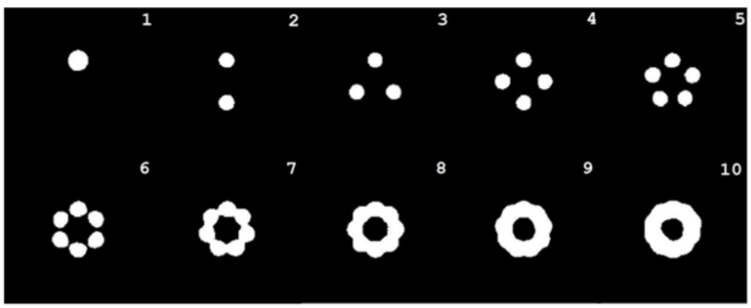

Figure 5: Number of spot sizes observed as a function of the number of fibers surrounding the drill.

Fig. 6 illustrates that as the drill is moved away from the detector surface (which could represent the bone or tissue surface that blocks an underlying structure of interest), the spot size increases and the beam profile changes from a torus to a Gaussian beam, where it is most intense at the center. This transition occurs at a distance of approximately 12-13 mm from the fiber tips, which corresponds to approximately 6-7 mm from the drill tip as shown in Fig. 6 (because the fibers are set back 5.6 mm from the drill tip).

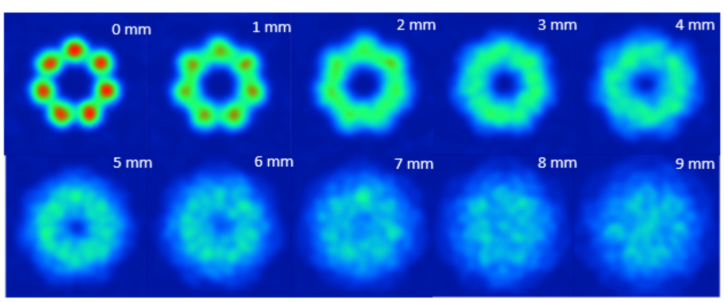

Figure 6: Near and far-field beam profile results demonstrating the transition of the beam profile from a torus to a Gaussian as the distance between the drill tip and the surface increases from 0 mm to 9 mm.

3.3 Light Delivery System Prototype

The simulation results provided design requirements for our light delivery system prototype, which are summarized in Sec. 2.3. The prototype consists of a 1-to-7 fiber splitter that surrounds the drill, and the seven fibers are held in place with a customized 3D printed part, as seen in Fig. 7.

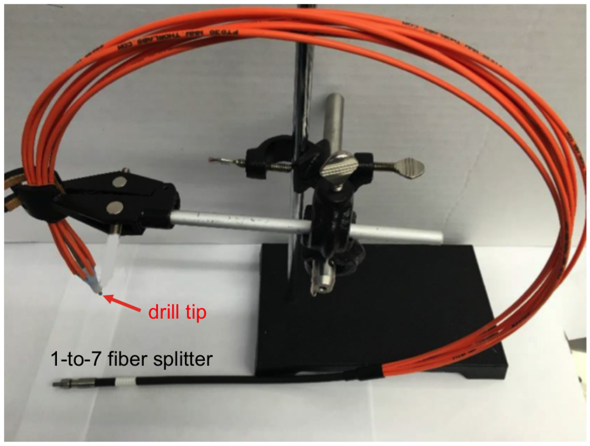

Figure 7: Light delivery prototype with optical fibers surrounding the surgical drill and secured in a customized 3D printed part.

4. DISCUSSION

We successfully designed and built a prototype light delivery system to surround a surgical tool, taking into account the optimal placement and number of fibers for this system based on simulation results. This is the first multifiber light delivery design to surround a surgical tool, particularly intended for interventional photoacoustic imaging.

Because the use of multiple fibers surrounding the tool tip increases the maximum achievable spot size compared to the fused fiber bundle approach,[2] we can now use a higher energy input to make photoacoustic images (assuming that the merged Gaussian beams from each individual fiber is sufficient to ensure that the average energy over an area corresponding with 1/e times the maximum energy will not exceed ANSI laser safety limits). Considering that at least 1.2 - 6 mJ is required to visualize blood through bone thicknesses ranging 0-2 mm,[2] we can potentially use higher energies without increasing patient risk, particularly when the bone is thicker than 2 mm (but thinner than 4-5 mm). Therefore, it is reasonable to conclude that the use of higher energies for higher bone thicknesses will simultaneously improve signal-to-noise ratios without needing to average multiple frames and thereby increase the real-time frame rate for image display.

We note that the custom 3D printed plastic part used to hold the fibers in place could potentially act as a mechanical bushing that enables drill rotation and operation while the multifiber locations remain stationary. In the future, this 3D printed part will be attached to the stationary handle of the surgical drill for testing while the drill is in motion, and the fiber jackets will be stripped to minimize the bulk of our light delivery system prototype for use during an actual surgery.

5. CONCLUSION

We have reported our success with designing and building a multifiber light delivery system to surround the tip of a surgical tool. In particular, the optical design reported in this paper is optimized for a neurosurgical drill. For a 2.9 mm spherical drill tip, the optimal fiber distance from the 2.4 mm drill shaft was identified as 2 mm. At this optimal distance, the optical profile merges with 7 or more fibers. The increased spot size with a 1-to-7 fiber splitter decreases fluence, and enables higher energies within safety limits. The methodology used to obtain these results may be applied to design and build custom multifiber light delivery systems for an entire suite of surgical tools.

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