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# Multi-scale Functional and Molecular Photoacoustic Tomography

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## Abstract

Photoacoustic tomography (PAT) combines rich optical absorption contrast with the high spatial resolution of ultrasound at depths in tissue. The high scalability of PAT has enabled anatomical imaging of biological structures ranging from organelles to organs. The inherent functional and molecular imaging capabilities of PAT have further allowed it to measure important physiological parameters and track critical cellular activities. Integration of PAT with other imaging technologies provides complementary capabilities and can potentially accelerate the clinical translation of PAT.

### Keywords

photoacoustic tomography; photoacoustic microscopy; photoacoustic computed tomography; multi-scale imaging; functional imaging; molecular imaging

## Introduction

Photoacoustic (PA, also termed optoacoustic) tomography (PAT) is a hybrid imaging modality that acoustically detects the optical absorption contrast of biological tissue [1–4]. In PAT, the object is usually irradiated by a short-pulsed laser beam. Some of the incident photons are absorbed by biomolecules (e.g., hemoglobin, water, lipids and melanin), and their energy is partially or completely converted into heat. The heat-induced pressure propagates in tissue as wideband ultrasound waves, which are detected outside the tissue by an ultrasonic transducer or transducer array to form an image that maps the original optical energy deposition in the tissue [5].

PAT seamlessly combines the rich optical absorption contrast of biological tissue with the high acoustic resolution at depths. In the optical excitation phase, a given percentage change in the optical absorption coefficient of the tissue yields the same percentage change in the linear PA signal amplitude; hence PAT has 100% sensitivity to optical absorption contrast. In the acoustic detection phase, the ultrasound waves undergo only weak scattering in tissue,

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which enables high-resolution PA imaging at depths beyond the optical diffusion limit (~1 mm in tissue) [6].

Benefiting from fast advances in laser and ultrasound technologies, PAT has developed remarkably since the early 2000s [7–9]. Almost all the key performance aspects of PAT, including spatial resolution, penetration depth, imaging speed and detection sensitivity, have been continuously improved through collaborative efforts involving engineering, mathematics, materials science, chemistry and biomedicine. Building on these technical advances, PAT has become increasingly popular in preclinical studies, especially in studying tumor angiogenesis, cancer hypoxia, brain functions, tissue remodeling and drug delivery [1, 2]. Most encouragingly, PAT has started to be used in clinical practice, for procedures such as breast cancer screening [10–12], sentinel lymph node mapping [13–18], melanoma staging [19–24] and endoscopic examination [25, 26]. All of these preclinical and clinical applications of PAT are important maturing steps towards human healthcare.

In light of the deep connections between PAT and ultrasound imaging [5, 27], this concise Review aims to introduce PAT technologies to the general ultrasound community, highlighting its inherent functional and molecular imaging capabilities. Representative PAT applications in both fundamental studies and clinical practice will be reviewed. Integration of PAT with other imaging modalities will also be discussed, and finally, potential PAT technical breakthroughs will be envisioned.

## Multi-scale PAT: seeing clearly at depths

Photon propagation in soft tissue can be loosely classified into four regimes, which are approximately related to the penetration depths of representative optical imaging modalities (Figure 1a) [5, 28]. Conventional planar optical microscopy works within the aberration limit (~100 μm, ballistic regime), where photons have undergone no scattering [5]. Modern optical imaging modalities, such as confocal microscopy, two-photon microscopy and optical coherence tomography, are limited by diffusion to ~1 mm in soft tissue (quasi-ballistic regime), where scattered photons still retain strong memory of original propagation direction [4]. Diffuse optical tomography (DOT) and PAT are able to provide penetration into the quasi-diffusive and diffusive regimes where photons have almost completely lost their memory of the original propagation direction [5, 29, 30]. However, DOT and PAT are still limited by dissipation to less than 10 cm in soft tissue, where the optical fluence ($J/m^2$) is significantly attenuated due to both absorption and scattering. Finally, when the optical scattering is largely corrected for by using wavefront engineering technologies, it is believed to be possible to conquer the dissipation limit and approach the absorption limit (~1 m) for whole-body penetration in humans [31, 32]. For brevity, the following use of “diffusive” and “ballistic” in this Review also refers to “quasi-diffusive” and “quasi-ballistic”, respectively.

In contrast, ultrasound attenuation in soft tissue is dominated by absorption instead of scattering [33, 34]. For example, at 5 MHz, the ultrasound scattering coefficient of the human skin is 0.014 $cm^{-1}$ while the ultrasound attenuation coefficient is 0.38 $cm^{-1}$. The ultrasound attenuation coefficient is approximately proportional to the ultrasound frequency

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over a wide frequency range [34–36]. In fact, a good rule of thumb is that the ultrasound attenuation coefficient for a variety of soft tissues is 0.5–1.0 dB/cm/MHz [34]. Such frequency-dependent attenuation in ultrasound imaging results in the trade-off between its spatial resolution and penetration depth. Figure 1b shows the approximate ultrasound penetration depths at typical frequencies and the corresponding biomedical applications.

Since PAT utilizes optical excitation and acoustic detection, its resolution can be either optically or acoustically determined, depending on the targeted penetration depth. Classified by how its resolution is determined, PAT has two major implementations [1]: acoustic-resolution PAT with an imaging depth beyond the optical diffusion limit; and optical-resolution PAT with an imaging depth less than the optical diffusion limit. Readers are referred to recent Review articles for comprehensive details about different PAT implementations [3, 37–39].

## Acoustic-resolution PAT

Acoustic-resolution PAT targets deep tissue imaging with acoustically determined resolutions in all dimensions. Acoustic-resolution PAT can be implemented by raster-scanning weakly focused optical illumination and a spherically focused ultrasonic transducer, a technology typically referred to as acoustic-resolution photoacoustic microscopy (AR-PAM). Acoustic-resolution PAT can also be implemented by using wide-field optical illumination and parallel acoustic detection with an ultrasonic transducer array, a technology typically referred to as photoacoustic computed tomography (PACT). In acoustic-resolution PAT, both the spatial resolution and imaging depth are highly scalable with the ultrasound frequency. PAT operating with lower ultrasound frequency can penetrate deeper with relaxed resolution.

As a rule of thumb, different acoustic-resolution PAT embodiments have generally achieved a depth-to-resolution ratio (DRR) of ~200 [1]. For example, by using a 50 MHz focused ultrasonic transducer, the first AR-PAM achieved a lateral resolution of ~45 μm, an axial resolution of ~15 μm and an imaging depth of ~3 mm in tissue [8, 40]. Such an imaging depth is sufficient for melanoma staging [40] and pain treatment evaluation [41] in the human skin. By using a similar design but with a 5 MHz transducer, photoacoustic macroscopy (PAMac) increased the imaging depth to ~38 mm, with a relaxed lateral resolution of ~500 μm and axial resolution of ~144 μm [42, 43]. This imaging depth is sufficient for whole-body mouse imaging (Figure 2a) [43] and deep sentinel lymph node mapping [13–18].

Acoustic-resolution PAT has proven powerful in deep tissue imaging, such as human breast cancer detection [11–13, 44–46], prostate adenocarcinoma imaging [47, 48], atherosclerotic plaque characterization [26], ophthalmic imaging [49], gastrointestinal (GI) tract imaging [50], small animal whole-body imaging [51–53] and thyroid imaging [54]. Of all the clinical translations of PAT, human breast imaging has progressed furthest. So far, several groups have reported PA breast imaging on a total of more than 100 patients [37]. To accommodate the shape of the uncompressed breast, Kruger and his colleagues used a semi-spherical transducer array with rotational scanning for dense spatial sampling (Figure 2b) [12].

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However, this configuration suffers from low imaging speed and may have difficulty in detecting deep tumors. Kitai and his colleagues used a 2D planar transducer array, where the breast was gently compressed from the side between a glass slide and the transducer array [45]. The side compression can reduce the effective beast thickness and thus help detect deeper tumors. However, the planar detection geometry suffers from limited view for accurate reconstruction. Alternatively, our laboratory has developed an integrated PAT and thermoacoustic tomography (TAT) system, where the breast is compressed from the front (the nipple side) to form a cylindrical shape [55]. The illumination is directed from the front, and the ultrasonic transducers scan around and along the cylindrical breast to obtain a full 3D data set. In this configuration, deep tumors close to the chest wall can potentially be imaged.

## Optical-resolution PAT

Although acoustically determined resolutions in acoustic-resolution PAT are adequate for many biomedical applications, it becomes challenging to improve the spatial resolutions to the cellular level by simply increasing the ultrasound frequency without severely compromising the penetration [34, 56, 57]. Different from ultrasound imaging, PAT can use fine optical focusing to provide optically defined lateral resolution within the optical diffusion limit, while the axial resolution is still derived from the time-resolved ultrasonic detection.

Optical-resolution PAT is traditionally referred to as optical-resolution photoacoustic microscopy (OR-PAM) [52, 58–74]. Limited only by optical diffraction, the lateral resolution of OR-PAM can easily reach the cellular and subcellular level, and is scalable with the optical wavelength and the numerical aperture (NA) of the optical objective. The first OR-PAM, reported in 2008, achieved a lateral resolution of ~5 μm (objective NA: 0.1 in air) and an imaging depth of ~1 mm. Single capillaries in a mouse ear can be clearly resolved (Figure 2c) [64]. The lateral resolution of OR-PAM was later improved to ~220 nm by using a water-immersion objective (objective NA: 1.23 in water), allowing single red blood cells to be clearly resolved (Figure 2d) [75]. Further, in recently developed optical-resolution PACT systems, the field of view is simultaneously excited by an array of diffraction-limited optical foci, and the resultant PA waves are detected by a linear- or ring-shape ultrasonic transducer array [69, 76, 77]. The parallel excitation and detection enable fast wide-field imaging, while the imaging depth is still restricted to ~1 mm.

Recently, OR-PAM has succeeded in sub-optical-diffraction imaging, taking advantage of various nonlinear mechanisms such as two-photon absorption [78–80], photobleaching [81], absorption saturation [82] and thermal relaxation [82, 83]. Nonlinear OR-PAM has provided optically determined resolutions in all dimensions including the axial or depth direction. In particular, a sub-diffraction lateral resolution of ~80 nm has been achieved by photoacoustic nanoscopy using the absorption saturation effect, enabling single mitochondria in fibroblast cells to be resolved (Figure 2e) [82]. Meanwhile, an axial resolution on the level of sub-micrometers has been achieved, which is about two orders of magnitude finer than the acoustically determined axial resolution in traditional linear OR-PAM [81, 82].

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## Functional and molecular PAT: seeing more than anatomy

While functional ultrasound imaging is largely limited to Doppler ultrasound and molecular ultrasound imaging mostly detects microbubbles in the vasculature [27], PAT is inherently suited for diverse functional and molecular imaging, with a wealth of endogenous and exogenous contrasts.

## Functional PAT

So far, PAT has measured a number of functional parameters at various length scales, including total hemoglobin concentration [12, 63, 84–86], blood oxygenation [85–92], temperature [93–95], blood flow [96–102], pH [103–106], blood glucose level [107], pulse-wave velocity [108] and metabolic rate of oxygen ($MRO_2$) [22, 109, 110]. Here we discuss several widely measured functional parameters, especially for cancer diagnosis and therapy.

Hemoglobin in red blood cells carries most of the oxygen needed by the body to power its functions [111]. Using hemoglobin as the endogenous contrast, PAT can quantify the total hemoglobin concentration (HbT) and oxygen saturation of hemoglobin ($sO_2$) with high sensitivity (Figures 3a–b) [112]. For accurate measurement of absolute HbT and $sO_2$, it is necessary to correct for light attenuation by using empirical or model-based methods [89–91, 113–116]. Recently, single-wavelength-based and dynamics-based PA methods have also been developed for absolute $sO_2$ measurement [88, 117]. Notably, HbT and $sO_2$ measurements are proven useful in cancer diagnosis and prognosis. For example, in PA breast cancer imaging, increased HbT in the tumor region is highly correlated with cancer angiogenesis (Figure 3c) [45], while decreased $sO_2$ in the tumor core typically indicates an adverse cancer progression [118, 119].

During thermotherapy, it is necessary to monitor the local temperature for safe deposition of heat and efficient destruction of abnormal cells. Using the temperature dependence of the Grueneisen parameter [120–127], PAT can measure temperature changes from relative changes in PA amplitude alone [93, 128]. A temperature detection sensitivity of 0.15 °C was achieved with a temporal resolution of 2 seconds [93]. Furthermore, a recently published method can quantify the absolute temperature in deep tissue with AR-PAM and PACT, using the dual temperature dependences of the Grueneisen parameter and the speed of sound [129, 130]. In addition to measuring temperature, the temperature dependence of PA signal can also be used to measure the blood flow speed in deep tissue [101, 131], enhance the spatial resolution of OR-PAM [82] and ameliorate the limited-view problem in linear-array-based PACT [131].

Blood flow ensures the transportation of oxygen, nutrients and metabolic wastes throughout the body. PAT can measure blood flow by using either Doppler methods [97, 99, 102, 132–135] or feature-tracking methods [98, 100, 101, 136, 137]. Compared with ultrasound flowmetry, PA flowmetry has much higher detection sensitivity because of the excellent contrast between the blood vessels and the surrounding tissue. A flow detection sensitivity of 50 μm/s has been demonstrated by using a structured-illumination PA method [135]. In addition, thermal-tagging methods have measured flow speeds at 5 mm depth in tissue [100, 101, 136, 137]. Consequently, by simultaneously measuring multiple parameters such as

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HbT, $sO_2$ and blood flow, PAT can quantify the metabolic rate of oxygen ($MRO_2$) of tissue, which can potentially be used for early cancer detection [22, 109, 110].

## Molecular PAT

In addition to functional imaging, by using targeted or untargeted contrast agents, PAT can track molecular processes in living organisms [21, 138–145]. The choice of contrast agents for molecular PAT is much greater than for molecular ultrasound imaging, because all molecules have their own absorption wavelengths, hence they can be imaged by PAT [38, 39]. In addition, many molecules can reach the extravascular space. So far, many contrast agents have been imaged in molecular PAT [37, 38, 146], including organic dyes (Figure 4a), nanoparticles (Figure 4b), fluorescent proteins (Figure 4c), microbubbles and reporter gene products (Figure 4d). These contrast agents, especially nanoparticles [141], can be specifically engineered for different PA applications.

The detection sensitivity of molecular PAT is relevant to a number of factors, such as the absorption cross-section of the molecule, the sensitivity of the ultrasonic transducer, the imaging depth and the permitted light exposure [147]. Roughly, in molecular PAT, the reported noise-equivalent detectable concentration at ≥3 mm depth in tissue is on the level of millimolar for microbubbles, micromolar for organic dyes, nanomolar for proteins and picomolar for nanoparticles [44, 148–153].

Molecular PAT has been proven reliable in early cancer imaging, circulating tumor cell (CTC) detection, glucose uptake monitoring and sentinel lymph node mapping (Figure 3e) [151, 154, 155]. For example, PAT has been successfully used for detecting pigmented circulating melanoma cells, due to their strong light absorption [21, 156, 157]. Recently, non-pigmented circulating breast cancer cells have also been detected *in vivo* by PAT, using targeted gold nanoparticles [158] or genetically-encoded green fluorescent protein [159] as the contrast agent. Multi-spectral PA measurements can help unmix different molecules to enhance the detection of CTCs [142] and early-stage tumors [160]. Magnetically modulated PA detection can highlight the signals from magnetic nanoparticles targeted to tumor cells [142]. Activatable organic dyes and nanoparticles can also help detect cellular activities of interest, with improved sensitivity (Figure 3f) [54, 156].

It is worth noting that multi-wavelength illumination is often required in functional and molecular PA imaging. However, commercially available wavelength-tunable lasers (e.g., dye lasers, optical parametric oscillators, or Ti:sapphire lasers) cannot switch wavelengths at a high speed [161]. Several methods have been developed to achieve fast wavelength tuning. Dean-Ben and his colleagues customized a 50 Hz optical parametric oscillator that allowed wavelength change on a per-pulse basis for their hemi-spherical-array-based PACT [162]. Wang et al. have developed a digital-mirror-device based wavelength multiplexing method for OR-PAM with a wideband dye laser [163]. A 2 kHz wavelength tuning speed has been achieved with a wavelength tunable range of ~20 nm. In addition, two lasers with different wavelengths can serve as an alternative to wavelength tuning, with increased system cost [164].

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## Integration of PAT with other imaging modalities

A major challenge for quantitative PAT is the unknown local light fluence. This issue can potentially be addressed by integrating PAT with DOT, which measures the optical properties of the tissue. Multiple groups have reported various DOT-PAT systems for different applications [165–168]. Studies have shown that DOT allows better quantitative reconstruction in PAT [165, 167, 169]. However, the optical properties derived from DOT measurements typically possess much poorer spatial resolution than that of PAT. Iterating between PAT and DOT reconstructions may help resolve this issue [167].

Speed of sound (SOS) heterogeneities also deteriorate PAT image quality. One remedy is to combine PAT with ultrasound tomography (UST). In UST-PAT systems, UST provides the SOS map of the tissue to improve the PAT reconstruction. Unlike the combination of PAT with DOT, which requires additional light sources and detectors, the addition of UST to PAT can be implemented with existing ultrasonic transducers as long as the ultrasonic transmission capability is enabled [170]. Alternatively, PAT can be integrated with commercial UST by adding a pulsed light source [13]. In addition to heterogeneous SOS, acoustic attenuation and aberration also deteriorate PAT image quality [170–173], especially in human brain PAT; but their effects can be potentially investigated with UST-PAT as well.

In addition to correcting for optical and acoustic inhomogeneities, different PAT embodiments have also been integrated with other imaging modalities to provide complementary contrasts [174]. Deep-penetration imaging modalities, such as magnetic resonance imaging (MRI), X-ray computed tomography (CT) and positron emission tomography (PET), have been used in conjunction with PAT for studies involving multimodality contrast agents [175–177]. PAT has been integrated with confocal microscopy [178, 179], optical coherence tomography [180–183] and ultrasound imaging [13, 18, 26, 50, 184–186], sharing either the same optical components or ultrasonic transducer(s) (Figure 5a) [50]. A recently developed tri-modality optical imaging system combines OR-PAM with a commercial confocal and two-photon microscopic system (Figure 5b) [187]. The three imaging modalities can readily share the optical illumination path and scanning system.

## Discussion

In conclusion, PAT is a unique imaging modality that complements other imaging techniques: rich optical absorption contrast provides inherent functional and molecular imaging capabilities, and acoustic detection allows high resolution imaging at depths. Although we can cover only several representative studies in this concise Review, they have clearly showed the scale and momentum of PAT development. Here, we will also discuss a few potential breakthroughs of PAT technologies.

One new frontier is PA-based optical wavefront engineering. The ability to focus light deep into tissue will have tremendous impacts on imaging and therapy. By using PA signals as the feedback, the wavefront of the excitation light can be optimized through iterative algorithms so that it can be confined to either an acoustically or optically determined region

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in the tissue [188, 189]. The advantage of PA detection is its high resolution and deep penetration, while the slow optimization process remains to be a major challenge.

Miniaturizing the PA probe is essential for endoscopic and intravascular applications. Recently, several PA endoscopes have been developed with optical detection of the acoustic signals [190–193]. Since the detection sensitivity of these optical detectors (e.g., micro-ring resonators and Fabry-Perot sensors) is not directly related to the detector size, the PA probe size can be significantly reduced. However, none of those optical detectors have been used for acoustic-resolution endoscopic imaging in deep tissue, due to the lack of acoustic focusing capability. Alternatively, internal optical illumination through the body cavity can be combined with external acoustic detection [194], avoiding the problem of acoustic detector size.

There is also great interest in hand-held PA probes for imaging surface regions such as the face, neck and arm, which are difficult to access with traditional table-top PAT systems. Several hand-held PA probes have been developed for microvascular imaging of port-wine stains on the face [195, 196], blood oxygenation imaging of the jugular vein in the neck [197], sentinel lymph node mapping in the breast [198], and melanoma imaging on the skin [199, 200]. For these hand-held PA probes, the imaging speed still needs to be improved to mitigate motion artifacts from both the patient and operator.

Although PAT is relatively inexpensive compared with MRI and PET, further reducing its cost can accelerate clinical translation. The most expensive components in PAT are typically the pulsed laser source and the ultrasonic transducer array. Several groups have explored inexpensive light sources [183, 201–205]. For example, Li et al. have developed an OR-PAM system using a low-cost Blue-ray DVD pickup head with a 405 nm laser diode [205]. An inexpensive ultrasound detector is also of interest, because the ultrasound receiving system becomes significantly more costly as the number of transducer elements increases. A PAT system based on low-cost acoustic delay lines for acoustic detection has been reported [206, 207]. Nevertheless, the acoustic coupling efficiency and the number of the acoustic delay lines still need to be increased.

At last, with its nonionizing radiation, high image quality, and functional and molecular imaging capabilities, PAT holds great promise for disease detection, staging and treatment evaluation. Although many challenges remain, none of them are beyond reach. PA breast cancer screening is now most ready for clinical translation. PA endoscopic examination of the GI tract will provide functional information at unprecedented depths. PA melanoma detection at depths up to a few millimeters will significantly improve the cancer's staging accuracy. Finally, PA human brain imaging through intact scalp and skull, although technically challenging, is considered one of the most promising tools for functional brain studies, with breakthroughs expected soon.

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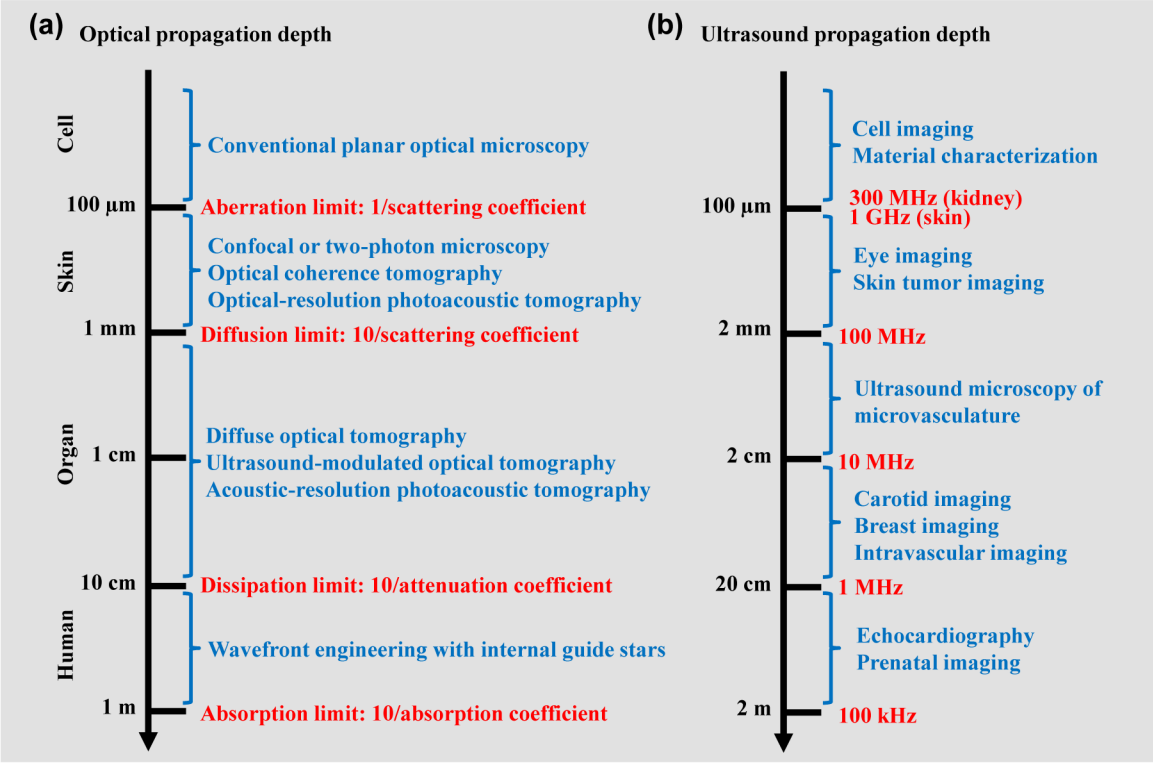


**Figure 1. Photon and ultrasound propagation regimes in soft tissue**

(a) Photon propagation regimes in soft tissue, which are approximately related to the penetration depths of representative optical imaging modalities [5, 28]. The four regimes are divided at photon propagation depths of 0.1 mm, 1 mm, 10 cm and 1 m, with typical optical absorption coefficient of 0.1 $cm^{-1}$, optical scattering coefficient of 100 $cm^{-1}$ and anisotropy of 0.9. The classification holds in optical scattering dominant media. (b) Ultrasound propagation regimes at typical ultrasound frequencies in soft tissue, with corresponding biomedical applications [34]. The ultrasound attenuation coefficient is approximately proportional to the ultrasound frequency up to at least 300 MHz for skin [35] and 100 MHz for kidney [36]. Here, the −10 dB propagation depth at the frequency up to 100 MHz is estimated with an ultrasound attenuation coefficient of 0.5 dB/cm/MHz. The propagation depths at 300 MHz for kidney and 1 GHz for skin are extrapolated based on the literature data [35, 36], taking into account the ultrasound attenuation by water.

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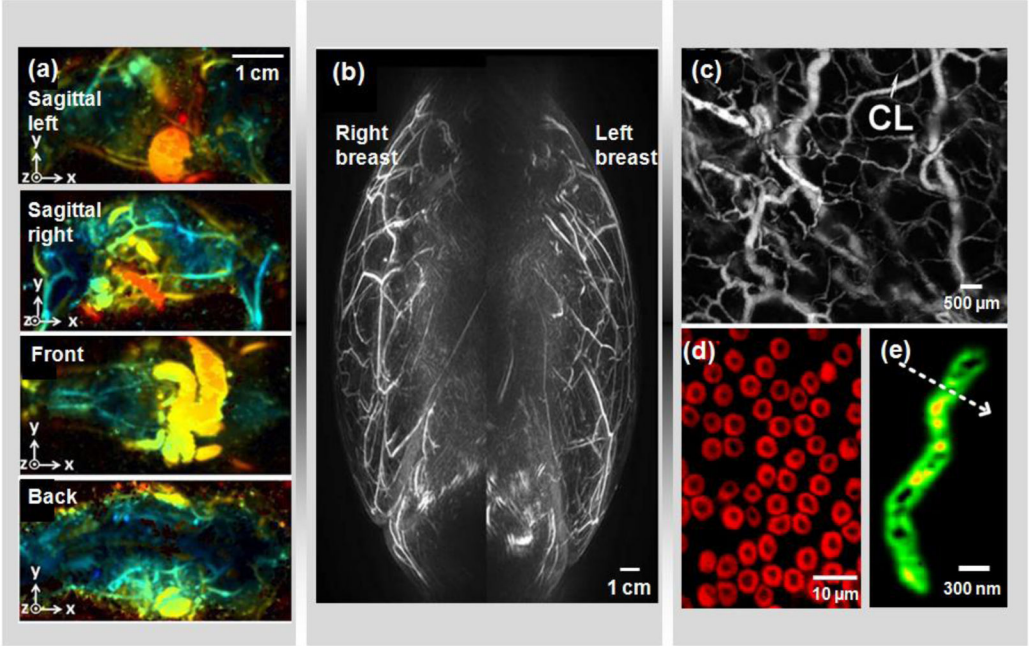


**Figure 2. Multi-scale PAT**

(a) Whole-body PAT of a mouse *in vivo* [43]. Acquired at 1064 nm, the maximum amplitude projection (MAP) images were extracted from sagittal, front and back views. The depths are color encoded from blue (shallow) to red (deep). (b) Medial-lateral MAP image of breasts of a healthy volunteer [12]. A semi-spherical transducer array was used with rotational scanning. (c) Optical-resolution photoacoustic microscopy (OR-PAM) of mouse ear vasculature, where single capillaries (CL) can be clearly resolved [64]. (d) Sub-wavelength OR-PAM of single red blood cells [75]. A lateral resolution of ~220 nm was achieved. (e) Photoacoustic nanoscopy of a mitochondrion in a fibroblast cell [208]. A lateral resolution of ~80 nm was achieved by using the optical absorption saturation effect. Images were adapted with permission from references [12, 43, 64, 75, 208].

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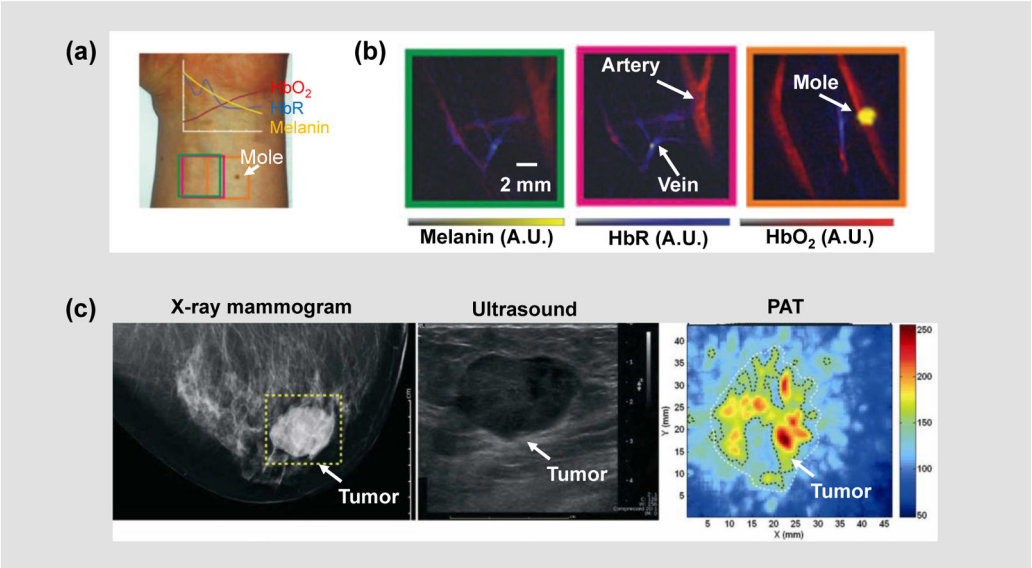


**Figure 3. Functional PAT**

(a) A photograph of a healthy volunteer's forearm, from which hand-held PACT images were taken [162]. The colored boxes correspond to the regions of interest imaged by PACT. HbR, deoxy-hemoglobin; $HbO_2$, oxy-hemoglobin. (b) Spectrally unmixed PA oxygenation images of the three boxed regions shown in (a). (c) PAT of an invasive ductal carcinoma in the right breast of a 57-year-old woman [45]. Left to right: X-ray mammogram (left); ultrasound image (middle); and PAT image (right). The higher PA signal strength in the PAT image was attributed to tumor angiogenesis. Images were adapted with permission from references [45, 162].

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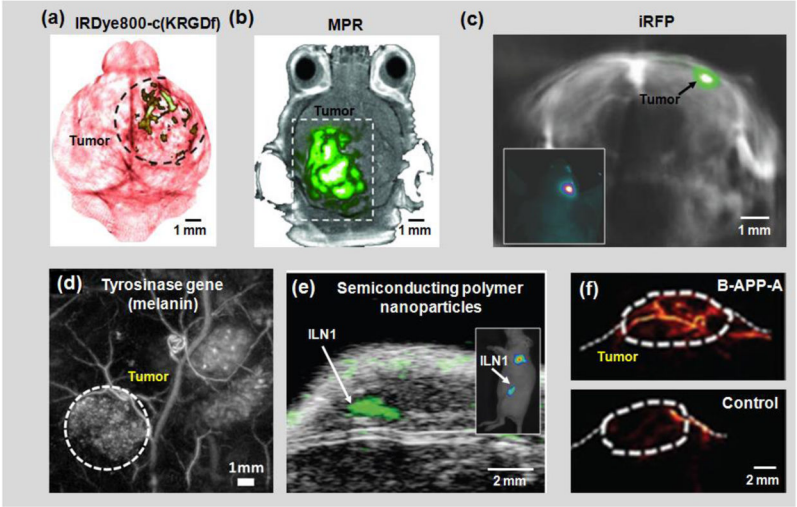


**Figure 4. Molecular PAT**

(a) PAT of a glioblastoma in a mouse brain enhanced by IRDye800-c(KRGDf), which targeted overexpressed integrin $\alpha_{v\beta3}$ in tumor cells [7]. (b) PAT of a glioblastoma in a mouse brain enhanced by tri-modality MRI-PA-Raman (MPR) nanoparticles [209]. (c) PAT of an iRFP-expressing glioblastoma in a mouse brain, 25 days post implantation [160]. The spectrally unmixed signals from the tumor (shown in green) are superimposed on the signals from blood (shown in gray). iRFP, infrared fluorescent protein. Inset: epi-fluorescence image of the mouse brain showing the tumor location. (d) PAT of a subcutaneous tyrosinase-expressing tumor [210]. The tumor is marked by the dashed circle. (e) Ultrasound (shown in gray) and photoacoustic (shown in green) co-registered image of mouse lymph nodes following tail vein injection of semiconducting polymer nanoparticles (SPN) [211]. ILN, inguinal lymph node. Inset: epi-fluorescence image of the mouse showing the ILN location. (f) PAT of the activatable B-APP-A in a FTC133 tumor in the mouse hind leg after tail vein injection of B-APP-A (top) and the control probe (bottom) [54]. Images were adapted with permission from references [7, 54, 160, 209–211].

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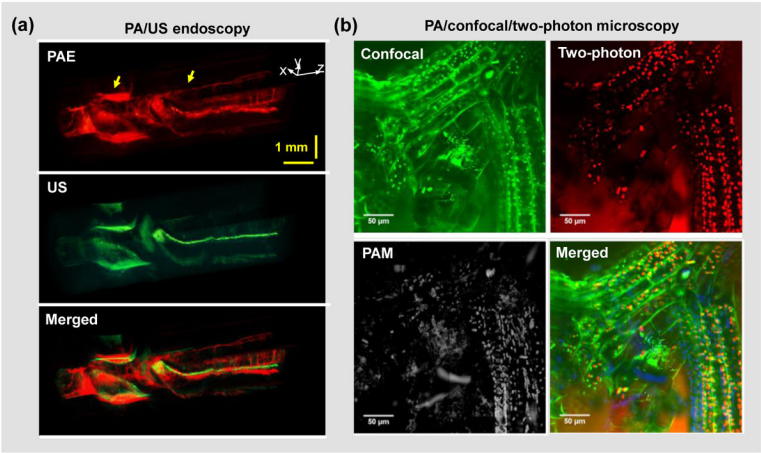


**Figure 5. Integration of PAT with other imaging modalities**

(a) Integrated PA/ultrasound (US) endoscopy of the upper esophagus of a rabbit *in vivo* [50]. The PA and US imaging share the same ultrasound transducer and scanning system. (b) Integrated PA/confocal/two-photon microscopy of a moss leaf [187]. The three imaging modalities share the same optical paths and scanning system. Images were adapted with permission from references [50, 187].