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Research article

Rapid scanning wide-field clutter elimination in epi-optoacoustic imaging using comb LOVIT

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ABSTRACT

Epi-style optoacoustic (OA) imaging provides flexibility by integrating the irradiation optics and ultrasound receiver, yet clutter generated by optical absorption near the probe obscures deep OA sources. Localised vibration tagging (LOVIT) retrieves OA signal from images that are acquired with and without a preceding ultrasonic pushing beam: Radiation force leads to a phase shift of signals coming from the focal area resulting in their visibility in a difference image, whereas clutter from outside the pushing beam is eliminated. Disadvantages of a single-focus approach are residual clutter from inside the pushing beam above the focus, and time-intensive scanning of the focus to retrieve a large field-of-view. To speed up acquisition, we propose to create multiple foci in parallel, forming comb-shaped ARF patterns. By subtracting OA images obtained with interleaved combs, this technique moreover results in greatly improved clutter reduction in phantoms mimicking optical, acoustic and elastic properties of breast tissue.

1. Introduction

Optoacoustic (OA) or equivalently named photoacoustic (PA) imaging shows optical absorption contrast inside tissue with ultrasound spatial resolution, based on detection of thermoelastic ultrasound that is generated upon optical absorption when irradiating the tissue with pulsed laser light. An image can be reconstructed by time-resolved detection of these ultrasound signals. The basic mechanisms of OA imaging and underlying signal processing are thoroughly described in review articles [1–3].

OA imaging provides functional information via the display of the blood oxygen saturation level based on the difference in absorption spectra of oxy- and deoxyhemoglobin [4–7]. Thus, it is a promising tool for diagnosis of vascular diseases and cancer [8,9], as well as for monitoring treatment response [10–12]. The combination of classical ultrasound (US) and OA imaging in one device has been demonstrated as a safe, real-time, multimodal imaging modality [13–17] with promise for clinical diagnosis [16]. For this combination, an epi-style setup is favored such that irradiation and detection is performed from the same side of the tissue. The optical components can then directly be integrated into the acoustic probe, e.g. as a diode-based miniaturized multi-wavelength laser source [18]. Such a combined setup allows flexible single handed probe guidance and accessibility to body parts otherwise not reachable in transmission mode due to bones, acoustically attenuating tissues or gas-containing hollow organs [19].

On the downside, the epi-style setup generates clutter signals which limit signal-to-background ratio and decrease the imaging depth to values substantially less than the several centimeters predicted by the electronic noise level, detector bandwidth and acoustic attenuation in the tissue [20]. Clutter is caused by strong OA transients that are generated by optical absorption at the tissue irradiation site in proximity to the detecting transducer [21], e.g. by melanin, blood capillaries or vascular lesions [22,23]. Detection of transients that propagate straight to the transducer from outside the imaging plane ('direct clutter') or detection of echoes when transients propagate into the tissue and are scattered by acoustic impedance fluctuations ('echo clutter' or 'reflection artefacts') generates artefactual background signals. These background signals can overlap in time and may then be confused with weaker optoacoustic signals from light absorbing structures deep inside the tissue strongly reducing the signal to clutter ratio. For particular aspects of anatomy or function, the signal to clutter ratio can be enhanced by employing OA contrast agents (e.g. [24–28]). This approach, however, increases the imaging depth for only those

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aspects of anatomy or function to which the agent localizes, and it is invasive.

To generally increase imaging depth towards the noise limited value, various clutter reduction techniques have been proposed:

(1) Displacement compensated averaging (DCA) employs the clutter decorrelation that results from quasi-static tissue deformation when palpating the tissue with freehand probe motion [29–31]. This technique is inherently limited by the maximum achievable deformation, typically resulting in a maximum threefold clutter reduction factor [17,29].
(2) Photoacoustic-guided focused ultrasound (PAFUSion) uses ultrasound pulse-echo acquisitions to mimic OA signal reflection artefacts, which can then be subtracted from the OA image for clutter reduction [32–35]. When using linear array probes for detection, this method can only mimic echo clutter that is generated by OA transients that propagate entirely within the imaging plane. Thus PAFUSion is well suited in a situation where the tissue must be irradiated directly below the probe aperture, to remove the significant echo clutter that stems from the plane-like OA transients generated by the skin melanin layer.
(3) Spatial coherence weighted OA imaging [36,37] exploits the fact that clutter frequently arises from OA sources, or echo-producing structures, that cover a volume that is much larger than the focal volume associated with a given image pixel, and thus the transients that they produce have low spatial coherence when they reach the detector. This allows their removal by a channel-level spatial coherence filter, but not without some loss of information, particularly from distributed but genuine OA sources.
(4) With the goal of allowing efficient clutter elimination that is independent of the clutter or signal origin, localized vibration tagging (LOVIT) was developed [38]. A long-pulsed (few 100 microseconds) focused ultrasonic beam generates acoustic radiation force (ARF) that induces localized tissue displacement at its focus (henceforth called single-focus LOVIT). OA images are acquired, with and without localized displacement of optically absorbing structures produced by preceding ARF pushes. Subtraction of the images results in a LOVIT image that highlights true OA signal at the focal region where the displacement is largest. Clutter signals that occur in the same region of the reconstructed image originate – by definition – from outside the focus. As the displacement outside the focal region is comparably small or even zero, clutter signals are reduced in the subtracted images. In a proof-of-principle study, efficient clutter reduction was demonstrated using a separate US transducer for ARF generation [38]. Recently, the performance of single-focus LOVIT was successfully demonstrated in a clinically realistic setup where the same linear array probe was used for ARF generation and imaging [39].

Combinations of the above approaches may eventually prove to be beneficial. For the moment however, LOVIT appears to be particularly promising and presents considerable opportunities for further development.

In single-focus LOVIT, clutter reduction is achieved only at one focal region at a time. Such an approach has the disadvantage of long acquisition times for scanning the focus position through an image, or a region of interest (ROI) within an image, which may limit clinical applicability in terms of real-time feedback. In addition, single-focus LOVIT does not completely eliminate echo clutter. It leaves residual echo clutter (henceforth, residual clutter): even though the displacement is largest inside the focal region of the ARF beam, significant displacement is generated also above the focal region and scatterers located there produce residual clutter inside the focal region if the total acoustic path length from a strong OA source to a scatterer and then to the detector equals that from the focal depth to the detector.

To circumvent these two problems, we propose a novel modification of LOVIT where multiple horizontally aligned foci are created simultaneously, forming comb-shaped ARF patterns. With this approach, imaging a large ROI can be accelerated compared to single-focus LOVIT, by a factor equal to the number of foci created within the comb. In addition to allowing a faster scanning time, comb LOVIT demonstrates a substantial reduction of residual clutter. To show this improvement, we compare the comb LOVIT and single-focus LOVIT approaches in a phantom study. Since detection of breast cancer is one of the promising application areas of OA imaging [40,41], we have chosen to compare the methods using a phantom that mimics optical, acoustic and elastic properties of breast tissue and contains optically absorbing inclusions mimicking blood vessels to allow signal to clutter ratios to be studied at various depths.

2. Theory

The acoustic radiation force (ARF) $f\left[N{\cdot}cm^{-3}\right]$ generated inside attenuating media by a focused ultrasonic beam can be calculated by [42]:

$$f(r,t) = \frac{2\alpha I(r,t)}{c}, \tag{1}$$

where $\alpha\left[cm^{-1}\right]$ is the ultrasound amplitude attenuation coefficient, $I\left[W{\cdot}cm^{-2}\right]$ the local intensity of the ultrasonic beam and c the speed of sound. ARF accelerates the tissue along the ARF beam axis, which results in localized axial tissue displacement relative to the state that existed at the time before ARF beam transmission, turning into shear waves that propagate mainly in the lateral and elevational directions perpendicular to the beam axis. Shear wave speed is related to the Young's modulus E of tissue in which the elastic modulus is orders of magnitude higher than the shear modulus via [43]:

$$c_s = \sqrt{\frac{E}{3\rho}}, \tag{2}$$

where ρ is the density of the tissue. When irradiating with a limited ARF push duration of few hundred microseconds, and with shear wave speeds typically a few m/s in soft tissues, shear waves can propagate only few millimetres before the push ends. As a result, the axial tissue displacement after the end of the push shows a spatial profile as indicated in Fig. 1a. At the ARF beam focus, it is characterized by a narrow region of maximum displacement (henceforth, 'focal region'), which is broadened relative to the ARF focus diameter due to shear wave propagation but still contained within few millimetres diameter. A significant displacement is also found above and below the focal region, albeit rapidly decreasing along the intensity profile of the pushing beam.

The goal of LOVIT is to differentiate between clutter and true OA signals and ultimately, eliminate clutter. In single-focus LOVIT, a first OA image is acquired without preceding ARF push (no displacement is present at the time of OA acquisition), and a second one at a short delay after an ARF push (a sketch of the scan protocol is shown in Fig. 1b). The subtraction of the two images results in a LOVIT image. The axial displacement of optical absorbers by the ARF beam leads to an OA signal phase shift between the two acquisitions and thus a non-vanishing signal from these absorbers in the subtraction (LOVIT) image. The signal amplitude in the LOVIT image is thus determined by the ARF-induced displacement that occurs in the interval between the two OA acquisitions. For displacements that are substantially below