Wide-field multiphoton imaging through scattering media without correction

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Optical approaches to fluorescent, spectroscopic, and morphological imaging have made exceptional advances in the last decade. Super-resolution imaging and wide-field multiphoton imaging are now underpinning major advances across the biomedical sciences. While the advances have been startling, the key unmet challenge to date in all forms of optical imaging is to penetrate deeper. A number of schemes implement aberration correction or the use of complex photonics to address this need. In contrast, we approach this challenge by implementing a scheme that requires no a priori information about the medium nor its properties. Exploiting temporal focusing and single-pixel detection in our innovative scheme, we obtain wide-field two-photon images through various turbid media including a scattering phantom and tissue reaching a depth of up to seven scattering mean free path lengths. Our results show that it competes favorably with standard point-scanning two-photon imaging, with up to a fivefold improvement in signal-to-background ratio while showing significantly lower photobleaching.

INTRODUCTION

A suite of powerful, disruptive optical imaging approaches across the physical and biomedical sciences has recently emerged. Super-resolution imaging led to new studies looking at nanometric features within cells that have revealed intricate aspects of subcellular processes (1–5). At the larger scale, methods such as optical coherence tomography (6) and light-sheet imaging (7) are taking hold in fields such as ophthalmology, neuroscience, and developmental biology. In tandem with the requirement for a fast, wide-field visualization and super-resolved imaging across biomedicine, a grand challenge is to perform such imaging through highly scattering (turbid) media, namely, tissue. In particular, this is essential to move from superficial surface imaging to functional imaging at depth (8–11), which is crucial for biomedical areas including neuroscience. To address this area, aberration correction can be implemented (12). However, this does not readily take into account the properties of the medium, and actual retrieval of the emitted signal from depth in the medium can still be challenging. Key advances have emerged by a consideration of the propagation of light within a complex medium. In this field, a number of approaches use dynamic wavefront shaping for illumination of the sample with a calculated input complex wavefront (13–16), which can focus light upon an embedded guide star. In essence, one determines the transmission matrix of the sample in this process (11, 17, 18). While this is powerful, the requirement of a guide star restricts the approach. Furthermore, it requires determination of the properties of the medium at one or more individual points, making it very challenging to implement for wide-field imaging.

An important advance would be the realization of a fast, wide-field imaging approach that would deliver and retrieve light from any given plane within a sample, even in the presence of turbidity. This would be without the requirement to characterize or even actively correct the aberrating effect of the turbid medium. Our approach to achieving this goal exploits temporal focusing (TF) microscopy (19, 20). By using the temporal rather than spatial degree of freedom, scanning of the optical axis for image reconstruction is avoided. Consequently, TF may record wide-field multiphoton images (19, 21). In addition, a little-recognized facet of TF is its ability to deliver light through scattering media. This ability has been used to project optical patterns for applications such as optogenetics, providing photostimulation at remarkable depths (9, 22, 23). Although TF can deliver light through a scattering medium very efficiently, collecting the emitted fluorescent light back through the same medium (i.e., truly achieving imaging) has not been accomplished to date. Separately, there has been the emergence of single-pixel detection, sometimes termed computational ghost imaging (24). In this form of imaging, known patterns illuminate an object, a single-element photodetector records the light intensity that is either transmitted or backscattered by the object, and images are reconstructed with the appropriate algorithm (25, 26).

However, while these studies in TF microscopy and in single-pixel detection have shown promise, none of them has addressed the challenge of correction-free wide-field imaging through turbid media. The scheme that we present here, which we call TempoRAL Focusing microscopy with single-pIXel detection (TRAFIX), uses a judicious combination of TF illumination with single-pixel imaging to obtain wide-field images of fluorescent microscopic samples within or even beyond biological tissues, in the presence of multiple scattering, without aberration correction or characterization of the turbid medium.

RESULTS

Principle of the technique

TF is based on decomposing an incident ultrashort pulsed light field into its constituent wavelengths with a diffraction grating. Each wavelength propagates along an individual path in the optical system, and these wavelengths constructively recombine to regain the original pulse duration only at the plane conjugate to the grating, generating axially confined multiphoton excitation. In TRAFIX, orthonormal light patterns (in a Hadamard basis) are temporally focused through a turbid...
medium to illuminate a fluorescent microscopic sample of interest. The use of TF for this projection ensures the retention of the integrity of these patterns at any given plane within the turbid media (Fig. 1A). This can be regarded as due to the fact that ballistic photons remain unperturbed all the way to the object plane and arrive at the same time, contributing to the reconstitution of the pulse. In addition, the superposition of wavelets of slightly different wavelengths at the focal plane results in nearly speckle-free propagation through long distances in scattering media, as recognized by Papagiakoumou et al.\(^9\). We confirm these aspects here with a numerical simulation. The same principle has been previously used for reducing out-of-focus excitation in line-scanning multiphoton microscopy (20). A scattering medium may affect the spatial and temporal degrees of freedom of an input field differently. In the time domain, the temporal profile of femtosecond pulses is not significantly distorted at substantial imaging depths such as 1-mm-thick brain tissue (27). As a consequence, TF may induce much more efficient multiphoton excitation when compared to standard point-focusing where spatial speckle greatly reduces the photon density at the focal spot. Consequently, TF is more robust than conventional focusing, resulting in a more intense fluorescence signal generated at large depths (28), which is a major attribute for our approach.

The total intensity emitted by the fluorescent sample under each illumination pattern is collected by the same objective after passing through the scattering medium a second time (epifluorescence geometry), and the total intensity measured by a single-pixel detector. In a single-pixel measurement, the fluorescent target is sequentially illuminated with Hadamard patterns \((\psi_n)\), and the total intensity detected is stored as a coefficient \((\omega_n)\). Gray background in the second column denotes regions of zero intensity. By adding up the Hadamard patterns weighted by their respective coefficients, an image of the fluorescent sample is reconstructed.
suitable for a suite of biomedical applications. An additional microscope takes reference images of the fluorescent sample in a transmission geometry analogous to previous reports (9, 22). Reference images are taken with a CCD camera under uniform TF illumination across the field of view (FOV) (see Materials and Methods). We stress that this additional reference system is not required for imaging. Once all patterns have been sequentially projected and their intensity coefficients have been measured, images are reconstructed using an orthogonal matching pursuit (OMP) algorithm (30). The OMP algorithm determines which patterns contribute most effectively to the image reconstruction and sums them up to create an image (see the supplementary materials). The number of pixels in the retrieved image is determined by the size of the Hadamard basis used in the measurement. An \( n \times n \) pixel image requires a Hadamard basis containing \( N = n^2 \) patterns. Therefore, depending on the pixel resolution required, a different number of Hadamard patterns (typically 4096 or 1024) are encoded on an SLM. The acquisition time of the microscope is given by \( T = 2n^2(t_{exp} + t_{SLM}) \), where \( t_{exp} \) is the exposure time of the camera used as a single-pixel detector and \( t_{SLM} \) is the time required to refresh the Hadamard patterns on the SLM (including data transmission).

As TRAFIX is based on patterned illumination, it lends itself to compressive sensing measurements (26, 31). One of the main advantages of compressive sensing is that sparse signals can be reconstructed with fewer samples than required by Nyquist sampling theory. In terms of microscopy, it means that one does not need to measure with the full set of Hadamard patterns to obtain a good-quality image. The compression ratio (CR) = \( N/M \) denotes how many patterns are used to reconstruct the image in relation to the total number of patterns in the Hadamard basis (26). Here, \( M \) is the number of patterns used in the reconstruction algorithm. For example, a 64-pixel by 64-pixel image requires a measurement with a Hadamard basis containing 4096 patterns. Consequently, a CR of 2 corresponds to using only half of the total patterns to reconstruct the image (i.e., 2048 patterns), a CR of 4 uses only 1024 patterns, and so on.

As we describe below, to demonstrate the performance of TRAFIX, we imaged various microscopic fluorescent samples making use of full Hadamard bases to obtain a high-quality image. Additional compressed images were obtained a posteriori to demonstrate that compressive sensing measurements are possible in this configuration (see the supplementary materials).

**Imaging through scattering media**

To begin with, 400-nm-diameter green fluorescent beads and fixed human embryonic kidney (HEK) 293T/17 cells labeled with green fluorescent protein (HEK293T/17-GFP) were imaged through scattering phantoms, designed to mimic the scattering properties of biological tissue, and through unfixed human colon tissue. A custom-made fluorescent microstructure was then imaged through fixed rat brain tissue. As scattering dominates over absorption in the range of wavelengths considered in our investigation (32, 33), we use scattering mean free path \( l_0 \) as a reference value to quantify imaging depth. The \( l_0 \) values for the scattering media used in the experiments presented here are approximately 140, 85, and 55 \( \mu m \) for the scattering phantom, colon tissue, and brain tissue, respectively (see Materials and Methods). Full Hadamard bases of either 32-pixel by 32-pixel or 64-pixel by 64-pixel resolution were projected onto the samples, and the resulting images were reconstructed with different CRs. The lateral resolution of the microscope is defined as twice the pixel size in the reconstructed images, and thus, it depends on the FOV. Using a Hadamard basis containing 32 pixels by 32 pixels and an FOV of 90 \( \mu m \) by 90 \( \mu m \), the resolution is 5.6 \( \mu m \), and for a larger basis of 64 pixels by 64 pixels, it is 2.8 \( \mu m \) (fig. S7). We measured the depth resolution in the absence of any scattering layer for a 40\( \times \) numerical aperture (NA) = 0.8 objective to be 4.7 ± 0.5 \( \mu m \) (fig. S4) and initially tested the microscope’s performance in imaging beads and HEK cells without scattering (see the Supplementary Materials).

To quantify image quality, we measured the signal-to-background ratio (SBR) for all images presented in this work (see Materials and Methods). All values are summarized in table S1. To assess the effectiveness of this technique in photolocalization, we estimated the spacing of fluorescent beads and the size of cells in the reference image and in the reconstructed images and calculated their deviation from the reference value (see the Supplementary Materials).

Figure 2 shows images of the fluorescent beads and HEK cells obtained through 500 and 540 \( \mu m \) of scattering phantoms (3 to 4 \( l_0 \)) respectively. As a result of the strong scattering, very few illumination photons reach the sample plane, generating a very dim fluorescence signal; therefore, a long exposure time or large binning is required to even obtain a good reference image under uniform TF illumination in transmission. In standard TF microscopes, those few emitted fluorescence photons would need to travel back through the thick scattering medium, losing all their spatial information and hence resulting in uniform noise on the camera (29). Consequently, the SBR would essentially be unity, which means that no signal can be extracted from the background noise. However, in TRAFIX, there is no need to measure spatial information as total intensity coming from the sample under each illumination pattern is detected and used as a coefficient in the reconstruction algorithm. This results in a significant enhancement in signal detection and higher SBR values.

The achievable level of compression depends on the sparsity of the image. If an image contains very little information, then high compression is possible with negligible information loss. As the images of fluorescent beads are sparse, image quality is preserved for large values of CR. A good image can be faithfully reconstructed with CR = 8 (12.5% of the total number of patterns). Similar CRs have been achieved in the literature, although with different optical imaging systems (26). Nevertheless, we did not achieve a high compression with cells in the present configuration. They covered a larger area of the image and thus cannot be compressed as much without a significant loss of information with the present algorithm and pixel resolution. In addition, the 800-nm excitation laser used in the experiments is not optimal for excitation of GFP, which has its highest absorption in the range of wavelengths between 870 and 920 nm (34). As a result, the cells appear very dim. Despite these issues, an SBR exceeding 4 was achieved at CR = 2 (i.e., using only 50% of the full Hadamard basis in the image reconstruction algorithm).

In the next stage, the performance of TRAFIX was tested with unfixed human colon tissue as a scattering medium. One of the main problems in imaging through biological tissue is autofluorescence. Coda et al. (35) showed the single-photon excitation spectrum of human colon tissue at different wavelengths. Under 435-nm single-photon excitation, colon tissue presents a very intense autofluorescence signal at the green part of the spectrum, overlapping with the light emitted from the beads or cells used in our experiment. Single-photon excitation at 435 nm is relatively similar to two-photon excitation at 800 nm, so it poses a big obstacle in the experiments because it reduces the number of photons that can reach the beads or HEK cells and it also generates undesired background light. The optical sectioning capability...
Fig. 2. Images of fluorescent microscopic samples through scattering phantoms. Fluorescent beads of 400 nm in diameter and fixed HEK293T/17-GFP cells were imaged through 500- and 540-μm of scattering phantoms, respectively. (A and F) Images taken from the reference imaging system under uniform TF illumination across the FOV. Exposure time was set to 20 s, and camera binning was 4 × 4 (beads) and 2 × 2 (cells). (B to E, G, and H) Images obtained in epifluorescence configuration with TRAFIX using a Hadamard basis containing 4096 illumination patterns. They were reconstructed with different CRs corresponding to 100% (CR = 1), 50% (CR = 2), 25% (CR = 4), or 12.5% (CR = 8) of the total patterns. Each measurement under individual illumination patterns was taken with a binning of 64 × 64 and an exposure time of 0.5 s. The spacing between beads was measured in all five images obtaining deviations smaller than 3% from the reference image (table S3). The diameters of the cells in (F) were measured to be 20.7 and 14.3 μm, respectively, and their values in (G) and (H) differ less than 4 and 12% from the reference value (table S2). The SBR is shown for all reconstructed images. Scale bars, 10 μm.

Fig. S12 shows how the focused spot of 2PM and various TF Hadamard patterns of TRAFIX get distorted at different depths through unfixed human colon tissue. After 400 μm, the 2PM focused spot is not discernible and turns into a complete speckle pattern. At the same depth, low-frequency Hadamard patterns retain their shape reasonably well, although having substantial intensity inhomogeneities. These results align well with the data presented by Rowlands et al. (22) using...
optical paths produce only an anisotropic smoothing. As a result, large paths produce an overall smoothing of the beam, while slightly different tissue, is transformed into a spatial speckle pattern, as expected. However, one important feature of TF is that each monochromatic portion going through a highly scattering layer, such as a 400-μm-thick medium, is basically speckle free, scattering of light in this case does not heavily distort the imprinting process. Larger intensity variations are due to inhomogeneities in light transmission through the highly anisotropic scattering medium. This also applies to figs. S7, S9, S10, and S12. The SBR is shown for all reconstructed images.

**Numerical simulation**

To illustrate how TRAFIX behaves in scattering media, we performed a simplified one-dimensional simulation of the imaging process. We took into account the propagation of TF Gaussian beams through a scattering medium and the detection of fluorescent light after backward propagation through the same medium (see the supplementary materials). Our simulation shows that monochromatic light propagating through a highly scattering layer, such as a 400-μm-thick brain tissue, is transformed into a spatial speckle pattern, as expected. However, one important feature of TF is that each monochromatic portion of the beam propagates through a different optical path. As identified by Papagiakoumou et al. (9), waves traveling with very different optical paths produce an overall smoothing of the beam, while slightly different optical paths produce only an anisotropic smoothing. As a result, large features of the initial beam profile are remarkably well conserved as shown in Fig. 3A. Although the main source of two-photon excitation in TRAFIX is typically generated by ballistic photons, at large depths, the excitation caused by scattered photons becomes important when compared to the highly attenuated ballistic light. As the TF beams are basically speckle free, scattering of light in this case does not heavily distort the Hadamard patterns but, in fact, contributes into making them brighter with the only disadvantage of generating softer edges in the features of the pattern.

In the second part of the simulation, we estimated the total fluorescent light that would reach our detector with respect to the total laser power deposited on the sample surface. It is clear to see in Fig. 3B that it decays markedly as the thickness of the scattering medium increases. Owing to single-pixel detection, in our experimental measurements, we could use those very few photons to form images through fixed rat brain tissue of up to 400 μm thickness.

**DISCUSSION AND CONCLUSIONS**

The combination of patterned TF illumination with single-pixel detection achieves remarkable imaging depths for wide-field multiphoton
microscopy as it provides a way of exciting fluorescent structures deeper inside turbid media than existing imaging techniques and can efficiently collect the emitted light in an epifluorescence configuration. We have demonstrated the effectiveness and potential of TRAFIX by imaging fluorescent beads of 400 nm in diameter and fixed HEK293T/17-GFP cells through a layer of a scattering phantom with a thickness of more than 500 μm (~7 Ls) without any aberration correction, guide star, or detector placed in/beyond the turbid media. We then imaged a bright custom fluorescent microstructure through rat brain tissue of hundreds of micrometers in thickness, reaching a maximum imaging depth of ~7 Ls. In addition, we showed that TRAFIX works well under typical biological research conditions by imaging both fluorescent beads and HEK cells through depths of more than 3 Ls of unfixed human colon tissue and even in the presence of intense background fluorescence.

The main factor that limits imaging depth of TRAFIX is the penetration depth of the TF Hadamard patterns. Propagation of TF beams through very large distances in scattering media results in distortions on the illumination patterns mainly caused by refractive index changes in the sample. These distortions generate a basis mismatch that may result in deformities in the reconstructed images (38). In addition, the use of a one-dimensional diffraction grating generates horizontal shifts in the illumination Hadamard patterns that could potentially be minimized by dispersing the patterns isotropically. A future embodiment using dispersion in two dimensions with two perpendicular diffraction gratings would be an improvement. In the present embodiment of TRAFIX, the acquisition speed is limited by the exposure time of the EMCCD camera and the slow refresh rate of the SLM (see the Supplementary Materials). A typical image obtained using a full basis of 1024 patterns (32 pixels by 32 pixels) through a moderate thickness of a scattering sample currently takes ~5 min. This time is increased to up to 1 hour when imaging in high resolution (64 pixels by 64 pixels) with a full basis through the most challenging conditions shown in this article. To speed imaging up, the EMCCD camera would be replaced with a fast, sensitive photodetector such as a photomultiplier tube (PMT) and the imaging speed would no longer be limited by the exposure time of the detector. In addition, as TRAFIX currently uses binary Hadamard patterns, the SLM may be replaced with a significantly faster digital micromirror device, which may run at tens of kilohertz. These changes, combined with the new advances in compressive imaging, suggest that frame rates for
128-pixel by 128-pixel images can be increased to more than ~30 Hz (43, 44), enabling studies in time-varying turbulence (45).

Since our scheme can be easily implemented in a standard multi-photon microscope, we believe that one of its main applications will be in the field of optogenetics where it would lend itself to achieve long-term simultaneous imaging and photoactivation of neuronal networks with minimal photodamage deep inside the brain. Last, as the polarization state of the illumination light does not change over propagation in scattering media through the range of distances normally considered for imaging (see the Supplementary Materials) (46), TRAFIX could also be combined with polarization-resolved imaging techniques (47).

In summary, TRAFIX is a novel approach for deep multiphoton imaging that presents an increased SBR compared to the ubiquitous 2PM while also reducing photobleaching of the sample. In addition, the almost speckle-free propagation of TF illumination patterns suggests that TRAFIX may surpass the maximum imaging depth limit of 2PM and may be very beneficial for long-term biological studies, particularly in neuroscience.

**MATERIALS AND METHODS**

**Experimental setups**

**TempoRAI Focusing microscopy with single-pixel detection**

An illumination laser (Coherent Chameleon Ultra II) delivers 140-fs pulses with an 80-MHz repetition rate, up to 4-W average output power at a variable central wavelength between 680 and 1080 nm. The central wavelength of the laser was set to 800 nm for all the experiments performed. The illumination beam was expanded four times to cover the active area of a phase-only SLM (LCOS-SLM, Hamamatsu Photonics). The SLM was then imaged onto a blazed reflective grating (1200 g/mm) with a 4f (f = 400 mm) telescope to create wide-field TF illumination. The first diffraction order from the SLM was transmitted through an iris in the telescope, while all other orders were blocked. The beam was diffracted from the grating, and all wavelengths were collimated with an f = 400 mm lens relayed onto the back focal aperture of the illumination objective. Two different illumination objectives were used in this work. A water dipping objective (40× NA = 0.8; Nikon), which is enclosed in a custom-made chamber filled with water, generates a TF illumination plane with a size of 90 μm by 90 μm. The highest average laser power per unit area used in this configuration is 64 ± 5 μW/μm². An air objective (20× NA = 0.75; Nikon) was used for additional studies presented in the Supplementary Materials, as accordingly specified. Backscattered fluorescent light propagated through the turbid media and was collected by an EMCCD camera run without amplification (iXon EM+ 885, Andor Technology) via the same illumination objective in epifluorescence configuration. To provide reference images for this paper, forwardly emitted photons from the sample were collected by a CCD camera (Clara, Andor Technology) in transmission via a long working distance air objective (100× NA = 0.7; Mitutoyo). Appropriate short-pass filters were used to reject the illumination laser at 800 nm and transmit fluorescence below 700 nm. In contrast to other single-pixel imaging approaches (25, 26), the EMCCD camera with 64 × 64 binning was used as a bucket detector instead of using a single-element detector such as a PMT or an avalanche photodiode. Using high binning helps reducing the effect of readout noise. All objectives, samples, and cameras were attached on the body of an inverted microscope (Eclipse Ti, Nikon) accordingly.

**Point-scanning two-photon microscope**

The 2PM shares the same setup as TRAFIX, except for the diffraction grating and a lens that are replaced with a mirror to obtain a focused beam on the focal plane of the illumination objective. A Nikon 20× NA = 0.75 air objective was used for all experiments. A variable iris was used to adjust the size of the focused spot. An X-Y-Z motorized stage (Nanomotion-LP200, Mad City Labs) scanned the sample in a stepwise motion across the focused beam covering the entire FOV. The same binned EMCCD camera run with no amplification (iXon EM+ 885, Andor Technology) collected the fluorescent light emitted by the sample.

**Fluorescent and scattering samples**

**Fluorescent layer and fluorescent micropatter**n

A 200-nm-thick layer of super-yellow polymer spin-coated on a glass coverslip was used to characterize the profile and depth resolution of the TF beam (see the Supplementary Materials). It was also used to generate a fluorescent micropattern that was then imaged through scattering media. The negative of a pattern of interest was encoded on the SLM, and the thin fluorescent layer was placed at the focal plane of the microscope without any scattering layer. The laser power was set to the maximum, and the negative pattern was photobleached on the fluorescent film. Therefore, the only portion of the FOV that remained fluorescent was exactly the desired micropattern.

**Fluorescent beads**

Green fluorescent polymer microspheres (G400, Duke Scientific) with a diameter of 0.39 μm were used to test the performance of the imaging system. A very small amount of beads was deposited on a glass coverslip and placed on top of the scattering samples to image them through the turbid media.

**HEK cells**

HEK293T/17-GFP cells were used to demonstrate the capability of the microscope in imaging real biological samples through scattering. HEK293T/17 cell line obtained from American Type Culture Collection was cultured in Dulbecco’s modified Eagle’s medium GlutaMAX-I supplemented with 10% fetal bovine serum and 1% penicillin/streptomycin and was transfected using TransIT-LT1 transfection reagent with the Vesicular Stomatitis Virus glycoprotein (VSV-G) pseudotyped lentivirus vector and the packaging plasmid psPAX2 to deliver the plasmid pLenti-GFP-Puro. Regarding the sample preparation, the HEK293T/17-GFP cells were replated on World Precision Instruments FluoroDish poly-D-Lysine–coated cell culture dishes at a low density to achieve ideal imaging conditions. The day after plating, before imaging, the HEK293T/17-GFP cells were fixed in a phosphate-buffered saline 4% paraformaldehyde solution. The scattering samples were attached directly on the bottom of the dishes containing the cells.

**Scattering phantom**

Polystyrene beads of 1 μm in diameter were used as scatterers to simulate a turbid sample. The beads were purchased in a 1% concentration solution in water (Microbead NIST Traceable Particle Size Standard, 1.00 μm; Polysciences). The solution was thoroughly stirred in a vortex mixer and then mixed with a 1% solution of agarose in water (preheated above melting point). Agarose and beads were mixed in the vortex mixer again and placed into sample wells of variable height. The wells consisted of a 100-μm glass slide with multiple 90-μm vinyl spacers stacked on top of each other. An additional coverslip was placed on top to seal the well. The concentration of polystyrene beads in the sample was chosen to roughly match the scattering coefficient of real biological tissue (48). Using an on-line Mie scattering calculator (49),...
we determined the reduced scattering coefficient of our phantom to approximately be \( \mu_s' \approx 7.5 \text{ cm}^{-1} \), corresponding to a mean free path of about \( l_s = (1 - g) / \mu_s' \approx 140 \mu m \), where \( g \) is the anisotropy factor \( (g \approx 0.9 \text{ for most biological tissue at the wavelength considered in this investigation}) \). Different scattering phantoms were used in some experiments in the Supplementary Materials as appropriately specified.

**Rat brain tissue**

Similar to our previous work (50), rat brain tissue was obtained from adult Sprague Dawley rats, in accordance with the UK Animals (Scientific Procedures) Act 1986. It was fixed and then sectioned into slices at thicknesses of 200 and 400 \( \mu m \). The mean free path of the rat brain tissue was estimated by measuring the ratio of the incident laser intensity \( I_0 \) with the intensity of the ballistic photons \( I_b \) and by applying an exponential law \( I_b = I_0 \exp(-L/l_s) \), where \( L \) is the thickness of the brain tissue (18). The obtained value of \( l_s \approx 55 \pm 9 \mu m \) is consistent with other measurements reported in the literature for rat (51) and mouse brains (52).

**Colon tissue**

Thin fragments of unfixed normal human colon tissue were used as scattering samples. They were stored in a freezer at –80°C and mounted between two coverslips. Their thickness ranged from 200 to 250 \( \mu m \). The mean free path of the colon tissue was estimated by measuring the ratio of the incident laser intensity \( I_0 \) with the intensity of the ballistic photons \( I_b \) and by applying an exponential law \( I_b = I_0 \exp(-L/l_s) \), where \( L \) is the thickness of the brain tissue (18). The obtained value of \( l_s \approx 55 \pm 9 \mu m \) is consistent with other measurements reported in the literature for rat (51) and mouse brains (52).

**Image quality quantification**

**Signal-to-background ratio**

SBR is defined here as \( \text{SBR} = \mu_{\text{sig}} / \mu_{\text{bg}} \), where \( \mu_{\text{sig}} \) and \( \mu_{\text{bg}} \) are the average values of the signal and the background, respectively. In the case of cells and fluorescent microstructures, two small regions of interest were defined—one containing fluorescent structures and the other corresponding to the background. In the images of beads, the highest-intensity pixels in the beads were used as signal value, and the maximum intensity pixels in the rest of the image were used as background noise. Uncertainty is given by the SD of all averaged measurements. All values of SBR presented in this work are summarized in table S1.

**Cell size**

As the cells used here appear approximately spherical, their diameter was estimated by taking a measurement of their size in two orthogonal directions and averaging the obtained values. Their diameter was determined by fitting a Gaussian function to the intensity values and measuring its full width at half maximum. Table S2 shows the diameters of the cells that appear in the different images presented in this work. Deviations in cell size between reconstructed and reference images are expressed as a percentage error.

**Bead spacing**

The central pixel for each individual bead was located, and the distance between beads was measured in the reference image and in all retrieved images with different CRs (table S3). Deviations in bead spacing between reconstructed and reference images are expressed as a percentage error.

**REFERENCES AND NOTES**


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