Contact focusing multimodal microprobes for ultraprecise laser tissue surgery

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Abstract: Focusing of multimodal beams by chains of dielectric microspheres assembled directly inside the cores of hollow waveguides is studied by using numerical ray tracing. The device designs are optimized for laser surgery in contact mode with strongly absorbing tissue. By analyzing a broad range of parameters it is demonstrated that chains formed by three or five spheres with a refractive index of 1.65-1.75 provide a two-fold improvement in spatial resolution over single spheres at the cost of 0.2-0.4 attenuation in peak intensity of the central focused beam. Potential applications include ultra precise laser ablation or coagulation in the eye and brain, cellular surgery, and the coupling of light into photonic nanostructures.

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1. Introduction

Focusing of light is widely used in optical microprobes where a stable and well-confined beam of photons is scanned or directed over a biological sample or photonic structure. Potential applications of such technologies are numerous, including ultra-precise laser surgery [1], nanoscale patterning [2], piercing of a cell [3], and biomedical optical spectroscopy [4]. Typically, these applications require a combination of high spatial resolution and high power transmission. However, sharp focusing of light is made difficult in such applications because of the multimodal structure of the beams delivered by flexible hollow waveguides (HWG) or multimode fibers. In principle, this problem can be solved by using single mode fibers;
however they are not readily available in the mid-infrared (IR) range. In addition, they have limited coupling efficiency with many practical radiation sources, and limited power transmission properties as well.

An important potential application of such microprobes is for use in ultraprecise ophthalmic laser surgery [5]. Over the past two decades various optical elements including spheres [6], hemispheres, dome, cone, and slanted shapes [7,8], cylindrical gradient index (GRIN) lenses [5], and tapered fibers have been tested as tips for ophthalmic laser scalpels. It should be noted, however, that these devices are usually designed to operate in non-contact mode in air at a fixed working distance from the tissue. However, the working distance is difficult to control during surgery and may result in a variable laser spot size and intensity in tissue.

In this work we focus on developing contact microprobes which ablate the tissue under touching conditions with the tip of the optical scalpel. Although contact surgery may allow for better surgical control and protection for adjacent healthy tissue, it has been used only in a limited number of studies [9]. It should also be noted that due to the refractive index of tissue conventional microprobes designed to operate in air lose their focusing capability when operated in contact with tissue [5–8]. Ultraprecise contact surgery tools should provide compact beam sizes in tissue in both the depthwise and transverse dimensions. It is well known that the optical penetration depth is reduced by strong absorption. For example, Erbium:YAG laser ($\lambda = 2.94 \mu m$) radiation closely matches a major water absorption peak in tissue, providing an optical penetration depth and peripheral thermal damage zone of $\sim 10-30 \mu m$ during incision of ophthalmic tissues, when the Er:YAG laser is operated in free-running (long-pulse) mode with a pulse length of $\sim 300 \mu s$ [10–12]. Minimization of the transverse beam diameter in tissue, however, requires solving the problem of sharp focusing of multimodal beams of light delivered by HWGs or multimodal fibers.

Recently, it was demonstrated that a small wavelength-scale microsphere with a refractive index ($n$) of approximately 1.6 produces a narrow focused beam, termed “photonic nanojet” [13–17]. Such a nanojet propagates with little divergence for several wavelengths into the surrounding medium, while maintaining a subwavelength transverse beam width. This concept is attractive for designing focusing devices with high optical transmission properties. It should be noted, however, that photonic nanojets from single spheres require strictly plane-wave or conical illumination which is not readily available in many applications.

Another approach to this problem is based on the use of chains of spheres. Initial studies have concentrated on the optical coupling effects between whispering gallery modes (WGMs) in microspheres [18–24]. The notable advance in this area has been the identification of two distinct mechanisms of optical transport by using the generalized multiparticle Mie theory: evanescent WGM-based coupling and nanojet coupling [25]. Both mechanisms have been observed in chains of polystyrene microspheres assembled on substrates [19,26–30]. The key result for developing applications was the observation that the coupled nanojets decrease in size along the chain, reaching wavelength-scale dimensions [26]. It was also shown that the nanojet coupling leads to formation of “nanojet-induced modes” (NIMs) with the periodicity of the focused beams corresponding to the size of two spheres and with very small propagation losses on the order of 0.1 dB/sphere [26,27].

More recently, it was suggested that the chains of microspheres can be used in contact surgery because the last nanojet appearing in close proximity to the surface of the end sphere is not strongly affected by the external medium, since focusing is accomplished primarily inside the chain [31–34]. Additional advantages of these structures are their simple integration with HWGs currently used as a flexible delivery system in laser surgery and their ability to focus multimodal input beams.

In this work we develop guidelines for designing high resolution contact focusing multimodal microprobes based on numerical ray tracing. By analyzing a broad range of parameters it is shown that chains of spheres with $n$ around 1.65–1.75 provide a two-fold improvement of the spatial resolution over single spheres. A power coupling efficiency to the last photonic nanojet on the order of several percent can be obtained in such structures.
2. Theoretical model

The aim of our modeling was to create a generic design based on standard optical components, rather than one specific to a single application. As illustrated in Fig. 1 (a), we assumed that a multimode fiber with a core diameter of 150 \( \mu \)m is inserted into a HWG with an internal diameter of 300 \( \mu \)m. These dimensions are available for mid-IR waveguides [35]. The wavelength \( \lambda=2.94 \mu \)m was selected to match the emission line of the Er:YAG laser. Smaller HWG diameters would allow more compact focusing of the beams, however diameters smaller than \( a=300 \mu \)m are rarely used in practice because of the strongly increasing propagation losses \( \alpha \sim 1/a^3 \), where \( \alpha \) is the attenuation coefficient [35]. Previously HWGs tapered to \( a=100 \mu \)m have been used for achieving compact beams [36], however such tapers have significant insertion losses and are challenging to fabricate.

Chains of \( N \) identical touching microspheres with diameter \( D=300 \mu \)m and index of refraction \( n \) from 1.4 to 2.0 were placed close to the HWG edge in a configuration in which half of the end sphere was extended from the HWG. The distance between the fiber and first sphere \( (d) \) was discovered to be not critically important for this design in the limit of sufficiently large separations, as explained in Section 3.1. We neglected the absorption of light by the HWG sidewalls which is a good approximation for \( d<1 \) m [35]. We also neglected the absorption of light by the material of microspheres.

The millimeter-scale dimensions of structures considered in this work make it difficult to find exact solutions for Maxwell’s equations based on numerical or analytical techniques such as an integral formulation [37]. On the other hand, application of geometrical optics is well justified in the case of \( D \sim 100 \lambda \) [6]. In this work we used ZEMAX EE ray tracing software which has many advanced capabilities for designing similar structures [38].

The mid-IR multimode fibers used in laser surgery can support hundreds of modes and have an approximately Gaussian radiant intensity distribution with a typical divergence angle 2\( \alpha=12^\circ \). In order to model this source in ray optics we assumed a random distribution of point sources of light at the front surface of the fiber core, as shown in Fig. 1 (b). A total of \( 2 \times 10^6 \) rays were traced in each calculation. To approximate a Gaussian distribution in the far field, a
weight factor (WF) was introduced for the irradiance of each ray, depending on its starting polar angle (θ): \( WF(\theta) = \exp(-20\theta^2 / \alpha^2) \), where \( \alpha = 6^\circ \) [6]. The calculated average WF was applied in ten equally spaced angular steps from 0 to 6\(^\circ\) along the cone of directions centered with the normal to the fiber surface. The rays were randomly distributed in azimuthal directions for each point source to take into account skew modes.

Inside the device, the ray tracing takes into account the mirror reflections by the sidewalls of HWG and the Fresnel reflections and refractions at the spherical interfaces. This leads to a gradual attenuation of initial rays once they propagate through the structure and to the emergence of new rays. We traced the rays down to a level of \( WF=10^{-4} \) which provided good convergence of the numerical solutions. The intensity distributions created by the device were calculated from the density of the rays at the detector plane with their WF taken into account. Due to the cylindrical symmetry of the problem the calculated intensity profiles are not dependent on the direction of linear polarization of the point sources. All calculations were performed with a random polarization of light.

To take into account the short penetration depth of light in tissue at \( \lambda=2.94 \mu m \) we placed a totally absorbing detector in a contact position with the end sphere in the \( xy \)-plane, as illustrated in Fig. 1(a). The square detector with 5000x5000 pixels had 300x300 \( \mu m \) size. It should be noted that the use of a flat detector in our modeling is only an approximation to realistic surgical conditions due to a variable gap between the surface of the end sphere and detector introduced by the curvature of the sphere. In practical surgery the laser scalpel can be pressed slightly against the tissue surface providing close contact with the tissue along the surface of the end sphere. We proved numerically, however, that the gap between the surface of the sphere and detector is too narrow to significantly deflect the rays sideways. We increased the index of the medium in this gap region up to 1.33 and observed changes of the calculated intensity profiles at the detector plane of only few percent for reasonably well focused beams with widths below 20 \( \mu m \). Thus, we concluded that the use of a flat detector in contact with the end sphere provides a good estimate of the transverse dimensions of the beam in strongly absorbing tissue.

It should be also noted that previously we reported the focusing properties in the presence of non-absorbing tissue with a refractive index of 1.33 [32–34]. In this case the refractive index of the external medium may affect the focusing ability of the spheres. In our designs, however, this influence was minimized due to the fact that the focusing was provided primarily inside the device, so that the last focused beam was produced at the surface of the end-sphere. This is the key difference between the contact and non-contact focusing devices.

Finally, we should point out that during surgery collateral damage can propagate in tissue away from the focused beam due to emission of shock waves and cavitation bubble formation [5]. These processes, however, are beyond the scope of the present work where we determine the minimal dimensions of the light beams.

### 3. Results and Discussion

#### 3.1 Role of the Distance between the Fiber and First Sphere (d)

We start with the focusing properties of a multimode fiber in contact with a single sphere (\( d=0 \)), as illustrated in Fig. 2. It should be noted that in principle the tighter focusing can be achieved with smaller spheres. The diameter of spheres, however, should be larger than the diameter of the core of the multimode mid-IR fiber (150 \( \mu m \)) to preserve high optical throughput. The numerical ray tracing is illustrated for single spheres 300 \( \mu m \) and 150 \( \mu m \) diameters in Figs. 2(a) and 2(b), respectively. (HWG is not shown since the reflections by the HWG sidewalls are not important in such designs). Using a Gaussian approximation for the beam intensity profile the full width at half maximum (FWHM) around 17 \( \mu m \) and 10 \( \mu m \) were estimated for high index (\( n=2 \)) spheres with 300 \( \mu m \) and 150 \( \mu m \) diameters, respectively, as seen in Fig. 2(c). It should be noted that fixing the microspheres at a precise and stable position centered with the axis of the fiber core can be a rather difficult problem in such structures, especially in surgical applications where the tip of the end sphere is in contact...
with the tissue. It is interesting that in the case of bundles of single mode fibers the positions of much more compact 2 μm polystyrene spheres have been defined due to selective wet-chemical etching of a microwell array [39]. A similar approach can in principle be used for fixing larger spheres at the edge of a multimode fiber; however the refraction properties of light in such microwells and the robustness of such structures require further studies.

Fig. 2. Ray tracing for a single (a) 300 μm and (b) 150 μm sphere with n=1.9 in contact with the core of the multimode fiber. (c) Calculated FWHM of the central intensity peaks as a function of n for two structures shown in (a) and (b), respectively.

Next, we consider a typical example illustrating how the transverse intensity profiles depend on a fiber to sphere distance $d$ for a HWG structure with a single 300 μm sphere with $n=1.7$, as illustrated in Fig. 1(c). For a small separation of 1 mm the intensity distribution in the detector plane has a complicated shape with a large width of ~40 μm. It should be noted that for $d<1.5$ mm the rays do not hit the sidewalls of the HWG. The single sphere works as a thick lens with a focal distance of 32 μm. The detector plane intercepts the rays before they converge in the imaging plane, resulting in broad intensity distributions.

In contrast, for $d>5$ mm the intensity distribution is more similar to a Gaussian shape with a much narrower FWHM of approximately 13 μm. The shape of the intensity profile was found to be not strongly dependent on $d$ for $d>20$ mm, as illustrated in Fig. 1 (c). This result can be understood based on a mirror effect produced by multiple reflections at the HWG sidewalls that leads to the formation of multiple virtual sources of light. Due to a larger $d$ the images of these sources are better focused at the plane of detector where they are overlapped. Generally, this leads to the formation of an averaged envelope intensity profile which is not strongly dependent on $d$. For this reason in our further analysis we fixed $d=20$ mm in order to focus on studies of other essential parameters of the system such as $n$ and $N$. An additional advantage of this approach is that our results are independent of the way multimode beams are coupled into the HWG. For example, similar input beams can be produced by direct coupling of the multimode laser beam into the HWG without use of multimode fiber.
3.2 Focusing and Coupling Properties

In terms of energy conservation the optical microprobes considered in this work create three types of fluxes. First, they produce a focused beam with the total power $P_f$ and peak intensity $I_f$ along the axial direction. Second, in some cases they produce a broad background illumination in the forward direction with the total power $P_b$ and intensity $I_b$. Finally, a part of the incident power ($P_i$) is reflected by the spheres in a backward direction inside the HWG. In the absence of absorption $P_i= P_f + P_b + P_r$, where $P_r$ is the reflected power.

We start with a single sphere case illustrated in Figs. 3(a-c). The calculated intensity distributions were found to have an approximately Gaussian shape with FWHM presented in Fig. 3 (d). Producing small spots at the detector plane requires sharper focusing of light at the tip of the sphere. It is well known that in the paraxial approximation it can be achieved at $n=2$. Due to nonparaxial rays in the HWG the role of spherical aberrations was found to be significant, resulting in smaller indices $1.8<n<1.9$ required for minimization of the spot sizes.

![Fig. 3.](image)

Power coupling efficiency to the central focused beam ($\eta=P_f/P_i$) was estimated by using a small circular detector with radius equal to FWHM. According to this procedure, we calculated first the entire intensity profile over a broad area using a square detector, and then we used a small circular detector with the radius equal to FWHM to determine $P_f$. The calculated dependence of $\eta$ on $n$ is presented in Fig. 4(a). The single spheres provide relatively high values of $\eta\sim0.35$ in combination with FWHM~9 $\mu$m in the range of indices $1.8<n<1.9$. Although this combination of parameters is reasonable for ultraprecise surgery, the choice of the materials which can be used for fabricating microspheres with such high index of refraction and good optical properties at $\lambda\sim3$ $\mu$m is extremely limited.
The focusing properties of microprobes are found to be dramatically altered for chains of spheres, as illustrated for a five-sphere chain in Figs. 3(e-g). For \( n < 1.65 \) both three- and five-sphere chains produce relatively broad intensity profiles with non-Gaussian distributions in the detector plane. Large spot sizes are also produced by two- and four-sphere chains, as shown in Fig. 3(d) for a four-sphere chain at \( n = 1.6 \). Most importantly, three- and five-sphere chains in the range \( 1.65 < n < 1.75 \) are found to produce intensity distributions with a very narrow central peak superimposed on a much broader and weaker background \( (I_f/I_b \sim 10) \), as illustrated in Figs. 3(f,g). The Gaussian fit to the central peak shows that its FWHM can be as small as 3-6 \( \mu \text{m} \) for the range of \( 1.65 < n < 1.75 \), which is more than two times better spatial resolution over single spheres. It should be pointed out that the wavelength-scale dimensions of the focused beams indicate that we approach the limitations of geometrical optics in these designs. This greatly improved spatial resolution comes at the cost of the total power coupled to the focused beam. As shown in Fig. 4 (a), power coupling efficiency on the level of a few percent can be obtained. It should be noted, however, that the peak intensity \( (I_f) \) is attenuated in these structures only by a factor of 0.2-0.4 compared to single spheres with the same index, as shown in Fig. 4(b).

Interpretation of the observed effects is based on the fact that chains of spheres with an index in the range \( 1.65 < n < 1.75 \) effectively filter the rays which are focused in such structures with a period matching the twofold period of the structure \( (2D) \). These rays have a sharper focus and smaller propagation losses that result in a structural resonance in the optical properties with gradual decrease of the focused spot sizes. They can be considered as a geometrical optics analog of NIMs observed in chains formed by micrometer-scale spheres \[26,27\].

The role of absorption in such structures was estimated for silver sidewalls in HWGs in combination with a weakly absorbing sphere material such as sapphire in the mid-IR range. Only a minimal power loss of a few percent was observed due to absorption. Interestingly, we found that absorption can play a positive role in improving the ratio of \( I_f/I_b \) for multi-sphere structures because the rays propagating at large angles (contributing to the background illumination \( I_b \)) are most strongly attenuated by absorption effects in such structures.

4. Conclusions

In this paper, focusing multimodal microprobe designs capable of operating in contact mode with strongly absorbing tissue or samples were proposed. We modeled multimodal beams propagating in the flexible optical delivery systems used in laser surgical applications at the wavelength of the Er:YAG laser \( (\lambda = 2.94 \mu\text{m}) \) which is strongly absorbed in tissue. Our designs show more than two-fold improvement of the spatial resolution of multi-sphere microprobes over single sphere devices. It is demonstrated that transverse beam diameters of 3-6 \( \mu\text{m} \) are possible in three- and five-sphere structures with an optimized index of refraction.
of $1.65 < n < 1.75$. The focusing and attenuation properties of these structures are determined by
the mode filtering effects which have much in common with the properties of nanojet-induced
modes observed in chains of micrometer-scale polystyrene microspheres [26,27]. An
additional advantage of multiple-sphere optical microprobes is associated with the availability
of inexpensive microspheres with excellent surface quality and minimal mid-IR absorption,
such as sapphire spheres with $n=1.71$ at $\lambda \sim 3 \, \mu m$. Potential applications of these microprobes
include ultraprecise laser procedures in the eye and brain, cellular surgery, and coupling of
multimodal beams into photonic microstructures.

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