Abstract

Photoacoustic tomography (PAT) is a molecular imaging technology. Unlike conventional reporter gene imaging, which is usually based on fluorescence, photoacoustic reporter gene imaging relies only on optical absorption. This work demonstrates several key merits of PAT using lacZ, one of the most widely used reporter genes in biology. We show that the expression of lacZ can be imaged by PAT as deep as 5.0 cm in biological tissue, with resolutions of ~1.0 mm and ~0.4 mm in the lateral and axial directions, respectively. We further demonstrate non-invasive, simultaneous imaging of a lacZ-expressing tumor and its surrounding microvasculature in vivo by dual-wavelength acoustic-resolution photoacoustic microscopy (AR-PAM), with a lateral resolution of 45 µm and an axial resolution of 15 µm. Finally, using optical-resolution photoacoustic microscopy (OR-PAM), we show intra-cellular localization of lacZ expression, with a lateral resolution of a fraction of a micron. These results suggest that PAT is a complementary tool to conventional optical fluorescence imaging of reporter genes for linking biological studies from the microscopic to the macroscopic scales.


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INTRODUCTION

A key finding of the Human Genome Project, completed in 2003, was that in our enormous genome of 20,000–25,000 genes, only 2% of the total encodes all of the proteins necessary for building the human body and executing its physiological functions [1]. We are now being challenged to identify the functions of the discovered genes, understand the molecular mechanisms of physiology and pathology, and develop personalized treatment of diseases [2]. These investigations require evaluating the expression patterns of the genes of interest. Traditional methods that measure mRNA in vitro have several limitations. First, in vitro observation may not correspond to what occurs in vivo due to the difficulty of replicating the native microenvironment. Second, they produce only a single data point from each cell culture or sacrificed animal. To get statistically robust data, procedures must be repeated many times, which is labor-intensive, time-consuming, and costly. Third, they are incapable of localizing where genes are expressed in cells. Fourth, due to the various translational and post-translational modifications of the gene products, the level of mRNA for a gene may not quantitatively correlate with the protein production of the same gene [3].

With the promise of overcoming these limitations, molecular imaging is rapidly being developed for interrogating gene expression in vivo [4], [5], [6], which allows visualization of the spatiotemporal distribution of gene expression in the native environment. The non-invasive nature of molecular imaging allows for the animals studied to serve as their own controls in a longitudinal study, minimizing the uncertainty caused by inter-sample variability. Thus, molecular imaging greatly reduces animal use, labor, and cost. Moreover, molecular imaging could deeply impact clinical practices by contributing to the early detection and fast staging of diseases, accurate evaluation of treatment outcomes, and rational design of novel therapies.

Two optical imaging tools, bioluminescence imaging (BLI) and fluorescence imaging (FLI), are currently the main techniques used in preclinical research to study gene expression. BLI can detect fM of protein while FLI has a typical sensitivity on the order of nM [5]. Both methods have high throughput rates, however they have limited imaging depth (~1 cm) [5], lack of depth resolution, and poor lateral resolution. Although µm-order resolution is achievable optically within a thin slide of tissue, deeper structures are poorly resolved in vivo due to optical diffusion.

None of the current mainstream molecular imaging modalities can visualize gene expression at both microscopic and macroscopic levels. Previous works have suggested a new approach to imaging gene expression in vivo using photoacoustic tomography (PAT) based on optical absorption [7], [8], [9]. In our previous work, a single-scale single-/dual-wavelength photoacoustic system was used to non-invasively and simultaneously image the morphology of a lacZ-marked 9 L gliosarcoma and its surrounding microvasculature in vivo [7], [8]. Different implementations of PAT allow the spatial resolution to be scaled with the desired imaging depth while maintaining a high depth-to-resolution
ratio, which emphasizes the unique multi-scale, high-resolution imaging capability of PAT in this paper [10], [11], [12]. In this work, we demonstrate that PAT can create multi-scale images of gene expression in living biological structures. We show dual-wavelength photoacoustic images with improved image quality and three-dimensional (3D) depiction of the tumor and surrounding vessels. We explore how deep in tissue PAT can detect lacZ expression, and also show intra-cellular localization of lacZ expression with a sub-micron lateral resolution. The results indicate that PAT can bridge the gap between the microscopic and macroscopic domains in molecular biology studies.

### Materials and Methods

#### Chromogenic lacZ Reporter Gene System

Although developed earlier, chromogenic reporter genes have been much neglected for molecular imaging, largely due to the lack of tools to visualize their expression inside living organisms. PAT remedies this and, in our opinion, will rejuvenate the development of new in vivo chromogenic reporter gene systems to solve current challenges in biological imaging.

In this study, we selected the lacZ gene, one of the most widely used chromogenic reporter genes, for multiscale PAT. Originating from E. Coli, lacZ encodes β-galactosidase, a bacterial enzyme responsible for metabolizing lactose into glucose and galactose. The expression of lacZ can be studied by several chromogenic assays. Among them, we chose to use 5-bromo-4-chloro-3-indolyl-β-D-galactosi de(X-gal) as our in vivo molecular probe. X-gal is a colorless analogue of lactose, in which the glucose fragment is replaced by an indole-derivative. β-galactosidase cleaves the glycosidic linkage in X-gal and yields galactose and an optical transparent indoxyl monomer, 5-bromo-4-chloro-3-hydroxyindole. Afterwards, two of the monomers are oxidized to form a stable insoluble blue product, 5,5’-dibromo-4,4’-dichloro-indigo [Figure 1(a)] [13]. Figure 1(b) shows the dramatic chromogenic color change 24 hours after addition of 20 mg/ml X-gal solution to the native cell lysate of transgenic 9 L gliosarcoma cells expressing the lacZ reporter gene. There are other colorimetric assays for β-galactosidase activity, such as o-nitrophenyl-β-D-galactoside (ONPG). However, the blue product from X-gal absorbs light strongly between 605 and 665 nm [8] which is located in the so-called “optical diagnostic window”, and allows us to detect the expression of lacZ at great depths in tissue.

![Figure 1. Detecting the expression of the lacZ reporter gene using the chromogenic X-gal probe.](http://www.plosone.org/article/info%3Adoi%2F10.1371%2Fjournal.pone.0043999.g001)
For *in vivo* imaging, the lacZ/X-gal reporter gene system has two noticeable advantages. First, X-gal is an activatable probe. X-gal alone is colorless. Strong optical absorption is generated only after it is cleaved by β-galactosidase. Thus, clearance of the un-cleaved X-gal before imaging is not required. As a result, the activity of β-galactosidase can be measured by PAT against a minimal background. Second, this enzymatic reporter system possesses an intrinsic signal-amplification mechanism. As an enzyme, a single β-galactosidase molecule can cleave multiple X-gal molecules to produce a large number of blue product molecules, allowing us to detect a low expression level of lacZ.

**Cell Culture**

A wild-type 9 L gliosarcoma cell line was provided by Dr. Michael Welch's laboratory in the School of Medicine of Washington University in St. Louis (Commercial source: ATCC, Manassas, VA). The 9 L/lacZ cell line was created by transfecting the 9 L cell with BAG replication deficient retroviral vector carrying the *E. Coli lacZ* gene. The 9 L/lacZ cells were purchased from American Type Culture Center (Manassas, VA). The cells were cultured in Dulbecco's Modified Eagle Medium (DMEM, Invitrogen, Carlsbad, CA) supplemented with 10% fetal bovine serum (FBS, Invitrogen) and 1 mM sodium pyruvate (Invitrogen).

**Preparation of the Native Cell Lysate**

After removal of the culture medium, the 9 L/lacZ cells were rinsed with phosphate-buffered saline (PBS, Invitrogen) and then collected into a 15 ml centrifuge tube. The cells were harvested after centrifuging at 4°C for 5 minutes and discarding the supernatant. Then we mixed ice-cold RIPA lysis buffer (Sigma-Aldrich, St. Louis, MO) with the cells and incubated them in ice for 40 minutes. During the incubation, the solution was re-suspended every 10 minutes. Finally, the solution was centrifuged for 20 minutes at 4°C. The supernatant, containing the active natively expressed β-galactosidase, was collected as the lysate of 9 L/lacZ cells.

**Animal Preparation for Imaging**

This animal study was approved by the Washington University in St. Louis Animal Studies Committee. For tumor inoculation, a 5-µl PBS suspension containing 5 million wild-type 9 L or 9 L/lacZ cells was subcutaneously injected into anaesthetized mice (Hsd:Athymic Nude-Foxn1NU, Harlan) using a 0.3-ml syringe with a 29-gauge needle. Tumors were left to grow and monitored daily. Imaging was performed after the progression of the tumor became obvious. Hair was removed before imaging. For detecting the expression of lacZ, ~0.4 mg of X-gal in 20 µl PBS and DMSO (1:1) was injected near the tumor 1 day before imaging. During imaging, the animals were was maintained under anesthesia by a gas mixture of oxygen and 1% isoflurane flowing at a rate of 1 L/min, and their blood oxygenation and heart rates were closely monitored. All mice recovered naturally after the experiments without photodamage.

**Macroscopic Photoacoustic Imaging [14]**

This system is capable of performing both photoacoustic and ultrasonic imaging of the same sample. The photoacoustic signal was excited by 6.5-ns laser pulses generated at a 10-Hz repetition rate and a wavelength of 650 nm by a tunable dye laser (PrecisionScan-P, Sirah) pumped by a Q-switched Nd:YAG laser (PRO-350-10, Newport). Light fluence on the tissue surface was measured to be ~2 mJ/cm², 10 times lower than the ANSI safety limit (20 mJ/cm²). The acoustic signal was detected by a 128-element linear ultrasonic array with a nominal bandwidth of 4–8 MHz (L8-4, Philips Healthcare). The imaging probe was scanned along the elevational direction to obtain a volumetric image. For each 2D frame of the photoacoustic image, measurements were averaged for 100
times, requiring an acquisition time of 10 s. Photoacoustic images were reconstructed using a Fourier-domain beam-forming algorithm [15]. The in-plane resolutions of this system are ~1.0 mm and ~0.4 mm in the lateral and axial directions, respectively, at 5 cm deep in biological tissue [16].

Acoustic-resolution Photoacoustic Microscopy (AR-PAM) [8], [17]

A tunable dye laser (CBR-D, Sirah), pumped by a Nd:YLF laser (INNOSLAB, Edgewave), provided laser pulses at two wavelengths, 584 nm and 635 nm. The pulse width was 7 ns, and the maximum repetition rate was ~5 kHz. The light focus coaxially overlapped with the focus of a high-frequency ultrasonic transducer (V214-BB-RM, Olympus NDT. Central frequency: 50 MHz). The incident energy density at the tissue surface was controlled to be below 6 mJ/cm². A mechanical stage drove the raster scanning of the imaging probe to obtain a volumetric dataset without averaging. The maximum photoacoustic amplitudes along each axial line were projected onto the skin surface to form a maximum-amplitude projection (MAP) image. The current system was quantified to have a lateral resolution of 45 µm and an axial resolution of 15 µm, and was capable of imaging ~3 mm deep into the skin [17].

Optical-resolution Photoacoustic Microscopy (OR-PAM) [18]

For photoacoustic excitation, a compact diode-pumped Nd:YVO₄ laser (SPOT 100–532, Elforlight, UK. Maximal pulse repetition rate: 50 kHz) generated 1.2-ns pulses at a wavelength of 532 nm. The light was delivered by a single-mode optical fiber and focused into a diffraction-limited spot inside the tissue. The generated photoacoustic signal was detected by an ultrasonic transducer (Central frequency: 40 MHz; NA: 0.5) in transmission mode. The scanning mode was the same as AR-PAM and could also form an MAP image after scanning. The lateral resolution of this setup was quantified to be 0.40 µm [18].

RESULTS

Macroscopic Photoacoustic Imaging

Figure 2(a) shows a photograph of a tumor-bearing nude mouse before imaging. 9 L tumors with and without lacZ transgene were grown in its right and left flanks, respectively. Both tumors were injected with the same amount of X-gal before imaging. Once the mouse was sacrificed after imaging, we removed the skin on top of the tumors, and found only the lacZ + tumor was stained blue, which further confirms the specificity of the detection of the lacZ expression by the chromogenic X-gal probe [Figure 2(b)]. The tumor-bearing region of the mouse was imaged in vivo with both photoacoustic and ultrasonic imaging. Figures 2(c–e) show the composite dual-modality images in volumetric visualization, a cross-sectional view, and projection view, respectively. Therein, photoacoustic images are colored green, while ultrasonic images are colored in gray scale. Figure 2(c) shows we could localize the lacZ + tumor in three dimensions. Also, we observed that although the tumor showed as a hypoechoic area in ultrasonic images, the lack of ultrasonic scattering was not a specific indicator for tumor [Figure 2(d)]. In contrast, using the lacZ/X-gal reporter strategy, PAT could detect the lacZ-marked tumor with a high specificity [Figure 2(e)]. Moreover, by overlaying chicken breast tissue on top of the tumor, we found the expression of lacZ remained visible at a depth of 5.0 cm in biological tissue, with a contrast of 3.04.

Figure 2. Imaging 9 L tumors with and without the lacZ reporter gene by an array-based photoacoustic and ultrasonic imaging system in vivo.
Acoustic-resolution Photoacoustic Microscopy

Blood vessels and the blue product have well-separated absorption peaks, and can be separately visualized by PAT at different optical wavelengths (Figure 3). Two wavelengths (584 nm and 635 nm) were selected to maximize the difference between the optical absorption of hemoglobin and the blue product within the efficient emission band of the DCM laser dye. While the 635-nm wavelength was selected to map the lacZ-marked tumor [Figure 3(a)], the 584-nm wavelength was used to visualize the microvasculature [Figure 3(b)]. The photoacoustic signal in Figure 3(b) represents the relative value of total hemoglobin concentration. A combined image [Figure 3(c) and 3(d)] shows the spatial relation between the tumor and the surrounding microvasculature.

Figure 3. Simultaneously imaging a lacZ-marked tumor and its associated microvasculature by dual-wavelength AR-PAM in vivo.

(a) MAP mage acquired at 635 nm showing tumor morphology. (b) MAP image acquired at 584 nm showing microvasculature. (c) Composite image showing the spatial relations between tumor and blood vessels. (d) 3D depiction of composite PA image showing the tumor and blood vessels (Video S1, MPEG, 5.75 MB). Green: tumor. Arrows indicate feeding vessels of the tumor. The scale bars represent 2 mm.

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**DISCUSSION**

Using the macroscopic photoacoustic system, we imaged the expression of lacZ at a depth of 5.0 cm in biological tissue. This demonstrated the advantage of PAT over other optical molecular imaging techniques, such as BLI and FLI, in penetration depth. In addition, all images were obtained by averaging over 100 measurements, but using a low optical energy at 1/10 of the ANSI exposure limit. Assuming noise was uncorrelated between measurements, we could potentially obtain images with the same quality at a speed of 10 frames/second while using the maximum laser energy allowed by the ANSI standard. The imaging speed can be further improved by adopting a high-repetition rate laser.

We further demonstrated simultaneous imaging of a lacZ-marked tumor and its surrounding microvasculature by dual-wavelength AR-PAM. Remodeling of the microcirculation plays an important role in oncology. Although photoacoustic computed tomography is able to detect the lacZ-marked tumor at great depth, it lacks the resolving power to visualize the microcirculation system associated with the tumor. AR-PAM uses a high-frequency focused transducer, and can achieve sufficient resolving power to visualize the microvasculature at the cost of imaging depth.

It is noted that the reaction product of X-gal is almost insoluble in water and is thus difficult to be cleared by the body. However, by using other galactopyranosides (e.g., chlorophenolred-β-D-galactopyranoside (CPRG) and ortho-Nitrophenyl-β-galactoside (ONPG)) whose cleavage products by β-galactosidase are water-soluble, chronic monitoring of gene expression in a single sample at multiple time points can be achieved.

We showed the sub-cellular localization of the lacZ expression using OR-PAM. By confining photoacoustic excitation...
by tight optical focusing, OR-PAM has been proven to obtain a lateral resolution of a fraction of a micron [18], and could potentially play an important role in studying molecular biology in cultured cells.

Currently, each of the imaging systems is independent, which means one PAT system achieves only one specific resolution. Thus, a single PAT system with switchable resolutions and penetration depths is of great interest. To integrate a macroscopic photoacoustic imaging system with AR-PAM, we simply need to replace the 50-MHz transducer in AR-PAM with a lower frequency transducer and adjust the laser exposure accordingly. Previous work has demonstrated that an AR-PAM with a 5-MHz transducer provides penetration up to 38 mm in chicken breast tissue. At the 19-mm depth, the axial resolution is 144 µm and the lateral resolution is 560 µm [19]. To integrate AR-PAM with OR-PAM, the laser output can be split into two beams, and a fiber bundle can be used to switch illuminations for AR-PAM and OR-PAM. A single core of the fiber bundle can deliver light for OR-PAM with high lateral resolution, while all cores of the fiber bundle can illuminate a larger area for AR-PAM. An integrated AR-PAM and OR-PAM system is under construction by our group.

In summary, we demonstrated several key merits of PAT as a promising candidate for molecular imaging. We proved that the expression of lacZ can be detected by PAT as deep as 5.0 cm in biological tissue. In addition, we showed that PAT could follow the gene expression at scalable depths and resolutions. We expect that PAT could become an important tool linking biological studies at the microscopic and macroscopic levels. With the future development of new chromogenic reporter gene systems, photoacoustic reporter gene imaging can impact both laboratory research and clinical practice.

SUPPORTING INFORMATION

Video S1.

3D depiction of composite AR-PAM image acquired at 635 nm and 584 nm showing the tumor and blood vessels. Green: tumor. Red: microvasculature.

(MPG)

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AUTHOR CONTRIBUTIONS

Conceived and designed the experiments: LL XC AK LVW. Performed the experiments: LL XC AK ZG TNE CZ. Analyzed the data: XC LL. Contributed reagents/materials/analysis tools: YZ YX. Wrote the paper: XC LL LVW.

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