Can photoacoustic imaging quantify surface-localized J-aggregating nanoparticles?

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Abstract. We investigate the feasibility of photoacoustic (PA) imaging to quantify the concentration of surface-localized nanoparticles, using tissue-mimicking phantoms and imaging with a commercial PA instrument at 815 nm and a linear-array transducer at a center frequency of 40 MHz. The nanoparticles were J-aggregating porphysomes (JNP) comprising self-assembling, all-organic porphyrin-lipid micelles with a molar absorption coefficient of $8.7 \times 10^5 \text{ cm}^{-1} \text{ M}^{-1}$ at this wavelength. The PA signal intensity versus JNP areal concentration followed a sigmoidal curve with a reproducible linear range of $\sim 17 \text{ fmol/mm}^2$ to $11 \text{ pmol/mm}^2$, i.e., $\sim 3$ orders of magnitude with $\pm 34\%$ error. For physiologically-relevant conditions (i.e., optical scattering-dominated tissues: transport albedo $>0.8$) and JNP concentrations above $\sim 330 \text{ fmol/mm}^2$, the PA signal depends only on the nanoparticle concentration. Otherwise, independent measurement of the optical absorption and scattering properties of the underlying tissue is required for accurate quantification. The implications for surface PA imaging, such as in the use of targeted nanoparticles topicaly to tissue as in endoscopic diagnosis, are considered.

Keywords: photoacoustic spectroscopy; nanophotonics; biomedical optics; optical properties.

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1 Introduction

Photoacoustic (PA) imaging is a rapidly developing imaging technique, combining the molecular specificity of light with the deeper imaging capability of ultrasound. Depending on the platform, photoacoustic imaging (PAI) is capable of imaging optical absorbers at depths of multiple optical pathlengths (PA tomography) or with high resolution (PA microscopy). This positions PAI in a unique position for medical imaging applications, wherein the ability to image optical absorbers (e.g., blood, optically active nanoparticles) provides complementary information to structural imaging. Many groups have demonstrated PAI’s potential for in vivo functional imaging, including mapping hemoglobin concentration and oxygen saturation, photosensitizing dyes, iron levels, and metal nanoparticles.

Our recent development of porphysomes has added another dimension to PAI, enabling disease diagnosis, treatment monitoring, and real-time monitoring of nanoparticle distribution in vivo. Porphysomes are organic nanovesicles (~100 nm) made from self-assembling porphyrin-lipid bilayers. They are photoacoustically active due to their extremely high and wavelength-tunable optical extinction ($\sim 3 \times 10^9 \text{ cm}^{-1} \text{ M}^{-1}$ for pyropheophorbide porphysome at 680 nm) that is comparable to that of gold nanoparticles. This same property makes them suitable to enhance and spatially localize energy deposition in photothermal therapy. Their porphyrin fluorescence and photodynamic activity are self-quenched in the intact state due to the high porphyrin concentration ($\sim 80,000$ porphyrin molecules per nanoparticle), but become active upon cell uptake and disassembly into monomers. Porphysomes can be biomarker targeted, for example, to tumors by antibody labeling. PAI would then, in principle, be able to quantify the concentration of porphysomes in the target tumor in vivo, giving a measure of the biomarker expression level, which is becoming increasingly important for diagnosis, staging, and therapeutic guidance, as, for example, in breast cancer to assess hormone receptors and epidermal growth factor receptor (EGFR). One example of the potential use of PAI is detection and localization of dysplasia in patients with Barrett’s esophagus (BE). As the most important risk factor (>30-fold) for esophageal adenocarcinoma, BE is a condition in which the esophageal squamous mucosa (whitish in appearance on standard endoscopy) transforms to intestinal-type mucosa (pink-red appearance) due to chronic gastroesophageal reflux. BE is readily distinguishable from normal esophagus by endoscopy, but detection of dysplasia from nondysplastic BE is challenging under current standard endoscopic guidance. The current clinical standard is a four-quadrant random biopsy every 1 to 2 cm along the length of the Barrett’s segment, leaving up to 99% of the tissue unsampled. Hence, there is a need for rapid and wide-field imaging for dysplasia diagnosis in BE. Many different optical techniques have been investigated to address this limitation, including diffuse reflectance spectroscopy, tissue autofluorescence, contrast-based fluorescence, Raman spectroscopy, optical coherence tomography, and narrowband imaging, but the challenge remains in detecting dysplasia in BE with high accuracy and to determine if there is submucosal spread of disease, which is critical in therapeutic decision-making. The potential advantages of PAI are its capability for rapid scanning of large tissue areas for detection, and imaging of the full esophageal thickness and beyond for staging. PAI devices suitable for esophageal endoscopy are under development. We are currently evaluating the feasibility and efficacy of intrinsic PAI (i.e., without any exogenous contrast agent), using ex vivo endoscopic mucosal
resection tissues from BE-surveillance patients. While this has shown that PAI is indeed able to produce high-quality images of the tissue microvasculature, the initial findings suggest that specific contrast will be required to achieve high enough sensitivity and specificity. Hence, we hypothesize that combining PAI with targeted porphysomes will improve dysplasia detection in BE. Either systemic or topical administration of the porphysomes can be envisaged, the latter having the advantages of low total dose (minimizing cost and potential toxicity) and low background.

One of the major challenges in PAI is accurate quantification of the PA signal produced by physiologically relevant chromophore concentrations. As is well known, the PA signal can be written as

\[ p_0 = \Gamma \mu_a \Phi, \]

where \( p_0 \) is the initial acoustic pressure distribution, \( \Gamma \) is the Gr"uneisen parameter (a tissue-dependent measure of the PA conversion efficiency), \( \mu_a \) is the chromophore’s optical absorption (optical extinction coefficient multiplied by concentration), and \( \Phi \) is the local light fluence. Solving for chromophore concentration from the PA signal is a difficult inverse problem due to these spatially varying parameters: \( \Phi, \Gamma, \) optical, and acoustic propagation. In addition, the PA signal is not uniform even from a homogenously absorbing tissue, but rather it is dominated by signals from tissue boundaries. Nevertheless, it is possible in principle to create a direct relationship between known chromophore concentration and PA intensity by strict control of the experimental conditions.

The purpose of the present study was to determine the feasibility and accuracy of photoacoustic tomography (PAT) in quantifying the areal concentration of topical porphysomes from the PA signal intensity, using tissue-mimicking optoacoustic phantoms as well-controlled models with varying optical properties at the PA wavelength. In PAT, the standard approach is to flood the field of view with pulsed laser light delivered to the tissue through fiber optic bundles integrated with the ultrasound transducer array. One can expect that the light fluence within tissue will not be homogenous and will attenuate with depth, depending on the tissue optical properties. In particular, the local light fluence experienced by surface-localized nanoparticles is the sum of the incident (primary) and backscattered (diffusely reflected) light. It is well established that the diffuse reflectance is a function of the transport albedo (\( a' \)) defined as \( a' = \mu_t/[\mu_a + \mu'_s] \), where \( \mu_a \) is the absorption coefficient and \( \mu'_s \) is the reduced scattering coefficient. Hence, measuring \( a' \) should enable correct normalization of the PA signal for the total light fluence experienced by the surface nanoparticles to calculate their areal concentration.

For the first part of this study, we measured the PA signal from varying concentrations of surface-localized porphysomes on tissue-mimicking phantoms comprising gelatin with known and varied absorption and scattering. The PA signal versus porphysomes concentration was plotted and the corresponding errors were calculated. We considered also how the phantom optical properties affect the PA signal.

\[ 2 \quad \text{Materials and Methods} \]

\[ 2.1 \quad \text{Photoacoustic System and Data Acquisition Procedure} \]

We used a commercial PAT imaging instrument (Vevo LAZR, Visualsonics, Toronto, Ontario, Canada), described in detail elsewhere. This instrument has an integrated fiber-optic and linear-array transducer (40 MHz center frequency, 256 elements) with a crossed-beam geometry for optical excitation. Ultrasound and PA images were collected at a frame rate of 5 Hz across a field of view of \( \sim 15 \) mm (depth) by 14 mm (width). The transducer was mounted on a motorized translational stage to scan over a three-dimensional (3-D) volume, and PA images were acquired in 0.1-mm increments at 815 nm. Due to the crossed laser beam geometry of the PAI transducer probe, there is an optimal PA zone within the field of view at \( \sim 9 \) to 11 mm from the probe surface. For consistency in the transducer orientation and incident light fluence (\( \Phi_0 \)), we positioned the surface of the phantom 9 mm from the transducer for all measurements.

\[ 2.2 \quad \text{Sample Preparation} \]

J-aggregate porphysomes (JNP) comprising bacteriophage-phospholipid A-lipid dye was used, as described elsewhere. These are of particular interest for PA monitoring and enhancement of photothermal therapy, where quantification is important as part of dose optimization. They have typical porphysome optical absorption spectra, as shown in the insert of Fig. 1. To simulate topically applied nanoparticles on tissue, we used JNP on the surface of tissue-mimicking phantoms made from gelatin and Intralipid, as shown in Fig. 1. The procedure was to pipette 50 \( \mu l \) of varying concentrations (0.355 nM to 118 \( \mu M \)) of JNPs in phosphate-buffered serum (PBS) onto the surface [Fig. 1(a)] and leave it overnight at room temperature to evaporate [Fig. 1(b)] while protected from light by aluminum foil to prevent photobleaching. A thin \( (\sim 0.5 \) mm) layer of clear gelatin was added on top [Fig. 1(c)] to avoid physical damage by the PAI probe and ultrasound coupling gel during imaging.

In the main experiments, the goal was to measure the PA signal as a function of both the JNP concentration and optical properties of the tissue-mimicking phantom base. Based on published data for normal esophageal tissue (\( \mu_a = 0.080 \pm 0.023 \) mm\(^{-1} \) and \( \mu'_s = 0.77 \pm 0.15 \) mm\(^{-1} \)) at 630 nm, we selected the range \( a' \) = 0.65 to 0.95 to represent the expected maximum range of esophageal tissue, as relevant to ongoing PAI studies in BE patients, while recognizing that the variability in Barrett’s tissue is likely much larger than that in normal squamous esophagus because of the high inflammatory component. The phantom bases were created using 5% weight/volume gelatin (GX50, Matheson Coleman & Bell, Norwood, Ohio) in reverse-osmosis water, with intralipid (Fresenius Kabi, Richmond Hill, Ontario, Canada) added at different concentrations to achieve the desired \( \mu'_s \). The optical properties of intralipid were experimentally verified using the added-absorber technique. Our experimental results show up to 40% reduction in optical scattering for the same intralipid concentration in water compared with the value in gelatin, consistent with the findings of Cook et al. However, the impact on \( a' \) is less marked. We created tissue-mimicking phantom bases with \( a' \) spanning the targeted range, with optical properties varied as per Table 1. The intrinsic absorption of Intralipid is negligible (<0.001 mm\(^{-1} \)) compared with that of 5% gelatin (0.051 mm\(^{-1} \), measured by an UV–vis spectrophotometer).

The dilution of JNP spanned five orders of magnitude (0.0011% to 33%) by diluting from the stock 355-\( \mu M \) concentration. The JNP concentration after deposition is defined in detail elsewhere.
PAI and ultrasound images were analyzed in MATLAB (Mathworks, Natick, Massachusetts), as shown in Fig. 2. A logarithmic scale [20 log$_{10}$(dB)] was applied to the onboard beam-formed PA signals. From the original 3-D volume [Fig. 2(a)], the PA signal was flattened to two-dimensions [Fig. 2(b)] by isolating the maximum intensity of each column perpendicular to the tissue surface. A PA intensity threshold was applied to this, and the JNP droplet was identified by visual inspection. We then applied an area threshold to each column of spots representing three replicates of JNP droplets, to exclude the surrounding discrete spots that, we believe, were due to tiny air bubbles trapped between the phantom layers during fabrication. The air bubbles are less visible and phantom albedo, confirming that there is a significant contribution to the PA signal from light that is diffusely backscattered to the surface. The increase and left-shifting of the floor of the sigmoid curves with increasing $a'$ are consistent with this interpretation and allows detection of lower JNP concentrations. For physiologically relevant conditions (i.e., optical scattering-dominated tissues: transport albedo $>0.8$) and JNP concentrations above $330 \text{ fmol/mm}^2$, the PA signal depends only on the nanoparticle concentration (note the sigmoidal curves for medium and high $a'$ values are practically indistinguishable above $330 \text{ fmol/mm}^2$). In the (biologically unlikely) case of very high JNP areal concentration, the PA signal becomes essentially independent of concentration, i.e., the signal saturates.

We defined the concentration estimate error as the difference between the sigmoid fit and its 95% confidence intervals, as shown in Fig. 4. Using medium $a'$ as an example, we first calculated the confidence intervals [Fig. 4(a)], shown as the dashed lines to the left and right of the sigmoid. These are smallest in the linear portion of the signal response curves and increase as the curves flatten out at both ends. We then calculated the error for each JNP areal concentration. Using medium $a'$ as an example [Fig. 4(b)], a PA value of 39 dB corresponded to $\sim 330 \text{ fmol/mm}^2$, the PA signal depends only on the nanoparticle concentration (note the sigmoidal curves for medium and high $a'$ values are practically indistinguishable above $330 \text{ fmol/mm}^2$). In the (biologically unlikely) case of very high JNP areal concentration, the PA signal becomes essentially independent of concentration, i.e., the signal saturates.

As expected qualitatively, the PA intensity increased monotonically and consistently with increasing JNP concentration and phantom albedo, confirming that there is a significant contribution to the PA signal from light that is diffusely backscattered to the surface. The increase and left-shifting of the floor of the sigmoid curves with increasing $a'$ are consistent with this interpretation and allows detection of lower JNP concentrations. For physiologically relevant conditions (i.e., optical scattering-dominated tissues: transport albedo $>0.8$) and JNP concentrations above $330 \text{ fmol/mm}^2$, the PA signal depends only on the nanoparticle concentration (note the sigmoidal curves for medium and high $a'$ values are practically indistinguishable above $330 \text{ fmol/mm}^2$). In the (biologically unlikely) case of very high JNP areal concentration, the PA signal becomes essentially independent of concentration, i.e., the signal saturates.

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general, at a maximum error of $\pm 34\%$ for zero, low, and high $a'$ values, we can quantify the JNP areal concentration within 1 to 2 orders of magnitude (Appendix). This range is extended to almost 3 orders of magnitude (0.017 to 11 pmol/mm$^2$) for medium $a'$.

To further understand and confirm the increase and shift of sigmoid curves, we observed in Fig. 3, we created multiple tissue-mimicking phantom bases with different optical properties and measured the PA signal generated by each without added porphysomes. We found that the PA signal correlated best to the product of the $\mu_a$ and $a'$, as shown in Fig. 5.

4 Discussion and Conclusions

Our goal was to establish the relationship between the PA signal from the surface nanoparticles and their areal concentration using PAT. Given a measured PA value, we can use Figs. 3 and 4(c) to estimate the JNP areal concentration ± error, respectively. For instance, for a JNP PA signal of 65 dB on medium-$a'$ tissue, the predicted areal concentration is 0.407 pmol/mm$^2$ ± 4.6%. The quantifiable range depends on the required accuracy of the measurement. For example, for a maximum error of $\pm 34\%$ (1 standard deviation of a normal distribution) on medium-$a'$ tissue, the lowest and highest quantifiable JNP concentrations are 0.017 and 11 pmol/mm$^2$, respectively. To estimate the JNP concentration more accurately, knowledge of the underlying tissue’s optical properties may be required, in
particular when the PA signal from the nanoparticles is low (< ~ 60 dB) or when the transport albedo of the tissue is low. These optical properties could be measured in vivo by, for example, diffuse reflectance measurements.18,36

The increase in the PA signal with increasing JNP concentration is, of course, as would be expected qualitatively from Eq. (1), holding $\Phi$ and $\Gamma$ as constant while increasing the optical absorption. The S-shaped response curves of PA versus JNP concentrations show a central linear region with the detector noise floor and ceiling, typical of sensor response curves.37 A sigmoid was used as the simplest equation to fit data of this form, as for example in drug dose–response curves.38 The sigmoid-form relationship between PA versus JNP areal concentration should be universal for surface-localized chromophores. We also highlight the general shape of the quantitation error, as shown by the plot in Fig. 4(c). Quantitation error is lower at the center of the sigmoids (~10−1 to 100 pmol/mm²), and error increases as JNP concentration approaches the floor or ceiling of the sigmoids. This is expected as the floor and ceiling of the sigmoids are at the edge of the PA system’s dynamic range. PA signal contrast between different concentrations is reduced, resulting in higher quantification error.

As previously mentioned, in this experimental geometry, where PAT uses a pulsed laser to flood the field of view from the top, the fluence attenuates with depth depending on the tissue optical scattering and absorption coefficients. Higher scattering also results in higher backscattered (diffusely reflected) light reaching the tissue surface. The local light fluence ($\Phi$) on the tissue surface is

$$\Phi = \Phi_o + \Phi_R,$$

where $\Phi_o$ is from the incident (primary) light and $\Phi_R$ is the back-scattered light (which depends on $a^032$). The increase and leftshifting of the floor of the curves were found to correlate best with the product of the absorption and $a'$ of the underlying phantom, as seen in Fig. 5.
These results pose a question regarding the clinical translation feasibility of nanoparticle-contrast PA endoscopy, namely can a high enough surface density of biomarker-targeted topical porphysomes be achieved under typical in vivo conditions to give a reliable PAI signal intensity from which the porphysome surface concentration can be calculated? Estimating the binding affinity of targeted porphysomes is beyond the scope of this paper, but if the reasonable detectable JNP concentration we reported is orders of magnitude higher than the ideal case, where all tissue surface is bound to porphysome, then PAI is not likely to be a good platform for quantitative porphysome imaging. Assuming that the accessible tissue surface of interest (e.g., dysplasia or carcinoma in situ) is completely covered with porphysomes bound to cell-surface receptors and tightly packed, then the areal concentration would be $\sim 10^8$ JNP/mm². At $\pm 34\%$ error, the lowest quantifiable concentration is $\approx 7.81$ to $31.8 \times 10^5$ JNP/mm² (Appendix), at least 2 orders of magnitude lower than the ideal case of $10^8$ JNP/mm². Hence, PAI detection and quantification of surface-bound targeted porphysomes on tissue seems feasible.

These results pose another question: how well do our phantoms match general antibody-based tissue surface labeling? The distribution of JNP within our phantom spots is relatively uniform, as can be seen from the line profiles of a typical spot in Fig. 2(d). Using EGFR as an example, in vitro studies have shown uniform distribution of cell-surface receptors and anti-EGFR antibody conjugated gold nanoparticles bound on cancer cell-surface receptors. But it is unlikely for cell-surface receptors/nanoparticles to remain uniformly distributed throughout cancerous tissue, where there may be multiple cell types at different disease stages. Nevertheless, we should still be able to see JNP PA hotspots above the minimal quantifiable threshold, quantify the JNP areal concentration, make relative comparison of biomarker expression level between disease stages and potentially estimate absolute biomarker expression level. These are subjects of ongoing studies and will be experimentally investigated in the near future, initially in ex vivo human endoscopic mucosal resection specimens with topical applied antibody-conjugated porphysomes. PA quantification of other porphysomes applications (e.g., subsurface/porphysomes within tumor, porphysomes within blood vessels through systemic delivery) is also subject of ongoing studies.

We are mindful of the limitations of these results. Our goal here was to investigate PA nanoparticle quantification for the particular situation of topical applied nanoparticles using PAT, so the conclusions are restricted to this scenario. The specific values of, e.g., the minimum detectable nanoparticle concentration, will also be particular to the experimental parameters used, such as the laser power and transducer sensitivity and the nanoparticle (i.e., JNP). In addition, an underlying assumption is that $\Gamma$ does not change with JNP concentration, which is reasonable as PA increases linearly with JNP concentration. It is unlikely for $\Gamma$ to change within the linear range or at low JNP concentration, but it is possible at high JNP concentration $\Gamma$ may not remain constant, and contribute to the rate of PA signal flattening. Nevertheless, the results do indicate that topical nanoparticle quantification should be feasible with a sensitivity and accuracy that would be relevant to clinical applications, such as porphysome contrast-enhanced endoscopy.

In conclusion, PAT can quantify surface-localized JNPs. Our results enable quantification of the porphysome areal density, and the quantifiable range is determined by the required accuracy. At $\pm 34\%$ error on medium-$\alpha'$ tissue, the quantifiable areal concentration is in the range of $17 \text{ fmol/mm}^2$ to $11 \text{ pmol/mm}^2$, which should be sufficient for clinical applications, such as biomarker-based diagnosis, or guiding subsequent ablative therapies, such as porphysome-enhanced photothermal therapy. In correcting the measured signal for the effects of backscattered light from the tissue, we need to know the tissue optical properties, which can be measured from the diffuse reflectance. For endoscopic applications, one can then envisage ways to integrate both PAI and diffuse reflectance measurements in the same probe.

### Appendix: Estimate of Number of J-Aggregating Porphysomes Per Square Millimeter Surface Area

For porphysomes, the molar concentration refers to the concentration of the component porphyrin-lipid conjugate, which is bacteriochlorophyll (bchl) molecule in the case of JNP. There are $\sim 83,000$ porphyrin-lipid conjugates per 100-nm diameter JNP containing $95\%$ porphyrin-lipid and with 15 mol% of bchl in the JNP’s formulation. This gives:

Number of bchl molecules per JNP nanoparticle = 15/95 × 83000 = 13105

Thus

\[
0.195 \text{ pmol/mm}^2 = 0.195 \times 10^{-12} \text{ mol/mm}^2 = 0.195 \times 10^{-2} \times 6.022 \times 10^{23} \text{ bchl/mm}^2 = 1.17 \times 10^{11}/13105 \text{ JNP/mm}^2 = 8.96 \times 10^6 \text{ JNP/mm}^2.
\]

For each $\alpha'$, the lowest and highest quantifiable JNP concentrations at $\pm 34\%$ error are shown in Tables 2 and 3, respectively.

### Disclosures

The authors have no relevant financial interests in the paper and no other potential conflicts of interest to disclose.
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