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Coherence-Weighted Synthetic Focusing Applied to Photoacoustic Imaging Using a High-Frequency Annular-Array Transducer

Parag V. Chitnis¹, Orlando Aristizábal², Erwan Filoux¹, Ashwin Sampathkumar¹, Jonathan Mamou¹, and Jeffrey A. Ketterling¹

¹Riverside Research, F.L. Lizzi Center for Biomedical Engineering, New York, NY, USA

²Skirball Institute of Biomolecular Medicine, NYU School of Medicine, New York, NY, USA

Abstract

This paper presents an adaptive synthetic-focusing scheme that, when applied to photoacoustic (PA) data acquired using an annular array, improves focusing across a greater imaging depth and enhances spatial resolution. The imaging system was based on a 40-MHz, 5-element, annular-array transducer with a focal length of 12 mm and an 800- μ m diameter hole through its central element to facilitate coaxial delivery of 532-nm laser. The transducer was raster-scanned to facilitate 3D acquisition of co-registered ultrasound and PA image data. Three synthetic-focusing schemes were compared for obtaining PA A-lines for each scan location: delay-and-sum (DAS), DAS weighted with a coherence factor (DAS + CF), and DAS weighted with a sign-coherence factor (DAS + SCF). Bench-top experiments that used an 80- μ m hair were performed to assess the enhancement provided by the two coherence-based schemes. Both coherence-based schemes increased the signal-to-noise ratio by approximately 10 dB. When processed using the DAS-only scheme, the lateral dimension of the hair in a PA image with 20 dB dynamic range was between 300 μ m and 1 mm for imaging depth ranging from 8 to 20 mm. In comparison, the DAS + CF scheme resulted in a lateral dimension of 200 to 450 μ m over the same range. The DAS + SCF synthetic focusing further improved the smallest-resolvable dimension, which was between 150 and 400 μ m over the same range of imaging depth. When used on PA data obtained from a 12-day-old mouse embryo, the DAS + SCF processing improved visualization of neurovasculature.

Keywords

photoacoustic imaging; coherence weighting; beamforming; annular arrays

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Corresponding Author: Parag V. Chitnis, Department of Bioengineering, George Mason University, 4400 University Drive 1G5, Fairfax, VA 22032, USA. pchitnis@gmu.edu.

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Introduction

Photoacoustic (PA) imaging (PAI), which relies on local absorption of a brief (~ns) monochromatic light pulse and subsequent thermoelastic conversion of the optical energy into a broadband ultrasonic pulse,^{1,2} can depict vasculature when performed using an optical wavelength that is preferentially absorbed by blood. Similar to ultrasound (US), the spatial resolution achieved in PAI typically depends on the operational frequency and one-way receive properties of the ultrasonic transducer used to passively detect the PA signals. Also similar to US, the PA signals received by the transducer include signals originating from on- and off-axis targets. The off-axis signals distort the PA signals of interest and introduce undesirable clutter in the images.

When used for PAI, linear-array transducers are particularly prone to clutter due to their fixed elevation focusing. Signals from on-axis targets can be better isolated using single-element spherically focused transducers. However, these transducers have a relatively small depth of field (DOF). In regions beyond the DOF, image resolution deteriorates due to an expanding acoustic beam. An alternative to single-element and linear-array transducers is an annular-array transducer, which provides a radially symmetric beamwidth (similar to single-element transducers) as well as relatively uniform spatial resolution over the entire field of view (similar to linear arrays). We previously demonstrated that penetration and focusing can be improved along an increased imaging depth during US imaging by processing the pulse-echo radio-frequency (RF) data using a synthetic-focusing scheme involving delay-and-sum (DAS) at multiple depths along the acoustic axis.³⁻⁵ This work was performed at an operational frequency of 40 MHz to facilitate micron-scale resolution necessary to resolve anatomical features in mouse embryos.^{6,7} The 40-MHz annular array combined with the DAS beamformer also was recently used for PAI of mouse embryos.^{8,9} The DAS beamforming approach inherently de-emphasizes the signals from off-axis targets, which are not phase-matched to signals originating from on-axis targets.

DAS beamforming can be further improved by implementing a weighting function based on inter-element signal coherence. Coherence factor (CF) has been used for suppressing side and/or grating lobes in US imaging. CF, originally implemented for linear arrays, was defined as the ratio between the energy of the coherent sum and the total energy in the signals received across an aperture.¹⁰ CF is based on the fact that after focusing delays are applied to inter-element data, the signals on the main lobe are highly coherent and the signals from the side lobes are incoherent. This also implies that when the off-axis signals interfere strongly with the on-axis signals, the coherent energy of the signal is reduced in comparison with the total energy, resulting in a low CF value. When the output of the DAS beamformer is weighted using the CF, aberration effects from off-axis signals are suppressed.^{10,11} A generalized, spatial-frequency-domain version of CF was introduced by Li and Li for applying to objects with diffused targets.¹² The generalized CF includes low-frequency as well as the DC component of the spectrum while the original CF contained only the DC component. Camacho et al. introduced a sign-coherence factor (SCF) based on the dispersion of phases instead of amplitudes.¹³ SCF implementation is particularly useful for minimizing artifacts associated with spurious off-axis signals. The on-axis signals have an effective phase dispersion of zero (in phase), whereas the dispersion increases for the off-

axis signals and results in a low-weighting factor. The side and grating lobes are thus suppressed, and effective lateral resolution is improved.¹³

CF-based processing also has been implemented for improving image resolution during PAI. Several groups have demonstrated the utility of CF-weighted beamforming for reducing clutter due to off-axis signals when using linear arrays.^{14,15} Signal contamination from off-axis contributions is particularly troublesome in PAI applications involving a single, high-frequency, fixed-focus transducer that is mechanically scanned over the specimen. Synthetic-aperture focusing technique (SAFT) combined with CF-weighting can improve DOF and resolution.¹⁶ Deng et al. extended this approach to obtain vascular morphology from SAFT-derived 3D images and adaptively apply CF-weighted SAFT in a direction normal to the blood vessels for optimally enhancing the image resolution.¹⁷ CF-weighted SAFT also has been successfully applied to 2D planar arrays for 3D PAI.¹⁸

Despite the demonstration of its utility in improving US and PA image quality, CF-weighted beamforming has not been widely applied to annular arrays. Annular arrays also are sensitive to spurious PA signals originating from off-axis locations that result in higher side lobes and deterioration in signal-to-noise ratio (SNR) compared with US imaging performed using the same annular array.^{9,19} Passler et al. demonstrated the utility of DAS + CF beamforming for enhancing spatial resolution and DOF for PAI using a low-frequency annular array.²⁰ However, unlike other US-based studies, they computed the point-wise CF after envelope detection of the A-line data. In this study, we investigated the feasibility of improving the quality of PA images acquired from the 40-MHz annular array using adaptive DAS approaches based on inter-element signal coherence. Phantom-based studies were performed to quantitatively assess the improvement provided by DAS + CF and DAS + SCF schemes. Last, improvement in image quality provided by coherence-based strategies was demonstrated using image data of a mouse embryo at a gestational stage of 12 days.

Materials and Method

Annular Array

The annular array used in this study was fabricated through a process that was previously described in Ketterling et al.²¹ Briefly, the transducer was fabricated by bonding a 9- μm poly (vinylidene fluoride-tetrafluoroethylene) film to an array pattern etched onto a copper-clad polyimide film using a non-conductive epoxy. The array had five equal-area rings with 100- μm spacing between rings and a total aperture of 6 mm. The films were pressed against a stainless steel sphere to produce a radius of curvature of 12 mm. A 800- μm diameter hole was precisely drilled along the central axis of the array to accommodate coaxial laser illumination.⁹ The central hole reduced the surface area of the central element by 10%, which reduced the sensitivity by 2 dB, and did not alter the transducer-beam characteristics appreciably.⁹ The array had a central frequency of 38 MHz and a -6 dB pulse-echo bandwidth of 40%, which were characterized using pulse-echo measurements from a rigid quartz reflector while using a broadband pulser (Panametrics 5900, Panametrics, Waltham, MA) for excitation.²¹ The center frequency and bandwidth of the array are expected to be nominally greater when operating in PA mode.

The output beam of the excitation laser, which had a nominal beamwidth of 2 mm, was directed toward a periscopic assembly of collimation lenses and mirrors on kinematic mounts (Figure 1a) to reduce the beam size and precisely guide the laser beam through the center hole of the annular array.⁹ The periscopic assembly consisted of two portions: one that moved with the annular array in 2D (shown in dashed box) and one that moved along one of the scan axes (shown in solid box). This arrangement facilitated delivery of the laser beam through the center hole during mechanical scanning of the annular array. Using collimation lenses with focal distances of 100 and 25.4 mm, the beam was collimated and reduced to a spot size of 500 μm . The annular array was mounted on a manual 2D stage that allowed fine alignment of the array relative to the laser beam to ensure coaxial illumination of the specimen (Figure 1b).

Data-Acquisition System

The data-acquisition system consisted of three components: motion system, pulser and laser excitation, and digital-signal acquisition. The motion system was driven via a motion-control card (PXI-7534, National Instruments, Austin, TX) and consisted of a high-speed linear actuator (LAL35, SMAC, Carlsbad, CA) and a cross-axis stage (Newport, Irvine, CA), both with 1- μm precision.

A broadband pulser (Avtech AVB2-TA-C-CVA, Ottawa, Ontario, Canada) was used to excite the central element of the annular array with a 200-V_{pp} monocycle impulse. A light pulse from a frequency-doubled Nd:YAG laser (MiniliteII, Continuum, Santa Clara, CA) operating at a wavelength of 532 nm was used to excite the PA response of the tissue. The laser pulse was 5 ns in duration with a repetition rate of 10 Hz. To acquire simultaneous US and PA data, the pulser for generating US signals was synchronized to the Q-switch of the laser. A custom LabVIEW (National Instruments, Austin, TX) program controlled and automated mechanical scanning of the annular array and data collection. The RF line from each scan location contained the US and the PA data. Because the separation distance between the annular array and the proximal surface of the uterus was greater than the dimension of the embryo along the acoustic axis, the US and PA signals corresponding to the embryo were inherently separated in time, and simple time gating was used to parse out the US and PA data from the RF lines. Each receive channel had a low-noise amplifier (AU-1313, Miteq, Hauppauge, NY) with a fixed 46 dB gain. Three 2-channel digitizer cards (PXI-5152, National Instruments, Austin, TX) with an 8-bit resolution and a 500-MHz sampling rate were used to simultaneously record the five channels of RF data.

Coherence-Weighted DAS Processing

The multi-channel RF data were processed using a DAS algorithm,³ which shifted the focus of the transducer along the axial direction through post-processing of the transmit-to-receive (TR) data acquired from all five channels. To focus the array on receive at a depth d , the time delay t_i required for ring n of the array is approximately $t_i = [a_i^2(1/R - 1/d)]/2c$, where R is the geometric focus, c is the speed of sound, and a_i is the root mean square of the inner and outer radii of ring n .²² For US imaging, the total round-trip delay $t_i^{tot} = t_1^T + t_i^R$, with t_1^T on transmit and t_i^R on receive, was calculated for each of the five elements and applied to the acquired A-lines. The A-line data were then envelope-detected and log-compressed to

produce B-mode slices. In the case of PAI, excitation is considered instantaneous and the delays corresponded to the one-way acoustic propagation ($t_i^T=0$). In each case, the signals were dynamically delayed and summed to synthetically focus at many different depths d . The focused data windowed around each d were assembled to form a composite image with multiple focal zones.

CF-based processing was implemented on the 5-channel PA data for each scan location after performing DAS. An interpretation of CF is the ratio of coherent energy to total energy for the time-delayed, pre-summed RF data acquired using $N=5$ channels in the aperture. A point-wise CF (as a function of acoustic time t) was computed as

$$CF(t) = \frac{\left| \sum_{i=1}^N S_i(t - \Delta t_i) \right|^2}{N \sum_{i=1}^N |S_i(t - \Delta t_i)|^2}, \quad (1)$$

where $S_{i=1:N}(t - t_i)$ are the RF signals acquired using N channels and delayed by $\Delta t_i = t_i^{tot}$.^{12,23} In this manner, a coherence matrix that contains CF values corresponding to each data point in each of the DAS-processed A-lines was obtained, which represented a 2D spatial map (corresponding to the B-mode space) of inter-element signal coherence. The DAS-processed 2D RF data were then multiplied by the CF map, envelope-detected, and then log-compressed to produce a CF-weighted, synthetically focused B-mode image.

The point-wise SCF was computed in a manner similar to CF using the following expression:

$$SCF(t) = |1 - \sigma|^p, \quad \sigma = \sqrt{1 - \left(\frac{1}{N} \sum_{i=1}^N B_i(t) \right)^2}, \quad (2)$$

where p adjusts the sensitivity of the SCF ($p=1$ in this work). $B_i(t)$ is the point-wise sign bit of the time-delayed signal $S_i(t - t_i)$ such that $B(t) = 1$ when $S(t) \geq 0$ and $B(t) = -1$ when $S(t) < 0$.

Phantom Studies

Phantom studies were conducted to test the performance of the annular array while comparing three synthetic-focusing strategies: DAS, CF-weighted DAS (DAS + CF), and SCF-weighted DAS (DAS + SCF).

The PA performance of the annular array in depicting micro-scaled structures and the resolution enhancement provided by the coherence-based schemes were investigated *in vitro* using an 80- μm diameter hair suspended normal to the acoustic axis between two posts in a water bath. The 532 nm laser was operated at an output energy setting of 4 mJ. The

periscopic assembly was scanned cross-sectionally to the hair. The transducer position was varied in the axial direction from 8 to 20 mm relative to the hair and PA data for all the locations were combined to produce a composite data set containing the target at all axial positions. The cross-sectional dimension of the hair inferred from the PA image thresholded at the 20 dB dynamic range was used to quantify depth-dependent -20 dB beamwidth, which provided an effective PAI resolution when depicting structures similar to blood vessels in mouse embryos.

Embryonic Imaging

All animals used in these studies were maintained under protocols approved by the Institutional Animal Care and Use Committee at New York University School of Medicine. The mouse was anesthetized with 1.5% isoflurane and the intact uterus from a mouse embryo at the 12th day of gestation (E12.5) was externalized through a small slit in the rubber membrane into a fluid-filled Petri dish (Figure 1d). Laser-pulse energy incident at the uterine wall was 4 mJ/pulse. The transducer assembly was raster-scanned in 50- μ m increments along both scan axes to acquire 201 B-mode images resulting in a 3D-image volume of $7 \times 7 \times 10$ mm³. The exposed uterus is relatively immobilized except for the vertical motion resulting from the mother's respiration. The mother's respiratory cycle was monitored in real-time and respiratory gating was used to acquire RF data only during the resting phase of the respiratory cycle.²⁴ One RF line per channel was acquired for each laser pulse and no averaging was performed. A single image plane was obtained in approximately 15 s limited only by the 10 Hz pulse-repetition frequency of the laser.

The RF data were imported into custom-analysis software (MATLAB, Mathworks Inc., Natick, MA) that extracted the PA and US data from each RF line and applied the three beam-forming strategies for comparison. The magnitude of the PA signals in the synthetically focused A-lines, which was proportional to incident optical energy, was corrected to compensate for depth-dependent losses in optical energy. The correction factor was determined by measuring the optical-extinction coefficient of a surgically exposed uterus, which had a value of 0.35 mm⁻¹. This was measured with a fiber-based spectrophotometer in a trans-illumination configuration (not shown). B-mode US and PA images were generated directly from the envelope-detected, log-compressed A-lines. The US images were thresholded to a dynamic range of 50 dB, and the PA images obtained by DAS, DAS + CF, and DAS + SCF approaches were thresholded to the same dynamic range of 30 dB.

The PA images were processed with a local-means filter that had a kernel dimension of 100 μ m (approximately equal to the 6 dB lateral resolution) using ImageJ, which is a public-domain, image-processing software.²⁵ A mean projection image (MIP) of image slices contained within an approximately 500- μ m thick volume centered on the mid-sagittal section was obtained using a 3D-rendering software (Amira, FEI Visualization Sciences Group, Burlington, MA). These images were used for comparing the quality of *in vivo* vascular images obtained using the three synthetic-focusing schemes. The PA images were overlaid on the US images for providing anatomical context to the vascular features.

Results

PAI of Hair Phantom

Figure 2 shows the composite PA image of the hair located at different depths relative to the transducer face. A pulse energy of 0.4 mJ was found to be sufficient to obtain an SNR above 30 dB at the geometric focus. The PA-signal amplitude peaked at 10 mm and then decreased with increasing depth (data not shown here, but previously presented in Filoux et al.⁹). Depth-dependent reduction in transducer sensitivity and loss due to acoustic attenuation contribute to this depth-dependent signal decay. (Note that this PA-signal loss with increased depth is further compounded by loss in optical energy when imaging tissue.)

When processed using the DAS + CF strategy (Figure 2b), the side lobes and the noise floor in the PA-hair data were considerably reduced compared with the DAS-only scheme (Figure 2a). This reduction in noise and side lobes translated to an improvement in spatial resolution in depicting the hair in the PA image (shown with a dynamic range of 20 dB), which was observed consistently throughout the entire imaging depth. The PA image of the hair obtained using the DAS + SCF scheme was comparable with that obtained using DAS + CF. Representative lateral beam profiles obtained from the hair scans at 12 mm (geometric focus) and 20 mm are shown in Figures 3(a) and (b), respectively. Lateral beam profiles indicated that the DAS + CF scheme lowered the noise floor by approximately 10 dB. The DAS + CF approach significantly improved the 20 dB lateral beamwidth. The DAS + SCF focusing was marginally better than the DAS + CF scheme.

The lateral beamwidth obtained using all three synthetic-focusing approaches remained less than 500 μm (laser-beam diameter) across the entire imaging depth (data not shown), which indicated that the PAI lateral resolution was solely dependent on the focusing properties of the transducer. (This may change if the optical beam diameter is reduced to a dimension smaller than the acoustic beamwidth.) The width of the cross-sectional profile of the hair depicted in the PA image at each depth was used to obtain the values corresponding to -6 dB and -20 dB lateral resolution obtained from DAS, DAS + CF, and DAS + SCF strategies. The -6 dB resolution achieved by the DAS + CF and DAS + SCF schemes was only modestly (20% on average) better than that achieved by the DAS-only scheme (data not shown). The -20 dB resolution for the three synthetic-focusing schemes is plotted as a function of depth in Figure 4. When processed using the DAS-only scheme, the lateral dimension of the hair in a PA image with 20 dB dynamic range was between 300 μm and 1 mm for imaging depths ranging from 8 to 20 mm. In comparison, the DAS + CF scheme resulted in a lateral dimension of 200 to 450 μm , and the DAS + SCF synthetic focusing produced a lateral dimension of 150 to 400 μm . These measurements indicated that coherence-based schemes produced a significantly better (90% on average) apparent lateral resolution in the -20 dB thresholded hair image when compared with DAS-only strategy.

PAI of a Mouse Embryo

A representative A-line measurement acquired by the central element is shown in Figure 5. The PA and US signals are well separated in time because the PA waves undergo one-way propagation instead of the round-trip pulse-echo of US waves. The PA signals can have a

larger peak amplitude in comparison with the US signals, but the PA-signal amplitude decays rapidly with optical-penetration depth.

PA images of the mid-sagittal plane (color) of a 12-day-old mouse embryo overlaid on the corresponding US image (gray) are shown in Figure 6. All three PA images are displayed using the same dynamic range of 30 dB. The DAS-only PA image does show prominent blood vessels, but the image is contaminated with signal clutter likely as a result of off-axis PA signals detected by the annular array. A qualitative comparison between the images obtained using DAS-only and the coherence-weighted schemes indicate that the DAS + CF and DAS + SCF beamforming approaches significantly reduce the artifacts arising from the spurious off-axis signals, resulting in a sharper depiction of major blood vessels. The vessels in the DAS + SCF image are better delineated in comparison with the DAS + CF image, which is consistent with the depth-dependent lateral beam profiles shown in Figure 4. An apparent variation in signal intensity for different scan lines (visible in all three PA images) and corresponding striations in the horizontal direction was due to the presence of air bubbles. Occasional fluctuations in the laser-pulse energy to a value greater than 4 mJ resulted in production of air bubbles in the laser-beam path, which resulted in a transient reduction in incident optical fluence. Prominent vascular features visible in the DAS + SCF image were validated against a reference magnetic-resonance image (Figure 6d).

Discussion

We investigated the utility of adaptive beamforming techniques that rely on coherence-weighted DAS for reducing the contributions of spurious off-axis PA signals that deteriorate SNR and resolution during PAI using a 40-MHz annular array. In our configuration, the beam characteristics primarily depend on the transducer geometry and operational frequency. The relatively similar -6 dB lateral beamwidths achieved with DAS-only, DAS + CF, and DAS + SCF schemes are consistent with this notion. When performing PA imaging in tissue, aggressively thresholding the signal level, for example, at 6 dB, is highly impractical because of diminished optical penetration; the off-axis PA signals are particularly confounding when depicting PA image data at more practical thresholds, for example, at 20 dB. The improvement in resolution achieved using the DAS + CF scheme can be better appreciated by examining the lateral dimension of the hair in the 20 dB thresholded images acquired using the DAS-only and DAS + CF schemes (Figure 4). The dimension of the hair in this image shows the point-spread function of the hair target, which increases with increase in depth. This is because the synthetic focusing with the annular array has the effect of creating a variable F-number such that the lateral beamwidth of the US transducer increases as a function of depth.

Unlike the method implemented by Passler et al.²⁰ for processing annular-array PA data, we computed the point-wise CF and SCF from the raw inter-element RF data and not the envelope-detected signals. CF based on the inter-element envelope-detected signal did not produce a weighting function that substantially improved the resolution or the image quality of our 40-MHz annular array (data not shown). This can be attributed to the fact that the process of envelope detection, which is equivalent to demodulation, reduces the precision in the computation of coherence between inter-element RF data.

When applied to embryonic PA data, CF processing improved delineation of vasculature and decreased artifacts or erroneous depiction of vascularity in regions that were void of blood. When applied to US imaging, the use of coherence-weighted DAS is known to locally increase variance in the image and sometimes result in artifacts due to removal of speckle from the image. However, DAS + CF and DAS + SCF schemes did not adversely affect the PA images. This could in part be due to the use of the local-means filter applied to the 3D-image volume. The use of coherence-based approaches does result in a modest reduction in the absolute signal amplitude, which translates to a reduction in brightness of the depicted vasculature and might pose constraints in terms of maximum attainable imaging depth.

Currently, our dual-modality system is limited only by the 10 Hz pulse-repetition rate (PRF) of our laser. The slow PRF also was the reason why only the central element of the array was used for the US pulse-echo technique. However, US images acquired by transmitting using the central element and receiving on all five channels produced images with quality that was adequate for resolving anatomical features. A faster laser (3–5 kHz PRF) would facilitate real-time, dual-modality imaging and acquisition of full 3D-image data (while using all five elements in pulse-echo configuration) in under three min (currently possible when acquiring just US data). Because the calculation of CF can be performed on the 2D image data using matrix operators with modest computational expense, CF-weighted DAS does not add a significant computational burden to the DAS-only scheme, which allows the possibility of real-time processing and display of annular-array image data.

The *in vivo* result was obtained by imaging a live embryo in a surgically externalized, intact uterus. This was necessary because when the annular array and the DAS-only scheme were used for PAI through the intact abdominal wall, the embryonic vasculature was not well visualized due to inadequate SNR. Future works will continue the effort to improve the quality of PA images by using longer optical wavelengths to improve optical penetration and by using a laser with a faster repetition rate to facilitate signal averaging and faster frame rates. SNR enhancement, achieved using these proposed steps in conjunction with improvement in the PA spatial resolution afforded by the DAS + SCF scheme, might allow us to image embryonic vasculature in a completely non-invasive manner and facilitate longitudinal studies of embryonic development.

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Figure 1.

Periscopic integration of coaxial illumination. (a) Computer-aided drawing of the periscopic assembly for precisely delivering the laser beam through the center hole of the array. (b) Close-up of transducer-lens assembly. (c) Half-section of the transducer-lens assembly. (d) Photograph of array-lens assembly immersed in a phosphate buffered saline bath and scanned over the exposed uterus of an anesthetized mouse.

Figure 2.

Composite B-mode PA image of a hair phantom (80 μm diameter) in water obtained using (a) DAS, (b) DAS + CF, and (c) DAS + SCF. The images are shown in 20 dB dynamic range. PA = photoacoustic; DAS = delay-and-sum; DAS + CF = DAS weighted with an inter-channel coherence factor; DAS + SCF = DAS weighted with an inter-channel sign-coherence factor.

Figure 3.

Normalized cross-sectional PA profiles of hair located at a depth of (a) 12 mm and (b) 20 mm. PA = photoacoustic; DAS = delay-and-sum; DAS + CF = DAS weighted with an inter-channel coherence factor; DAS + SCF = DAS weighted with an inter-channel sign-coherence factor.

Figure 4.

Lateral resolution obtained from the composite PA image of the hair thresholded at -20 dB relative to the peak value. PA = photoacoustic; DAS = delay-and-sum; DAS + CF = DAS weighted with an inter-channel coherence factor; DAS + SCF = DAS weighted with an inter-channel sign-coherence factor.

Figure 5.

(a) A representative A-line measurement from the central element. PA and US signals are separated in time. Zoomed-in PA and US signals are shown in (b) and (c), respectively. PA = photoacoustic; US = ultrasound.

Figure 6.

MIPs of mid-sagittal plane of a 12-day-old mouse embryo obtained by processing the PA data using (a) DAS, (b) DAS + CF, and (c) DAS + SCF synthetic focusing. All images were produced using the same intensity scale, which ranged from 0 to 30 dB. The scale bar represents 1 mm. (d) A magnetic-resonance image (modified from <http://www-personal.umich.edu/~brdsmith/Research/Cardiovascular/figure2.html>) is shown for comparison. MIP = mean projection image; PA = photoacoustic; DAS = delay-and-sum; DAS + CF = DAS weighted with an inter-channel coherence factor; DAS + SCF = DAS weighted with an inter-channel sign-coherence factor.