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Multispectral upconversion luminescence intensity ratios for ascertaining the tissue imaging depth†

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1 Introduction

Fluorescence imaging has great potential in early stage cancer diagnosis because of its high sensitivity and resolution.1–4 Especially with the development of near infrared (NIR) light excitable lanthanide ion (Ln3+) doped upconversion nanoparticles (UCNPs), more and more attention has been paid on the upconversion scheme.4–19 NIR light excitation has minimal absorption/scattering in animal tissue and will not excite the biological environment, which make UCNPs superior in luminescence imaging over traditional fluorescence compounds like organic dyes and quantum dots (QDs) that need ultraviolet (UV) to visible (Vis) light for excitation. As Chen et al. demonstrated in 2012, the Tm3+ doped UCNPs can image up to 3.2 cm thick in pork tissue, thus UCNPs is an excellent luminescent probe for in vivo imaging of deep tissue.13 Zhang also synthesized several different UCNPs and systematically studied their microscopic luminescence imaging depths by embedding the nanoparticle labeled cells in different animal tissues.18 Moreover, benefiting from the abundantly discrete energy level structures of the doped Ln3+ ions, UCNPs show a unique optical property of multiband upconversion luminescence (UCL) spanning from ultraviolet to near infrared, and the spectrum can be modulated by simply varying the doping ions, e.g. Er, Tm, Ho, etc. and/or relevant concentrations.20–24 Based on this, multicolor imaging methods can be aptly achieved for simultaneously imaging several different lesions with single 980 nm excitation.25,26 We also developed a multifunctional nanoplatform for cancer cell imaging and photodynamic therapy upon the selective energy transfer from multicoloered NaYF4:Yb,Er UCNPs to surface covalently functionalized photosensitizers Rose Bengal (RB).27 All these efforts indicated the prospect of UCNPs in tissue imaging and/or therapy.

Despite this progress, how to relate these images to the exact position of the lesion, i.e. how to accurately locate the tissue depth of luminescence probe labeled cancer, remains a big challenge.28,29 In clinical oncology it has been proved that the invasion depth has a close relationship with cancer metastasis,30–32 and thus the determination of cancer depth is of great significance in cancer staging and prognosis. However, because of the intrinsic complex of the interactions between light and animal tissues (absorption, scattering, reflection, etc.), it is usually difficult to resolve the lesion from traditional single-color planar imaging (only lateral distribution of the luminescent probes is acquired) in which the detected signal intensity

†Electronic supplementary information (ESI) available: Absorption spectra of India Ink, Intralipid and pork muscles; NaYF4:Yb,Er upconversion luminescence spectra detected at different depths in tissue mimicking liquid phantoms and pork muscles. See DOI: 10.1039/c4nr02090a
has a nonlinear dependence on the propagation depth in surrounding tissue, especially when the concentration of luminescent probes is taken into account. In this aspect, fluorescence molecular tomography (FMT) has recently been developed to reconstruct fluorescence images.\textsuperscript{33,34} However, this effort is often interfered by the complex light source arrays and detection techniques, and the requirements of intensive computation and complicated data analysis. Moreover, most present FMT techniques have to combine with CT or MRI to improve the photon reconstruction and image visualization.\textsuperscript{35,36} Thus a simple and independent method of evaluating the lesion depth is very demanding.

In this work, we have established a theoretical model which can be used to have an easy but accurate assessment of the depth of luminescence probes embedded in tissue based on multispectral luminescence of UCNPs. The parameters in the deduced quantitative relation between the light propagation depth and the UCL spectrum were fixed from tissue mimicking liquid phantoms. The optical path-length on excitation and emission could be well separately adjusted and the corresponding UCL spectra were recorded by PMT respectively. The integrated intensity ratio of the green and red emission was used for sensing the depth. The deduced quantitative relation between the light propagation depth is very demanding.

In this work, we have established a theoretical model which can be used to have an easy but accurate assessment of the depth of luminescence probes embedded in tissue based on multispectral luminescence of UCNPs. The parameters in the deduced quantitative relation between the light propagation depth and the UCL spectrum were fixed from tissue mimicking liquid phantoms, and the setup is depicted in Fig. 1, where UCNPs were encapsulated in a capillary tube and embedded in the tissue mimicking liquid phantom. The optical path-length on excitation and emission could be well separately adjusted and the corresponding UCL spectra were recorded by PMT respectively. The integrated intensity ratio of the green and red emission was used for sensing the depth. The deduced quantitative relation between the light propagation depth and the UCL spectrum were recorded by PMT respectively. The integrated intensity ratio of the green and red emission was used for sensing the depth. The deduced quantitative relation between the light propagation depth and the UCL spectrum were recorded by PMT respectively. The integrated intensity ratio of the green and red emission was used for sensing the depth. The deduced quantitative relation between the light propagation depth and the UCL spectrum were recorded by PMT respectively. The integrated intensity ratio of the green and red emission was used for sensing the depth.
2.3 Characterization

Structural characterization was performed with a Philips Morgagni transmission electron microscope (FEI Company, US). UV-Vis absorption spectra of solutions in a quartz cuvette (1 cm) were recorded with a Hewlett-Packard/Agilent 8453 diode-array biochemical analysis UV-Vis spectrophotometer. The steady-state UCL spectra of UCNPs were detected using a SPEX Fluorolog-3 spectrophotometer (HORIBA JobinYvon, France) where a CW semiconductor diode laser of 980 nm was used for excitation.

2.4 Animal tissue depth evaluation using UCNPs

To validate the methodology of using multicolor UCL imaging to determine the tissue depth, layered pork muscle tissue (thickness = 0.65 mm) was utilized as the model. In the experiment, 50 μL of NaYF₄:Yb,Er UCNPs solution (10 mg mL⁻¹) were firstly dropped onto a layer of pork muscle, which can seep into the tissue within a few seconds. Then more layers of fresh pork muscle (label-free) were covered layer by layer onto the one labeled with UCNPs, and the corresponding UCL spectra at different tissue depths were recorded using an SPEX Fluorolog-3 system under 980 nm excitation (700 mW cm⁻²). The luminescence intensities at 540 nm and 650 nm were used for quantitative analysis. The real color UCL imaging was recorded using a Canon Power Shot S120 digital camera by putting an 890 nm short-pass filter (Semrock) in front to eliminate the scattered 980 nm laser light.

3 Results and discussion

3.1 Characterization of NaYF₄:Yb,Er nanoparticles

Fig. 2A is the transmission electron microscopy (TEM) image of the ligand free NaYF₄:Yb,Er nanoparticles and the average diameter is 39 nm. Fig. 2B is the corresponding selected area electron diffraction (SAED) pattern, which confirms the as-synthesized UCNPs are in hexagonal phase which is known to have high upconversion efficiency.³⁵

Fig. 3A is the energy level structures of Yb³⁺ and Er³⁺ co-doped UCNPs and there are two main UPL bands around 540 nm and 650 nm, respectively. Considering that the allowed excitation power density is limited in animal tissues, we began with the excitation power dependence of the UCL spectrum. The upconversion spectra shown in Fig. 3B were taken under relative weak excitation densities from 175 to 700 mW cm⁻², well below the UCL saturation threshold. The UCL in the visible region exhibits the feature of Er³⁺, a green band around 540 nm and a red one around 650 nm, corresponding to transitions of ¹⁴S₅/₂→¹⁴I₅/₂ and ⁴F₇/₂→⁴I₃/₂ in the doped Er³⁺ ions, respectively (Fig. 3A). The spectra demonstrate a monotonic increase with the excitation power without saturation. The excitation power density dependence of the two UCL bands is shown in Fig. 3C. From the slope of linear fitting in the log-log scale, it can be concluded that the upconversion emission has a quadratic dependence on the 980 nm excitation power, showing that the UCL originates from two-photon processes, no higher order process is significantly involved. An ideal luminescence marker should have a minimal or no bleaching effect under long time irradiation, thus we studied specifically the photostability of the two UCNP emission bands under 30 min continuous 980 nm excitation and the results are shown in Fig. 3D. There is no noticeable photodegradation. Based on these studies, we came to the conclusion that UCNPs could serve as ideal contrast agents for long-term luminescence imaging.

3.2 Depth dependent UCL in liquid phantom

To study the path-length effects on UCL spectra a 2-D translation platform was built up as shown in Fig. 1A, in which the excitation and emission processes could be separately controlled by simply adjusting the liquid phantom cuvette along different directions. Fig. 4 are the extinction spectra of the different components of the liquid phantom used in our study. India Ink and Intralipid were served respectively as the main absorption and scattering components. From the spectra we can see that their extinction coefficients at short wavelength (e.g. 540 nm) are higher than those at longer wavelength (e.g. 650 nm). Both the Intralipid and India Ink have linear response of extinction

Fig. 2 (A) TEM image of the NaYF₄:Yb,Er UCNPs. (B) Selected area electron diffraction (SAED) diagram of UCNPs.
coefficients to their concentrations (Fig. S1 and S2 in the ESL†), thus we could control the optical properties by modulating the relative concentrations of the two. Since the hemoglobin in real animal tissue has high absorption around 540 nm, Rose Bengal was also added into the liquid phantom to further enhance the absorption in this spectral region. Fig. 4B shows the extinction spectra of liquid phantoms with and without Rose Bengal. The small peak detected around 540 nm in the red curve can be attributed to the characteristic absorption of Rose Bengal. The UCL spectra recorded in Em-, Ex- and Ref modes are shown in Fig. S3A–C in the ESL† and the corresponding integrated intensities of the green and the red bands are given in Fig. 5A–C (mono-logarithm scale). In Ex mode (Fig. 5A), both the green and red emissions attenuate exponentially with the same slope (≅4.3), indicating that the spectral shape does not vary with the propagation path-length of the excitation light. Here the contribution of surface reflection is already excluded. In Em mode (Fig. 5B), however, the green band attenuates faster than the red one, which is understandable because the liquid phantom absorbs and scatters more at shorter wavelength (Fig. 4). The fitted attenuation slopes are −3.25 and −2.72 for the green and red bands, respectively. The slope difference between Ex- and Em modes is related to the two photon nature of the UCL process. Fig. 5C shows the fitted slopes of Ref mode; both emission bands attenuate significantly with depth; the fitted attenuation slopes are −7.57 and −7.01, respectively. The attenuation slopes in Ref mode are found to be exactly the sum of the slopes in Ex- and Em modes. In Fig. 5D, we show the penetration depth dependent intensity ratio of green/red UCL (G/R ratio). Exponential relation is found in Em- and Ref modes, whereas it remains almost constant in Ex mode. This indicates that the propagation path-length of excitation light has a negligible effect on the G/R ratio.

3.3 Theoretical model

In our experiments the 980 nm laser was collimated into a planar beam of 10 mm² to excite the UCNPs capillary tube embedded in the cuvette that is filled with liquid phantoms. The UCL was thus treated as a line light source, and the energy fluence attenuated isotropically in the tissue. Based on the optical diffusion theory,39,40 the distribution of the excitation light and the emission light along their propagation direction (z) inside tissue could be written as:

\[ \frac{d^2 \phi_x(z)}{dz^2} - \mu_{a,x} \phi_x(z) = -\mu'_{a,x} P_{0\phi} e^{-\mu'_{a,x} z} \]  

(1)

\[ \frac{d^2 \phi_m(z)}{dz^2} + \frac{2}{z} \frac{d \phi_m(z)}{dz} - \mu_{\text{eff},m}^2 \phi_m(z) = \frac{P_{0\phi}}{D_m} \delta(z) \]  

(2)

here \( \phi_x \) and \( \phi_m \) are the emission intensity of the excitation light and emission light inside the tissue, \( P_{0\phi} \) and \( P_{0\phi} \) are the initial intensities of the incident excitation light (e.g. 980 nm) and the emission light (e.g. 540 or 650 nm), \( \mu_{a,x} \), \( \mu_{s,x} \) and \( \mu'_{a,x} \) are the absorption coefficient, reduced scattering coefficient and the total attenuation coefficient for the excitation light, \( \mu_{\text{eff},m} \) is the effective attenuation coefficient for the emission light, \( D_x \) and \( D_m \) are the diffusion coefficients of excitation and emission, respectively. The solution for the emission diffusion equations is:

\[ \phi_m(z) = \frac{P_{0\phi} e^{-\mu_{\text{eff},m} z}}{4\pi D_m z} \]  

(3)

From eqn (3) we can see that the fluorescence energy fluence \( \phi_m \) is affected not only by the initial luminescence intensity \( P_{0\phi} \) but also by the tissue optical property \( \mu_{\text{eff},m} \). Regarding NaYF₄:Yb,Er UCNL, we can divide the fluorescence energy fluence into two parts \( \phi_{540} \) and \( \phi_{650} \) corresponding to the two emission bands around 540 and 650 nm, respectively. The intensity ratio \( R \) detected is therefore:

\[ R = \frac{\phi_{540}(z)}{\phi_{650}(z)} = \frac{P_{0,540}(z) e^{-\mu_{\text{eff},540} z}}{4\pi D_{540} z} \frac{P_{0,650}(z) e^{-\mu_{\text{eff},650} z}}{4\pi D_{650} z} = \frac{P_{0,540}(z)}{P_{0,650}(z)} \frac{D_{650}}{D_{540}} \left( \mu_{\text{eff},540} + \mu_{\text{eff},650} \right) \]  

(4)
The first item at the right side of eqn (4) is constant that is determined by the intrinsic optical properties of UCNPs, as proved in Fig. 5A. And the diffusion coefficients $D_\text{eff}$ are constant in the second part are also constant for a homogeneous tissue. Thus from this equation we can deduce that the G/R ratio detected at the surface follows an exponential decay pattern with increasing the tissue depth, and the attenuation slope can be calculated from the difference of effective attenuation coefficients at these two wavelengths.

3.4 Ascertaining the tissue imaging depth with multispectral upconversion luminescence

`Real tissue` contains hemoglobin and other chromophores, which lead to more absorption around 540 nm compared to 650 nm. To mimic this, studies were performed in liquid phantoms with different optical properties by varying the concentration of India Ink, Rose Bengal and Intralipid. The corresponding attenuation slopes detected in Ref mode are given in Fig. 6 (the corresponding spectra data are given in Fig. S4–S6 in the ESI†).

In sample A, the attenuation slopes are $-5.46$ and $-4.71$ for green and red bands, respectively (Fig. 6A). Adding more RB into the phantom, the slope of the green band changes to $-5.95$ while the red band remains almost constant ($-4.74$, Fig. 6B). This is because RB has maximal absorption around 540 nm, which makes the green band attenuate faster. In Fig. 6C, more Intralipid is in sample B, the scattering increases while the absorption remains the same around 540 nm and 650 nm. Sharper decreases of the intensities are observed with the slopes of $-6.12$ and $-5.23$, which is predictable since scattering is enhanced in both excitation and emission. Fig. 6D shows the G/R ratio of samples A, B and C, where the fitted slopes are $-1.72$, $-2.76$ and $-2.80$, respectively. Deviating from sample A, the slope variations are approximately the same for samples B and C although they have different amounts of Intralipid (the amount of India Ink/Rose Bengal was the same). This result tells us that the G/R intensity ratio is more sensitive to the absorption coefficient than the scattering coefficient. In fact it is in line with eqn (4) and (5), where the effective attenuation coefficient has a linear relationship with $\mu_a$ but a quadratic one with $\mu_s$.

So far we have built up the quantitative relationship between the propagation depth of UCNPs in tissue mimic liquid phantoms and the UCL spectra. In the following, we will validate the method employing layered pork muscle tissue. Fig. S7† shows the extinction spectra of pork muscle with different thicknesses (or layers). As pork muscles contain a high concentration of myohemoglobin which has relatively high absorption around 540 nm, the effective attenuation coefficient is thus higher than that of 650 nm. The photographs in Fig. 7A and B are the real color UCL images recorded in Ex- and Em modes, respectively.

The incident excitation power density at 980 nm was 700 mW cm$^{-2}$ at the surface. In Ex mode, although the emission intensity dropped proportionally with the tissue depth (the actual Fig. 6 (A), (B) and (C) are the depth dependent UCL intensities detected in three liquid phantom samples with different components. (D) is the corresponding G/R intensity ratio attenuation curves of the three samples. Error bars are marked in the figures.
excitation power decreased), the color remained unchanged. In contrast, the color of UCL in Em mode changed from green to red with the tissue depth, reflecting the higher absorption of muscle hemoglobin to 540 nm emission. More quantitative analyses were carried out by recording the UCL spectra at different depths of Em-, Ex- and Ref modes (data are shown in Fig. S8–S10 in the ESI†), and the UCL intensities around 540 nm and 650 nm are given in Fig. 7C and D. Fig. 7C is the depth dependent UCL intensity recorded in Ex mode, where a similar tissue penetration depth dependence is observed for the green and red emission. Fig. 7D shows the results of Ref mode; the slopes for green and red bands are −7.33 and −5.48, respectively. To determine the reproducibility of the results, all the spectra are recorded at least three times for each tissue depth. The results as shown in Fig. 7D are reproducible with the error bar less than 10%. Compared with the results on liquid phantoms, the G/R attenuation slope in pork muscle is much higher (4.74), as shown in Fig. 7E. This discrepancy might attribute to the higher effective coefficient difference of the two bands in the pork muscles than that in the liquid phantoms. As the G/R ratio detected is determined by the inherent properties of NaYF4:Er3+,Yb3+ UCNPs, which is independent of the absolute amount of UCNP labeled lesion.

4 Conclusions

In conclusion, a theoretical model has been established to relate the relative intensities of the UCL spectra to the tissue imaging depth of UCNPs. The method was validated in liquid phantoms and pork muscle tissue. Although in this work we have focused on NaYF4:Er3+,Yb3+ UCNPs, other upconversion materials can be similarly employed as well for even better penetration, e.g. introducing Tm3+. This new approach shall lift significantly the power of nanotechnology assisted luminescence imaging by providing also accurate information of the depth of UCNP labeled lesion.

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