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A study of reconstruction in photoacoustic tomography with a focused transducer

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ABSTRACT

So far most rigorous reconstruction algorithms for photoacoustic tomography (PAT), e.g., the modified back-projection algorithm, have been developed based on ideal point detectors. However, a flat unfocused transducer is commonly used in PAT, thus suffering from the finite aperture effect - tangential resolution deteriorates as the imaging point moves away from the scanning center. Based on a virtual-point-detector concept, we propose a PAT reconstruction with a focused transducer to improve the degraded tangential resolution. We treat the focal point of the focused transducer as a virtual-point detector, which means that delays applied in reconstruction are relative to the focal point. The geometric focus defines propagation path of photoacoustic signals. The simulation results show that compared with PAT with an unfocused transducer, PAT with a focused transducer having an f-number of 2.5 significantly improves tangential resolution by 29 μm up to 791 μm at the imaging positions of at least 4 mm away from the scanning center. The farther the imaging positions away from the scanning center, the larger the improvement. In the region of 4 mm away from the scanning center, PAT with a focused transducer slightly degrades the tangential resolution by up to 70 μm. The improvement in tangential resolution comes with a compromise of loss in radial resolution by 26 μm up to 79 μm depending on the distance from the scanning center. In terms of the significant improvement in tangential resolution, the loss in radial resolution is tolerable, especially for imaging of big objects, e.g., breast.

Keywords: Photoacoustic tomography, focused transducer, virtual detector, finite aperture effect

1. INTRODUCTION

Photoacoustic imaging is an emerging hybrid bio-photonic imaging technique that detects absorbed photons ultrasonically through the photoacoustic effect. The marriage of ultrasound and light in this technique overcomes the resolution drawback of pure optical imaging due to overwhelming light scattering in biological tissues, and possesses the merit and most compelling features of both optics and ultrasound—namely, high optical absorption contrast and sub-millimeter ultrasound resolution—up to an imaging depth of centimeters. Photo-acoustic imaging has been applied to vasculature structural imaging[1], breast tumor detection[2,3], oxygenation monitoring in blood vessels[4,5], epidermal melanin measurement[6,7] etc. Photoacoustic signals are induced as a result of transient thermo-elastic expansion when biological tissues absorb the pulsed laser energy[8]. These signals are then detected by acoustic transducers and reconstructed to form images representative of optical absorption distribution.

There are three modes of photoacoustic imaging, which have been introduced in the literature: the tomography mode[1], the forward mode, and the backward mode[8,9]. In this study, we focus on the tomography mode – photoacoustic tomography. In a typical photoacoustic tomography (PAT) setup, a short pulse is expanded and then used to illuminate the imaging object from the top, and a ultrasonic transducer is circularly scanned to collect photoacoustic signals required for image reconstruction of distribution of laser energy deposition. So far most rigorous PAT reconstruction

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algorithms, e.g., the modified back-projection algorithm[10] and the Radon transform with far field adaptation[11], have been developed based on "ideal point detectors", which can receive ultrasound from a very large acceptance angle. Nonetheless, in reality a flat unfocused transducer is commonly used due to the issue of poor signal-to-noise ratio from a point detector (or an unfocused transducer with the aperture size close to an acoustic wavelength). The major problem of photoacoustic tomography with a flat unfocused transducer is that it suffers from the "finite aperture effect"[12], as illustrated in figure 1. The aperture of the unfocused transducer blurs the tangential resolution greatly as the imaging point moves away from the circular-scan center while the PAT with an ideal point detector offers uniform point spread functions over all the imaging area. An analytic explanation of the finite aperture effect in PAT reconstruction can be found in the reference 12. The deteriorated tangential resolution can be approximated by the triangle formed by the transducer aperture and the circular-scan center. Such deteriorated tangential resolution is not proper for imaging of big objects, such as breast.

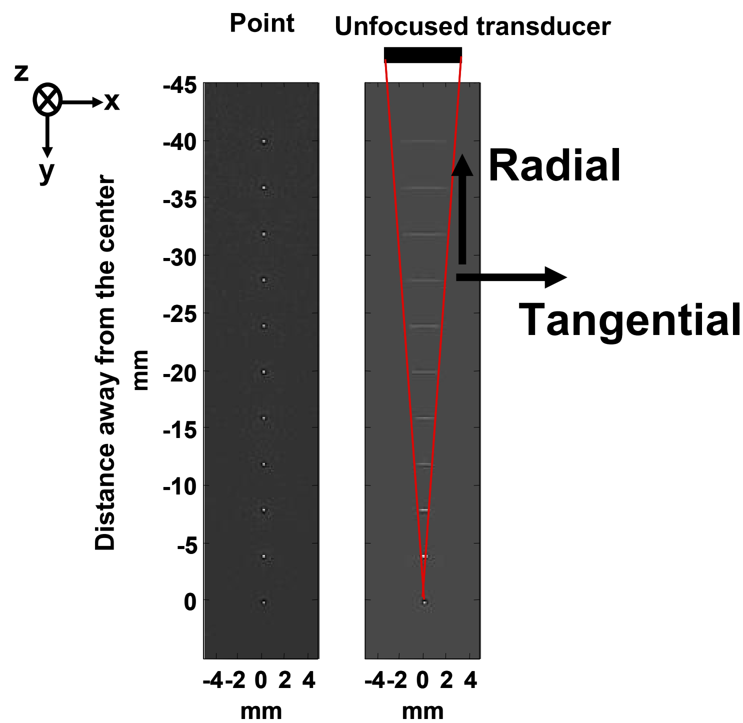

Figure 1. Illustration of the finite aperture effect in reconstruction of photoacoustic tomography. Coordinate (0,0) is the circular-scan center.

Previously, we presented a virtual-detector synthetic aperture focusing technique for photoacoustic microscopy with a focused transducer to improve the degraded resolution in the out-of-focus region[13]. Here, we extend this idea – a virtual point detector to photoacoustic tomography. Based on the virtual point detector concept, we propose a PAT reconstruction with a focused transducer to improve the degraded tangential resolution due to the finite aperture effect, as shown above.

2. VIRTUAL-POINT-DETECTOR BASED PAT RECONSTRUCTION

2.1 Virtual-point-detector concept

The virtual-point-detector concept enables reconstruction of photoacoustic tomography using a focused transducer. The concept of the virtual point detector is illustrated in Figure 2(a), which shows a focused transducer and its associate photoacoustic radiation pattern. In most cases, the photoacoustic radiation pattern is mainly determined by acoustic focusing parameters of the transducer owing to the overwhelming scattering of light in biological tissue.

Unlike an ideal point detector, photoacoustic radiation pattern of a focused transducer has certain angular extent. When biological tissues absorb pulsed laser energy, photoacoustic waves generated within the photoacoustic radiation pattern (after the focal point) will propagate toward the focal point, and can be viewed as being detected by the focal point. Therefore, the focal point of the focused transducer can be treated as a virtual point detector. As illustrated in figure 2(b), if a circular scan is performed, the scanning radius of the virtual point detector is the scanning radius of the focused transducer minus the focal length of the focused transducer. In addition, in the virtual-detector based PAT reconstruction, the reconstruction area is the region enclosed by the scanned virtual detector, as marked by slashed line in figure 2(b).

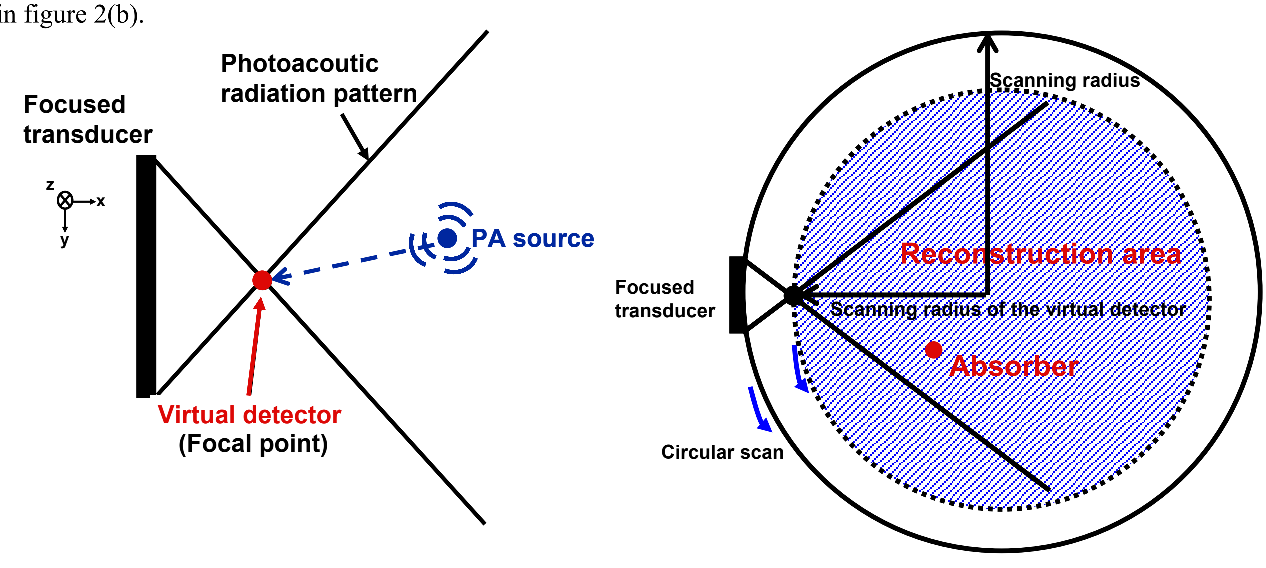

Figure 2. (a) Illustration of virtual-point-detector concept (b) Circular scan geometry of photoacoustic tomography with a focused transducer

2.2 Virtual-point-detector based back-projection

Here we only consider the 2-D imaging of a sample through a circular scan of photoacoustic signals around the imaging plane. Figure 3 shows the geometric illustration of virtual-point-detector-based back-projection for reconstruction of photoacoustic tomography, in which d_f is the focal length of the used focused transducer, r_0 is the position of the scanned focused transducer, r_0' is the position of the scanned virtual point detector (i.e., the focal point of the transducer), and r is the position of the point absorber. Note that the reconstructed absorber position r is currently limited to the region after the focal point of the transducer, i.e., the region enclosed by the virtual point detector, as shadowed by the slashed line in figure 2(b). When the scanning radius is much longer than the wavelengths of the photoacoustic waves that are useful for imaging, and the temporal profile of the short laser pulse is regarded as a Dirac delta function, for virtual-detector based PAT reconstruction, the distribution of optical absorption in the imaging plane can be reconstructed by the following modified back-projection algorithm in time domain, which is an integral over the angle θ_0 along the scanning circle[10]:

$$\phi(r) = C \cdot \int_{\theta_0} d\theta_0 \, \frac{1}{t} \frac{\partial p(r_0, t)}{\partial t} \bigg|_{t=(|r_0'-r|+d_f)/c_s},$$

where $\phi(r)$ describes the optical energy deposition at position r, $p(r, t)$ is the photoacoustic signal at position r and time t, and c_s is the speed of sound.

Note that compared with the equation reported in reference 10, the only difference is that the delay we apply in the reconstruction is calculated relative to the position of the virtual point detector (i.e., the focal point) r_0' instead of the transducer position r_0:

$t = (|r_0^{'} - r| + d_f)/c_s$.

That is, in our proposed virtual-detector-based PAT reconstruction, the back-projection of the modified quantity $(1/t)(\partial p(r_0,t)/\partial t)$ is along the circular contours (spherical shells in 3D reconstruction) that are centered at the virtual point detector (i.e., the focal point) and have a radius relative to the focal point: $|r_0^{'} - r|$, as illustrated in figure 3, because the geometric focus defines propagation path of photoacoustic signals at each scanning position. Note that in the following sections, a simple back-projection of $p(r_0,t)$ rather than $(1/t)(\partial p(r_0,t)/\partial t)$ is implemented to verify our idea.

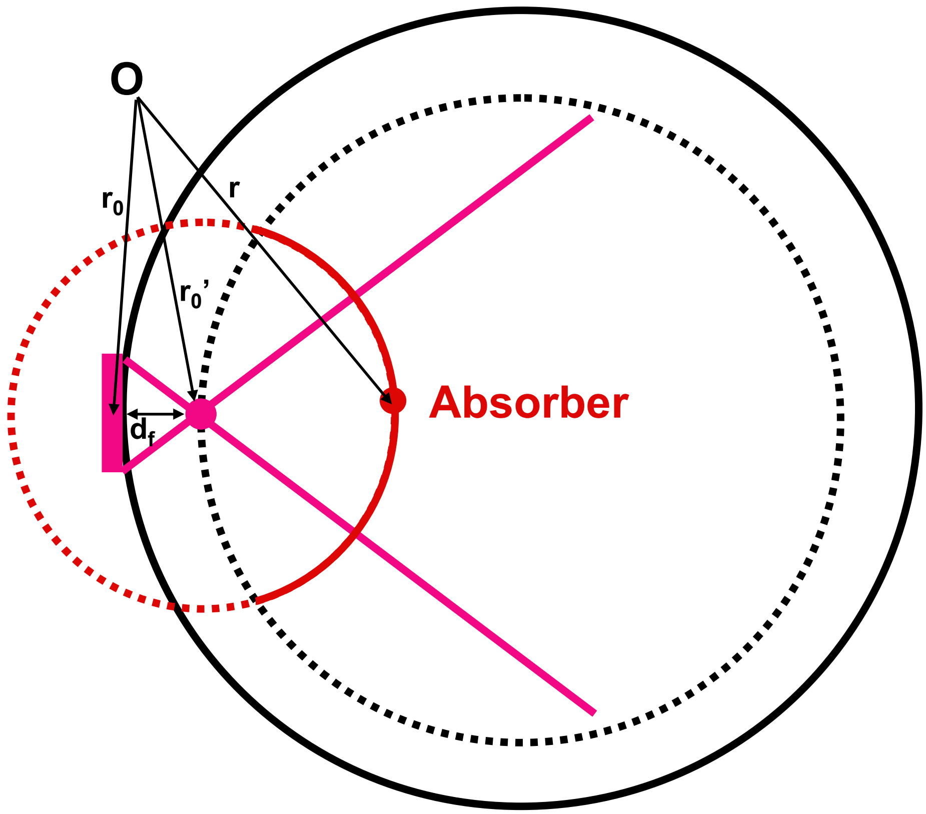

Figure 3. Geometric illustration of virtual-detector-based back-projection for reconstruction of photoacoustic tomography

4. SIMULATION RESULTS

The performance of our virtual-detector based PAT reconstruction was verified using the simulated point spread function (PSF) of PAT. The PSFs were reconstructed from simulated photoacoustic signals of point absorbers located at the imaging plane. An acoustic field simulator – Field II was modified here to simulate photoacoustic signals received by different types of detectors –point detectors, flat unfocused transducers, and focused transducers. The simulated detectors had a 5 MHz center frequency and 80% fractional bandwidth. The simulated point detector had a diameter of one 5 MHz acoustic wavelength. The diameter of the simulated unfocused and focused transducers was 6 mm. The circular scan radius was 50 mm. In both simulation and reconstruction, the speed of sound was assumed to be constant at 1.5 mm/μs.

Figure 4 shows the reconstructed PSFs of PAT with a point detector (a), an unfocused transducer (b), and a focused transducer (c). The focused transducer has a 2.5 f-number which is defined as the ratio of the focal length to the diameter of the transducer. Figures 4(a), (b), and (c) are shown in the same gray scale. Coordinate (0,0) is the circular-scan center. The lateral axis represents the tangential direction; the vertical axis is the radial direction. Figure 4(a) shows PAT with a point detector has uniform PSFs over the imaging plane while PAT with an unfocused transducer suffers the finite aperture effect – the extension of PSFs along the tangential direction (i.e., the tangential resolution) becomes wider as the imaging point moves away from the scanning center. If we compare figure 4(b) and figure 4(c), the tangential extension of PSFs from PAT with an focused transducer having an f-number = 2.5 (i.e., virtual-point-detector based PAT reconstruction) is narrower than that from PAT with an unfocused transducer from the distance of 4 mm away from the scanning center (i.e., improved tangential resolution) and is slightly wider within the distance of 4 mm away from the scanning center (i.e., slightly degraded tangential resolution).

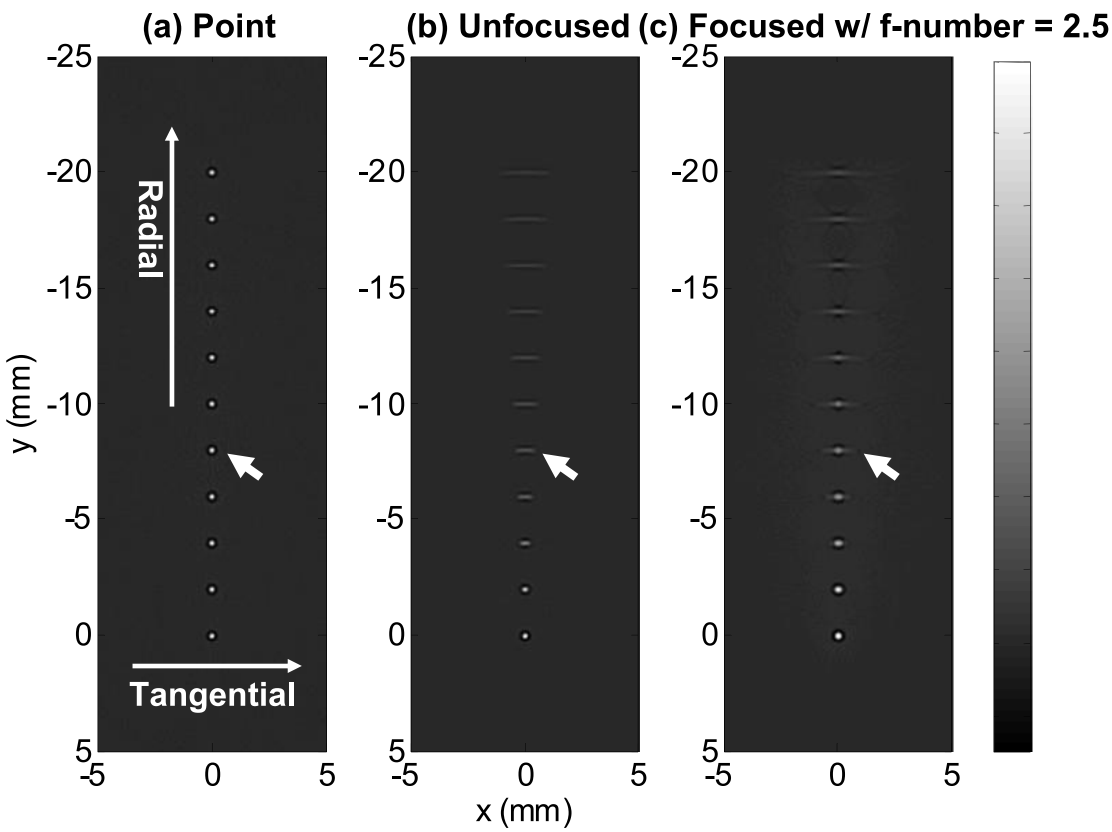

Figure 4. Photoacoustic tomography of simulated point targets by Field II. PAT images of (a), (b), and (c) are reconstructed with a point detector, an unfocused transducer, and a focused transducer with f-number = 2.5, respectively.

The 1D tangential profile and radial profile of the PSF at a distance of 8 mm away from the circular-scan center, as indicated by the white arrow in figure 4, are shown in figures 5(a) and 5(b), respectively. In figure 5(a), the lateral axis is the tangential extent, and the vertical axis is the normalized amplitude of the tangential profile. In figure 5(b), the lateral axis is the radial extent, and the vertical axis is the normalized amplitude of the radial profile. It is shown that compared with that from PAT with an unfocused transducer, there is great improvement in the mainlobe width of the tangential profile from PAT with a focused transducer although there are slight increase in the far sidelobes of the tangential profile, and in the mainlobe width of the radial profile, as shown in figure 5(b). Note that the sidelobe level of the tangential and radial profile from PAT with a point detector is much lower than those from PAT with focused and unfocused transducers although the mainlobe width in the radial profile is slightly wider than that of PAT with an unfocused transducer.

From the simulated PSFs in figure 4, the full width half maximum (FWHM) of the tangential and radial profiles at different imaging points are calculated. It is shown that compared with PAT with an unfocused transducer, PAT with a focused transducer having an f-number of 2.5 (i.e., virtual-detector based PAT) significantly improves FWHM of the tangential profile (i.e., tangential resolution) by 29 μm up to 791 μm at the imaging positions of at least 4 mm away from the scanning center. The degree of improvement depends the distance away from the scanning center. The farther the imaging positions away from the scanning center, the larger the improvement. At the imaging positions within 4 mm away from the scanning center, PAT with a focused transducer having an f-number of 2.5 slightly degrades the tangential resolution by up to 70 μm in terms of FWHM of the tangential profile. In addition, the improvement in tangential resolution also comes with a compromise of loss in radial resolution by 26 μm up to 79 μm depending on the distance from the scanning center.

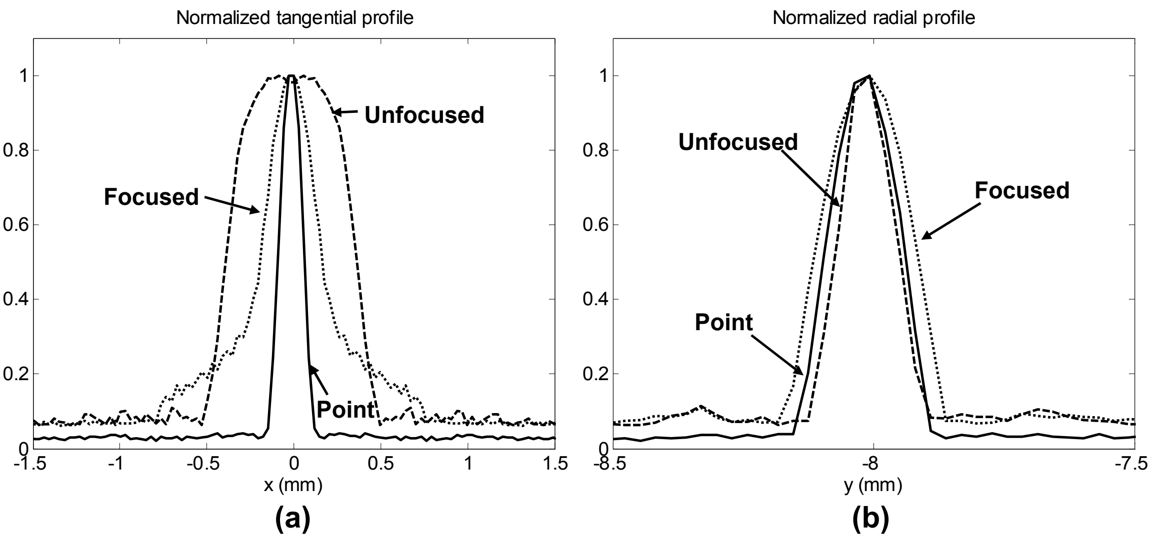

Figure 5. Tangential profile (a) and radial profile (b) for the reconstructed point spread function at a distance of 8 mm away from the circular-scan center. Solid line: point detector; dashed line: unfocused trasducer; dotted line: focused transducer with f-number = 2.5.

5. CONCLUDING REMARKS

In this paper, via the virtual-point detector concept, we show the feasibility of reconstruction of photoacoustic tomography with a focused transducer. Compared with PAT with an unfocused transducer, the PAT with a focused transducer (i.e., virtual-detector based PAT) significantly improves the tangential resolution although there is tradeoff between tangential and radial resolution. In terms of the significant improvement in tangential resolution, the slight loss in radial resolution is tolerable especially for imaging of big objects such as breast. Currently, another important parameter - the acceptance angle of the focused transducer is not incorporated into the reconstruction. The acceptance angle of the focused transducer is determined by the geometric focus (i.e., f-number) and should be used to limit the back-projection range. Taking acceptance angle into account, virtual-detector-based PAT reconstruction can potentially provide proper weightings to further reduce image artifacts and increase the uniformity of the image brightness. In addition, in our preliminary study, we also found that the f-number of the focused transducer will affect the reconstructed tangential and radial resolution. The f-number of the focused transducer should be optimized to obtain the best performance of the virtual-detector based PAT reconstruction.

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