Single fiber ghost imaging for extreme minimally invasive medicine

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Abstract

Optical scattering media, such as blood, disturbs us to diagnose diseases inside deep blood vessels of our body. This is one of age-old problems on optics. To date, optical correlation allows us to image any objects hidden by scattering media. However, constrained by the scattering condition and weak illumination with thought of our body safety, even advanced fiber endoscopes are impossible to image the objects in meso-field of a few 10 mm. To overcome this problem, we demonstrate a lens-less single fiber ghost imaging for extreme minimally invasive medicine. Present imaging with a diameter of 105 µm achieved a spatial resolution of 0.05 mm in observing area of 9 mm², image reconstruction under opaque scattering condition, and imaging at laser power density of 0.10 mW/cm², as compared with a conventional neuroendoscope at that of 94 mW/cm². Our strategy opens a new avenue for extreme minimally invasive endoscopic medicine.

Introduction

Advanced optical endoscopy achieving remarkable progress anticipates to obtain epoch-making images by tailoring a nature of light, such as broadband spectrum, coherence, and even polarization [1–4]. Such optical endoscopy helps us to diagnose disease inside deep blood vessel of our body, and to assist surgical guide [5, 6]. Their application has been widely expanded not only medical diagnosis but also industrial inspection [7]. To date, a lens-less fiber imaging has been attracting attention to enable minimally invasive diagnosis [8]. There are two solutions for minimally invasive fiber imaging, using a multimode optical fiber and an optical fiber bundle [9, 10]. Ultrathin fiber imaging with a diameter below 100 µm has been proposed with a conception related light shaping through a multimode optical fiber [10–14]. Thin fiber imaging has been demonstrated to create an image by analyzing light wave propagation in multimode optical fiber, and solving their inverse problems. This fiber imaging is capable of recording dynamic changes of macroscopic objects in far-field of several 100 mm. On contrary, holographic lens-less optical fiber bundle endoscope with a diameter below 1 mm, which consisted of several 10,000 optical fibers, achieved microscopic imaging of unstained biological tissues by high spatial resolution of 0.89 µm in close proximity of several 100 µm from a top of the fiber instrument [15–18]. Such fiber imaging becomes a candidate of endoscopy for rapid medical diagnosis.

However, even innovative research is limited to image “far field” or “close proximity” from the distal end of their optical fiber probe, because of speckle artifacts [9–18]. Meso-field between several mm and several 10 mm from the fiber probe is an unexplored region, which is unable to be captured by conventional fiber imaging, on single fiber imaging. In particularly, angioscopy in catheter treatment is often required to observe objects in the meso-field below several mm² at distance below few 10 mm from the end of the optical fiber [19]. Furthermore, conventional angioscopy using optical coherent tomography and small CMOS camera needs saline flush due to blood removal [20]. However, the flush has a risk that causes rapid changes of intravascular pressure inside thin blood vessel attached a plaque. Therefore, conventional optical angioscopy could not apply to an above-mentioned clinical environment, because of
high invasiveness. Optical angioscopy without saline flush such as an intravascular ultrasound has been strongly required, and thus one of remained challenges on advanced optical endoscopies is a resistance of imaging deterioration through opaque scattering condition, such as blood.

For this age-old problem, speckle correlation using rigorous phase retrievals proposed as predominant approach toward imaging under scramble light field [21]. Around the same time, differential ghost imaging, which achieved to create images with enhancing the signal-to-noise ratio from covariance between speckle pattern and scattering light, has been proved their superiority of image reconstruction through scattering media [22, 23]. Their fascinating studies were expected to open a next-generation optical fiber endoscopy breaking through the fundamental problems of optical endoscopes. Multi-core fibers illumination also restored images with an external detector [24–28]. However, their present studies are limited to create image by optical system different from an endoscope, because of difficulty of illumination stability. To the best of our knowledge, there are no reports of single fiber imaging under scattering media being achieved. Moreover, a small aperture on single fiber imaging makes it more difficult to collect photons back-scattered from an object. Illumination with high power density conversely causes burn injury on our body. Therefore, it is important for single fiber imaging to create high contrast image by weak illumination with low power density. Hadamard transform imaging also clarified to retrieve image under low signal-to-noise ratio condition [29]. This credible research is one of keys for extreme minimally invasive single fiber imaging with our body safety. For this background, we focused on solutions for unexplored region on ultrathin fiber imaging, power density of illuminating light, optical scattering problem for extreme minimally invasive medicine.

Here, we demonstrate a lens-less single fiber ghost imaging for extreme minimally invasive medicine. We clarified stability and relative coincidence of structured light through an optical fiber. For reconstructed images, spatial resolution of our imaging was determined. Compared with a neurosurgery fiber endoscope, we also investigated effectiveness of our imaging under extreme weak illumination. Finally, we show our imaging under opaque scattering condition. Our manifesting results promise a potentiality to solve age-old problems of optical endoscopy in clinical medicine.

**Results**

**Principle of single fiber imaging**

To achieve a fiber imaging, structured light, which enables us to register spatial coordinates, requires to form after passing through a multimode optical fiber. We noticed speckle pattern as their structured light although speckle has rather been treated as a nuisance in previous research of single fiber imaging [9–18]. Figure 1 shows a concept of our single fiber imaging for extreme minimally invasive medicine. Light generated from a pseudo-thermal light source is incident into a rotating diffuser. The scattered light after the diffuser is collected, and focused on a top of a multimode optical fiber by an objective lens. Speckle pattern generated after passing through the multimode optical fiber helps us to obtain spatial code. Light beam with the speckle pattern then passes through a diffuser. Here, the light beam is separated the back-
and forward-scatterings depicted by $b_s$ and $f_s$, respectively. Forward scattering light is illuminated on the object. Their scattered light by the object illuminates the diffuser, again. The scattering light represented by $f'_s$ is corrected by the multimode optical fiber. The light signal is detected by a single pixel detector. The essence of our single fiber ghost imaging is to detect the components of the speckle pattern scattered by the diffuser that are not scattered but propagate straight ahead by spatial sorting using optical fibers and extreme weak light detection of the ghost imaging. In our procedure, preregistered speckle patterns and light signal are correlated by a computer. Generated image was stored in memory on the computer. After we obtain two images $G_K$ with and without the object, the image of $G$ is retrieved by subtracting $A$ with $B$. Finally, the contrast of the object image is also improved by normalizing by the maximum and minimum values on the retrieved image.

Consider their speckle pattern control, as schematically shown in Fig. 2a. (see Materials and Methods and Fig. S1 for experimental setup) We employed a laser diode (LD), which is controlled by a thermal controller (TC), at central wavelength of 840 nm with spectral bandwidth of 1 nm. The light beam is incident into a rotating diffuser (RD) with precise mechanical control. We inserted an isolator composed by a Glan-Thompson prism polarizer (P) and a Fresnel rhomb (FR) to block their returned light from the diffuser. The light scattered by the rotating diffuser was collected by an objective lens with magnitude of $m = 20$ (OL), and was incident into a multimode optical fiber with a diameter of 105 mm and a numerical aperture of $NA = 0.22$. We employed a pair of optical fibers with two branch multimode optical fibers (MMF), due to avoid returned light from the end face of the optical fiber and an optical circulator. The diffuser was mounted by a rotation stage with resolution of 0.01°. We applied the output laser power of 0.287 mW through the optical fiber. From this experimental setup, we produced different speckle patterns at every angle of the rotating diffuser.

In advance, we conducted registration of speckle patterns with $I_n(x, y)$ using a two-dimensional sensor (2DS) at a distance of $z$ from a top of the optical fiber, where, $n$, $x$, and $y$ are indicated numbers and coordinates, respectively. The CMOS camera enables us to capture speckle patterns in time of 25 frame/s. By the rotating diffuser with every rotation angle of 0.06° during 4 min, we captured different speckle patterns with numbers of $n = 6,000$. To obtain more speckle patterns, the $x$-axis stage was used. By a displacement of the stage, the scanning area on the diffuser was slightly shifted. For this manipulation, we created different clusters of speckle patterns with $n = 6,000$. (see speckle patterns recorded in Supplementary mov. 1) After this registration, we replaced the sensor with a sample. By illuminating the same speckle patterns as the registration on the sample, the scattering light was passed through the other optical fiber of twin optical fibers, and detected the scattering intensity of $B_n$ by an avalanche photodiode (APD). A reconstructed image of $G$ by a ghost imaging, which uses calculus of optical correlation, is expressed as $G = \langle \Delta I_n(x, y) \cdot \Delta B_n \rangle$, where $\langle \rangle$, $D I_n(x, y)$, and $DB_n$ denote an ensemble average, intensity fluctuations of speckle patterns, and scattering light detected by the APD, respectively. Our ghost imaging was improved a differential ghost imaging to be suitable for an extreme minimally invasive single fiber imaging. (See Materials and Methods).

**Test for spatial resolution**
To estimate spatial resolution of this imaging, we measured their speckle sizes at \( z = 10 \) mm and 20 mm in Fig. 2b. Each speckle size was diameters of \( 0.08 \pm 0.01 \) mm and \( 0.13 \pm 0.01 \) mm, respectively. These results represent spatial resolutions of this imaging. In our experiments, we also evaluated a time-stability and difference by numbers in every speckle pattern. Figure 2c show a time-stability of the speckle pattern during 30 min. The speckle pattern was compared with an image at a time of \( t = 0 \) s. The speckle pattern during 4 min was stable with coincidence degree of 0.98. The coincidence degree between images was calculated by \( C = \frac{S_{AB}}{S_A \cdot S_B} \), where \( S_{AB} \), \( S_A \), and \( S_B \) denote a sample covariance between images of A and B, and standard deviations of two images of A and B, respectively. Note that we defined the coincidence degree to avoid a confusion with optical correlation in the ghost imaging. It is necessary for our imaging to be able to distinguish speckle patterns in each recorded image. We computed their coincidence degrees using 6,000 images, which are recorded different speckle patterns, at \( z = 10 \) and 20 mm as shown in Fig. 2d. Their averages of coincidence degrees at \( z = 10 \) and \( z = 20 \) mm were 0.14 and 0.04, respectively. Their results prove that our structured light generator suppresses memory effect of their speckle patterns.

Moreover, we evaluated a spatial resolution of our endoscopic fiber imaging. We herein fabricated two 0.5 mm-squares as a sample, and measured relative length of centers on two squares as the function of displacement given by a micrometer. (See Materials and Methods for sample preparation and Fig. S2 and S3) Note that their material of samples was employed an infrared reflective tape, which is in use of a conventional infrared marker, due to obtain scattering light. Figure 3a show two images with 100 × 100 pixels at micrometer position of \( x = 0.00 \) and \( 0.05 \) mm. During displacement of 1.00 mm, we measured the position of center on the right square. (see Fig. S4 for reconstructed images with displacement at every 0.05 mm pitches) Fig. 3b shows results of the measured displacement. For this result, we were reconstructed endoscopic fiber imaging with the spatial resolution of 0.05 mm. This resolution was accorded with the speckle size of 0.07 mm. Note that the spatial resolution was limited by the wavelength of laser diode, a numerical aperture of the optical fiber, and a pixel size on CMOS. (see Fig. S5 for the relationship of speckle size between wavelength and fiber diameter.)

**Comparison with conventional bundle fiber imaging with lens**

To demonstrate single fiber imaging of another sample with a complex shape, we fabricated a cross of 2 mm as a sample as shown in Fig. 4a. (see Materials and Methods and fig. S2) The sample was placed at \( z = 10 \) mm from the top of the optical fiber. To Compare with a conventional endoscopic image, we captured an image using a neuroendoscope (NEU-4, Machida Endoscope Co., Ltd), which is bundled by optical fibers of 5,000 with a fiber diameter of 400 nm and a lens on top of the fiber bundle, as shown in Fig. 4d. We the conducted single fiber ghost imaging of the cross using recorded speckle patterns. Figures 4c–4e show results using speckle patterns of 1,000, 10,000, and 30,000, respectively. (also see Fig. S6 for comparison with images reconstructed between GI and differential GI) Fig. 4e became sharpen edge on the cross as compared with Fig. 4b. In Fig. 4f, we determined the coincidence degree between their reconstructed images by our method and a conventional endoscopic image in Fig. 4b. Here, two
coincidence degrees were depicted by red triangles and blue squares, respectively. A reconstructed image and conventional endoscopic image of squares were shown in Figs. 4f1 and 4f2, respectively. Maximum coincidence degree of the cross and the square were rose up to 0.70 and 0.78, respectively. This value implies a strong correlation [30]. Their coincidence degrees were calculated with conventional endoscopic fiber images shown in Figs. 4b and 4f2, respectively. Note that the images shown in Figs. 4b and 4f2 were used as standard images for calculation of coincidence degree, in spite of not clear image. Laser powers used at the conventional endoscopic imaging and our method were 105 mW and 0.287 mW, respectively. Surprisingly, our fiber ghost imaging enables reconstructing images at extreme low laser power.

**Extreme minimally invasive single fiber imaging**

To investigate our single fiber ghost imaging by extreme weak illumination, we fabricated a letter of “s”, as shown in Fig. 5a. (see Materials and Methods and Fig. S2) Compared with the above-mentioned conventional endoscopic fiber imaging, we captured the sample at power densities of 94 mW/cm² and 1800 mW/cm², as shown in Figs. 5b1 and 5b2, respectively. The intensity on Fig. 5b1 was very weak. Moreover, image quality on Fig. 5b2 was also not good. Then, we conducted single fiber ghost imaging of the letter of “s”. Figures 5c-5e show images retrieved at power densities of 0.1, 1.0, and 2.5 mW/cm², respectively. In Fig. 5f1, we show coincidence degree as compared between the reconstructed image and Fig. 5b2. Note that red circles were normalized by common maximum and minimum in the power density from 0.10 to 0.50 mW/cm². On contrary, blue circles were calculated by individual maximum and minimum, respectively. At power density of 0.10 mW/cm², their maximum coincidence degree was 0.79. This value also implies a strong correlation [30]. (see Fig. S7 for comparison of images reconstructed between DGI with common and individual normalization) Moreover, their contrast noise ratios (CNR) obtained by $\text{CNR} = \left( \frac{\bar{A} - \bar{M}}{\sqrt{\sigma_A^2 + \sigma_M^2}} \right)$ were also determined in Fig. 5f2. Here, $\bar{A}$ and $\bar{M}$ denote average intensities on an aperture and a mask of the cross. $\sigma_A$ and $\sigma_M$ are standard deviation of their intensities, respectively. The maximum of CNR was 1.50 at the laser power density of 0.10 mW/cm². Interestingly, our endoscopic fiber image achieved to create an image by the extreme weak illumination with magnitude of 1/940. Normalization on our imaging method was effective for improvements of the coincidence degree and CNR of our imaging with extreme weak illumination.

**Single fiber imaging through opaque scattering media**

One of strong points of the ghost imaging is a resistance of imaging deterioration on scattering condition [18]. To obtain image reconstructed through a scattering media, we build up an optical layout shown in Fig. 6a. In this experiment, we set a sample of 0.5 mm-square and a CMOS at $z = 20$ mm, and inserted a diffuser sheet (#No. 4, Renian) at $z = 10$ mm. Note that the distance to the sample is different with previous experiment setup shown in Fig. 2, because of insertion of the diffuser. For comparison of our imaging, we observed the square using a conventional endoscope. Figure 6b shows the neurosurgical endoscopic image under a halogen lamp. Note that the image quality was not good because of low
spatial resolution of this endoscope. Figure 6c1 shows an image of the sample reconstructed after passing through the diffuser. This image of \( I \) was calculated by \( I = I_A - I_B \) from two images, where \( I_A \) and \( I_B \) denote their images through the diffuser with and without the sample as shown in Fig. 6c2 and 6c3, respectively. Note that the retrieved image shown in Fig. 6c1 lost sight of the sample, because of influence on scattering media.

To realize a single fiber ghost imaging with a resistance of imaging deterioration by scattering condition, we firstly captured an image without the squared sample through the diffuser (see Fig. S2). After obtained the image without the sample, we then captured the image with a square sample. Thirdly, the image \( G \) of the sample through scattering media was calculated as \( G = G_A - G_B \), where \( G_A \) and \( G_B \) indicate their reconstructed images with and without the sample. (see Materials and Methods for reconstructed image under scattering condition) Figs. 6d1-6d5 show images calculated by our imaging using speckle patterns registered of 1,000, 10,000, and 30,000 respectively. The image reconstructed by 1,000 speckle patterns was nothing identifiable. A part of the sample was started to see, as shown in Fig. 6d3, and an appearance of the sample was clearly observed in Fig. 6d5, respectively. Compared with the conventional image shown in Fig. 6b, we calculated the coincidence degree. Figure 6e shows their coincidence degree at numbers of recorded speckle patterns. Maximum coincidence degree arrived at 0.45 using 30,000 speckle patterns. This value implies a moderate correlation \([30]\). In spite of the coincidence degree was not strong, the shape of the image reconstructed in Fig. 6d5 was well similar with the microscopic image without the diffuser in Fig. 6f. Note that a center of the sample was low scattering under the optical microscope. Using more speckle patterns, we expected to obtain an image with high coincidence degree. To evaluate image under optical scattering condition, we also employed another diffuser sheets with different scattering levels. (See Figs. S8, S9, and S10 for reconstructed images under scattering condition.)

**Discussion**

We demonstrated a single fiber ghost imaging for extreme minimally invasive medicine. By different speckle patterns, spatial resolution of our fiber imaging was 0.05 mm at \( z = 10 \) mm. This spatial resolution depends on their speckle size of illuminating light. To generate light with small speckle pattern, we just have to choose a laser source with short wavelength and/or a use of a multimode optical fiber with large diameter. Employed a laser with wavelength of 520 nm, presented speckle size becomes to be 55 mm in the diameter. According to experimental results using our ghost imaging shown in Fig. 3, spatial resolution of the single fiber ghost imaging becomes small since their spatial resolution was improved by analyzing multiple speckle patterns. For our experiment shown in Fig. 2b1, the spatial resolution of the fiber imaging was 5/7 times of the speckle size. When the optical fiber with a diameter of 200 mm is also employed at wavelength of 520 nm, we estimated the spatial resolution of our fiber imaging becomes 25 mm by calculating the speckle size in Fig. S5e. Such estimated spatial resolution is sufficient for single fiber imaging.
In extreme minimally invasive medicine, strong points of our strategy state imaging in a clinical meaningful meso-field, image reconstruction under scattering condition, and extreme minimally invasive single fiber imaging with ultra-thin diameter and weak illumination. Such strong points are suitable for observation inside small blood vessels in our brain and heart. Imaging on meso-field is important for plaque removement and catheter ablation. In extreme minimally invasive diagnosis, it is necessary not only to be used ultra-thin fiber imaging but also to image an object under optical scattering condition without saline flush. Moreover, we also achieved single fiber imaging by extreme weak illumination with power density of 1/940 as compared with the conventional endoscope.

In the image reconstruction through a diffuser, reconstructed image was improved according with numbers of speckle patterns, as shown in Fig. 6. In this stage, however, we took 20 min to generate 30,000 speckle patterns. This is limited by stepping motor control used in the rotating diffuser. Using a rotation stage, which is controlled by a DC servo motor, with a high precision encoder and a high repetition pulse laser, our fiber imaging enables us to reconstruct an image of an object during speeds from 0.1 to 10 frame/s. Moreover, we expected that stable silicon photonics phased array [26] and a high-speed spatial light modulator [31] are utilized for high-speed fiber imaging. When we measured a cross in Fig. 6, the cross was coincidence degree of 0.70 using speckle patterns recorded of 30,000. For more improvements of the imaging quality and reduction in number of speckle patterns, using deep learning is a potentially effective approach [32–34]. Though employed single fiber composed twin fibers for illumination and detection in this experiment, we can also realize single core fiber imaging by introducing an optical circulator with anti-reflective coatings. Moreover, quality of the image reconstructed under the optical scattering condition would be further improved if detection at a time gate could be introduced by stabilized pulsed lasers.

Our fiber imaging with weak power density under scattering media satisfies to avoid heat damage by laser beam. For these reasons, our strategy opens a new route toward extreme minimally invasive medicine. Our approach also possesses various potentialities, such as biopsy imaging using hyperspectral imaging [35], three-dimensional imaging [7, 14], and even polarization imaging [36]. On contrary, presented single fiber imaging is limited for working as a rigidscope, because speckle patterns were changed by a bending of their optical fiber. However, various control of speckle pattern has been proposed in recent years. In particular, we note that multimode-interference is effective to generate various speckle patterns [37, 38]. Previous studies clarified several mm-lengths of a multimode optical fiber yields unique electric field propagation in the fiber. Spectral sweep laser enables generation of various speckle patterns. We anticipate that our single fiber ghost imaging can be applied to the medical design, which is useful for various applications in neurology and cardiology.

Materials and Methods

Experimental setup
We show an experimental setup for single fiber ghost imaging in Fig. 2 and Fig. S1. A laser diode (LD, LTC56B, Thorlabs) and a thermal controller (TC, TED200C Temperature Controller, Thorlabs) were employed to generate a stable light beam. A rotating diffuser (RD, DG100X100-120, Thorlabs) performed as a speckle generator. An isolator was composed by a Glan-Thompson prism polarizer (P, GTH10M-B, Thorlabs) and a Fresnel rhomb (FR, R600QM, Thorlabs) to block their returned light. The light with speckle pattern generated by the rotating diffuser was collected by an objective lens (OL, M-Plan Apo 20X, Mitutoyo), and was incident into an optical fiber for illumination. We herein employed two branch multimode optical fibers (MMF, BFY105, Thorlabs), due to avoid returned light from the end face of the optical fiber and an optical circulator. A rotation stage (PS60BB-360R, Coms) was used to control the speckle patterns. We captured their speckle patterns using a two-dimensional sensor (2DS, STC-MBS43U3V, Omron-Sentech) The back scattering light was passed through the other optical fiber, and detected the scattering intensity by an avalanche photodiode (APD, APD440A2, Thorlabs). The voltage signal from the photodiode was recorded by an input module (NI-9239, National Instruments). The rotation stage, the CMOS, the avalanche photodiode, and the input module were controlled by (Labview, National Instruments). Captured image was calculated by our source code fabricated by python.

**Sample preparation and laser drawing system**

Firstly, we declare that we didn’t conduct any experiments involving the use of human or animal blood. For this reason, our experiments do not require ethical review for human or animal experimentation.

To perform our single fiber ghost imaging, objects shown in Figs. 3, 4, 5, 6, S2a, and S2b which made of infrared reflected tape (Hansha, Custom), were fabricated by manual cuttings. Thus, dark area on the reconstructed images means air. On contrary, an object shown in Fig. 5 and fig. S2c was fabricated by laser drawing system controlled by two-axis motorized stages, as shown in Fig. S2d. Their dark area on the reconstructed image in Fig. 5 means burnt spots. (see fig. S2 for our fabricated samples.) We also show our laser drawing system in Fig. S2d.

**Decimation of captured images**

We employed a CMOS camera with 720 x 540 pixels. Images captured by the CMOS camera were resized to be 540 x 540 pixels. After resizing, the image resolution was reduced to be 100 x 100 pixels by pixel skipping known as decimation. Before and after the decimation, we compared their images as shown in fig. S2. The decimated image reflected an original image with 540 x 540 pixels.

**Differential Ghost imaging with normalization**

A differential ghost imaging has been proposed to enhance the signal-to-noise ratio of reconstructed images in Ref. [22, 23]. In Fig. 1, the speckle patterns registered with number of $n$ are captured by a CMOS camera as $I_n(x, y)$, and detected a bucket signal as $B_n$ by a an avalanche photodiode (APD). Conventional differential ghost imaging $G(x, y)$ is expressed as

$$G(x, y) = \langle \Delta I_n(x, y) \Delta B_n \rangle$$
where, \( \langle \cdots \rangle \), \( \Delta I_n(x, y) \), and \( \Delta B_n \) are an ensemble average, intensity fluctuations of speckle patterns, and that of bucket signal, respectively. Their fluctuations are rewritten as

\[
\Delta I_n(x, y) = I_n(x, y) - \langle I_n(x, y) \rangle \quad (2),
\]

\[
\Delta B_n = B_n - \langle B_n \rangle \quad (3).
\]

The bucket signal can be rewritten as

\[
B_n = t \int \int [I_n(x, y) \delta S(x, y)] dx dy
\]

\[
= B_{1-n} - \frac{B_{1-n}}{B_{2-n}} \quad (4),
\]

where, \( t \) and \( \delta S(x, y) \) are a transmittance of the optical fiber and scattering fluctuation of an object, respectively. \( B_{1-n} \) and \( B_{2-n} \) are bucket signals by the APD and the CMOS. Differential ghost imaging is characterized by the ratio of their bucket signals as shown in second term. The scattering fluctuation of the sample is obtained as

\[
\delta S(x, y) = S(x, y) - \langle S(x, y) \rangle \quad (5),
\]

\[
\langle S(x, y) \rangle = \frac{1}{t} \cdot \frac{B_{1-n}}{B_{2-n}} \quad (6).
\]

The ensemble average of the scattering property of the object is represented as ratio of their bucket signals related between \( B_{1-n} \) and \( B_{2-n} \). \( B_{2-n} \) is also expressed as

\[
B_{2-n} = \int \int I_n(x, y) \, dx \, dy \quad (7).
\]

Using maximum and minimum intensities on the reconstructed image of \( G(x, y) \), we retrieved an image of the object as follows,

\[
\langle O_s(x, y) \rangle = g \cdot \frac{G(x,y) - G_{\min}}{G_{\max} - G_{\min}} \quad (8).
\]

**Reconstructed image under scattering condition**

Essence of our strategy is to reconstruct an image of an object under scattering condition. As shown in Fig. 6, the object is placed through a diffuser sheet. Observed the object using a conventional CMOS camera at same situation, we obtained pictures in Fig. S6. Figures S8a, S8b, S8c, and S8d show images captured by no diffuser and diffusers with weak, middle, and strong scattering, respectively. In Fig. 6, we designed imaging method by scattering light after passing through their diffusers. In case of no diffuser, the scattering condition \( S(x, y) \) is expressed as
\[ S(x,y) = O_s(x,y) \] (9).

On contrary, we obtain the scattering condition without an object as
\[ S(x,y) = b_s(x,y) \] (10).

Inserted an object as a sample, the scattering light is consisted of backscattering light \( b_s(x,y) \), forward scattering light \( f_s(x,y) \), object scattering light \( O_s(x,y) \), and forward scattering light \( f'_s(x,y) \) after scattering the object. The scattering light gives as
\[
S(x,y) = b_s(x,y) + t_D \cdot f_s(x,y) \cdot O_s(x,y) \cdot t_D \cdot f'_s(x,y) + \cdots
\]
\[
= b_s(x,y) + t_D^2 \cdot [f_s(x,y) \cdot f'_s(x,y)] \cdot O_s(x,y) + \cdots
\] (11),

where, \( t_D \) is a transmittance of the diffuser. For these equations, we retrieve the object’s image \( G \) subtracting an image \( G_B \) without sample from an image \( G_A \) with an object.

**Declarations**

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**Author contributions**

T.W., Y.M., and T.H. conceived the concept of single fiber ghost imaging. T.W. designed the single fiber ghost imaging. Y.H. built up the experimental apparatus. Y.H. and R.K. carried out the imaging experiments. Y.H. performed image processing for the single fiber ghost imaging. Y.H. and R.K. prepared
experimental samples. T.W. developed the theoretical imaging model. Y.M. provides ideas for high contrast imaging. T.H. suggested weak illumination experiments. T.W, Y.M., and T.H. supervised this project. T.W., Y.H., R.K., Y.M. and T.H. participated in writing the manuscript.

**Competing interest**

All other authors declare they have no competing interests.

**Data and materials availability**

All data are available in the main text or the supplementary materials. Additional data related to this paper may be requested from T.W. (wakayama@saitama-med.ac.jp).

**References**


Figures
Figure 1

Concept of our single fiber ghost imaging for extreme minimally invasive medicine. Light from a pseudo-thermal light source is incident into a rotating diffuser. The scattering light is collected by an objective lens and focused on a top of a multimode optical fiber. Light beam, which is spatially coded by speckle pattern, illuminates an object through a diffuser. Here, the light beam is separated the back- and forward-scatterings of $b_s$ and $f_s$, respectively. Forward scattering light is illuminated on the object. Light scattered by the object illuminates the diffuser, again. The scattering light represented by $f$’s is corrected by the multimode optical fiber. The light signal is detected by a single pixel detector. Preregistered speckle patterns and light signal are correlated by a computer.
Figure 2

Speckle pattern property on single fiber ghost imaging. (a) Experimental setup. (b) Speckle patterns were generated through a multi-mode optical fiber at distance of (b1) z = 10 mm and (b2) 20 mm from the optical fiber, respectively. (c) Time-sequential coincidence degree of speckle pattern. (d) Histogram of degree of coincidence calculated by 30 ’ 30 pixels and N= 6,000. Their scale bars in (a) are 1 mm.
Figure 3

Resolution check for single fiber ghost imaging. (a) Reconstructed images at displacements of (a1) $x = 0$ and (a2) $x = 0.05$ mm, respectively. (a) Changes of displacement measured from gaps of two squares. Their scale bars in (a) are 1 mm.
Figure 4

Experimental results of single fiber ghost imaging without scattering media. (a) A picture of a sample. (b) Image by a conventional neuroendoscope. Reconstructed images by speckle patterns of $n = \text{(c)} 1,000, \text{(d)} 10,000$ and \text{(e)} 30,000, respectively. (f) Coincidence degree of cross and square at numbers of recorded speckle patterns. Images of square using (f1) our fiber imaging and (f2) conventional endoscopic imaging.
Figure 5

Single fiber ghost imaging with weak illumination. (a) A picture of a sample. (b) Images of a conventional neuroendoscope at power densities of (b1) 94 and (b2) 1800 mW/cm², respectively. (c)-(e) Images reconstructed by speckle patterns of 6,000 at power densities of (c) 0.1, (d) 1.0, and (e) 2.5 mW/cm², respectively. (f) Comparison of power density properties of coincidence degree and CNR by normalization criteria. (f1) Coincidence degree. (f2) CNR. Note that red circles were normalized by common maximum and minimum in the power density from 0.10 to 0.50 mW/cm². On contrary, blue circles were calculated from maximum and minimum in individual images, respectively.
Figure 6

Demonstration of single fiber ghost imaging under scattering condition. (a) Experimental setup. To compare our imaging with conventional fiber imaging, we conducted experiment using a conventional endoscope. (b) Images captured by a conventional neuroendoscope. Here, a square sample was observed without any scattering condition. (c) Neuroendoscopic image after passing through a diffuser as scattering condition. (c1) Image after computation of c2-c3, (c2) Image captured through the diffuser after inserting an object of the square. (c3) Image captured through the diffuser before inserting the object. (d) Image reconstruction process under scattering field. (d1)-(d5) Images reconstructed by speckle
patterns of $n = (f_1) 1,000, (f_2) 5,000, (f_3) 10,000, (f_4) 20,000$, and $(d_5) 30,000$, respectively. Their scale bars from $(b)$ to $(d)$ are 1 mm, respectively. $(e)$ Microscopic image of the object. $(f)$ Coincidence degree, which is compared with images shown in $(b)$, at numbers of recorded speckle patterns.

**Supplementary Files**

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