On the Complexity of Photoacoustic Tomography: A Trade-off Between Image Quality and Computational Cost

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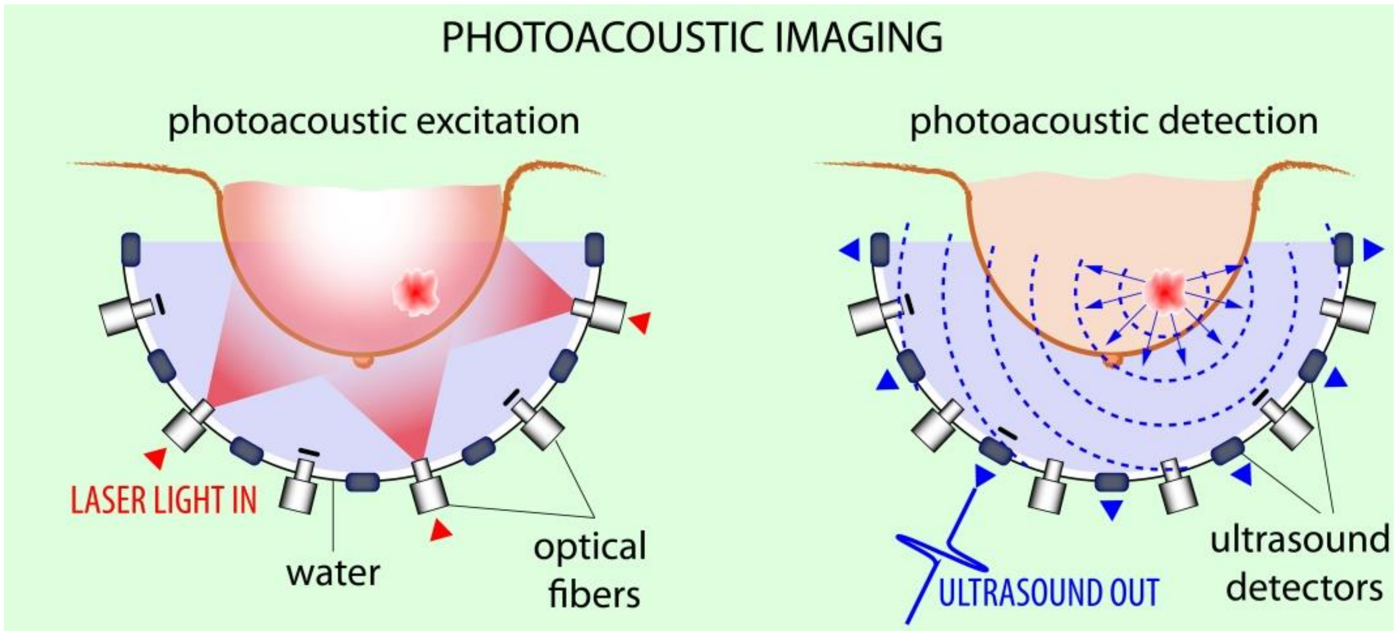
Overview

The photoacoustic tomography (PAT) is based on the fact that the chromophones in specific tissue, such as hemoglobin in veins and tumors, absorb the laser light. The absorbed light is transformed into heat and causes the thermoelastic expansion. This generates broadband ultrasonic waves in the tissue that can be captured by the ultrasound sensors.

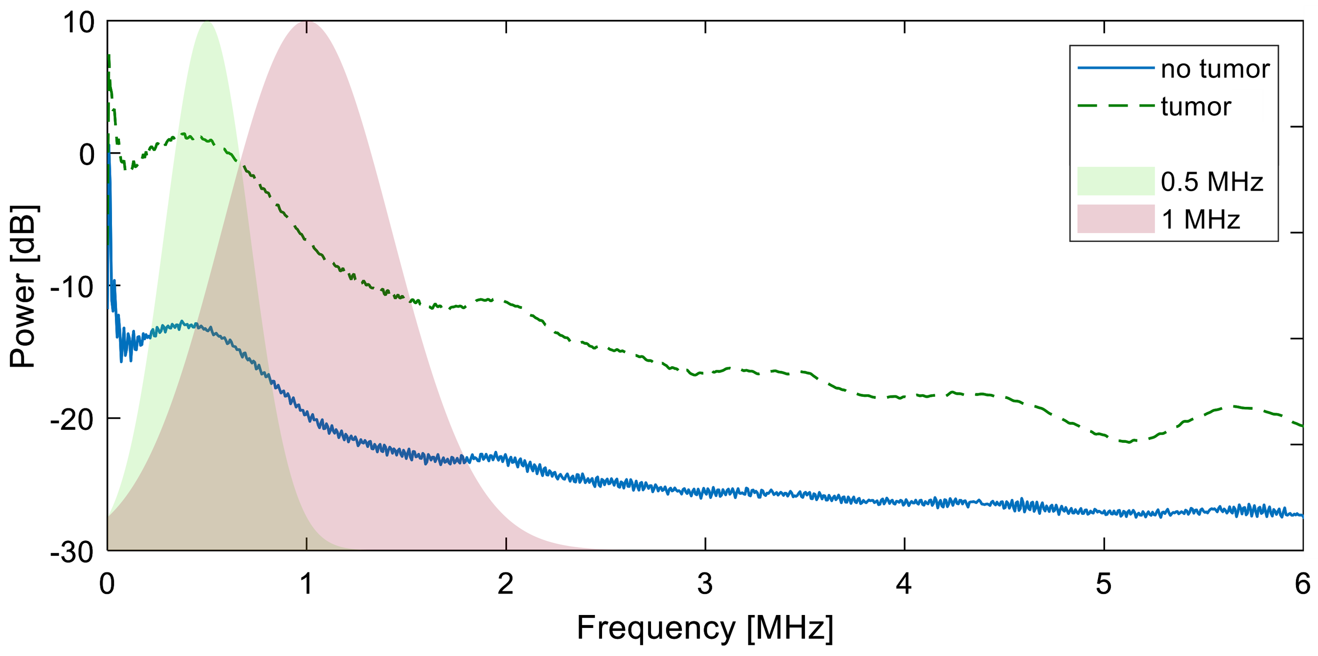


The goal of the PAT is to create a quantitative image showing the amount of energy created by absorbed light as origin of detected ultrasound waves. Several approaches are used to create the image, from which the model–based approach simulating the ultrasound propagation is considered as the least restrictive. The simulation is computationally demanding. Here, we examine an approach of using a recorded signal as a source signal with the reversed time axis.

Data Generation

Our reconstruction study uses a sensor bowl with a radius of 10 cm covered with broadband ultrasound sensors with a central frequency of 0.5 MHz. The simulation domain represents cuboid with dimensions of 20 cm × 20 cm × 13 cm.

We generate the detected signal from the MRI scan of the breast using the k-Wave toolbox with a fine grid supporting frequencies up to 2.4 MHz and grid spacing 0.2 mm. The recorded signal contains only a small portion of frequencies generated by the photoacoustic effect. Bandwidth of real signal, used sensors (0.5 MHz) and used simulation (central frequency of 1 MHz) is demonstrated on the picture below.



The generated signal is then filtered by the sensor frequency response and down-sampled to match the reconstruction grids with spacing 0.2 mm, 0.4 mm and 0.8 mm supporting frequencies up to 2.4 MHz, 1.2 MHz and 0.6 MHz respectively.

Reconstruction

The k-Wave toolbox is also used for the reconstruction. The parallel solver using one MPI process per socket was used on the Salomon cluster. The memory required for the biggest used matrix containing the source signal is displayed in the following table.

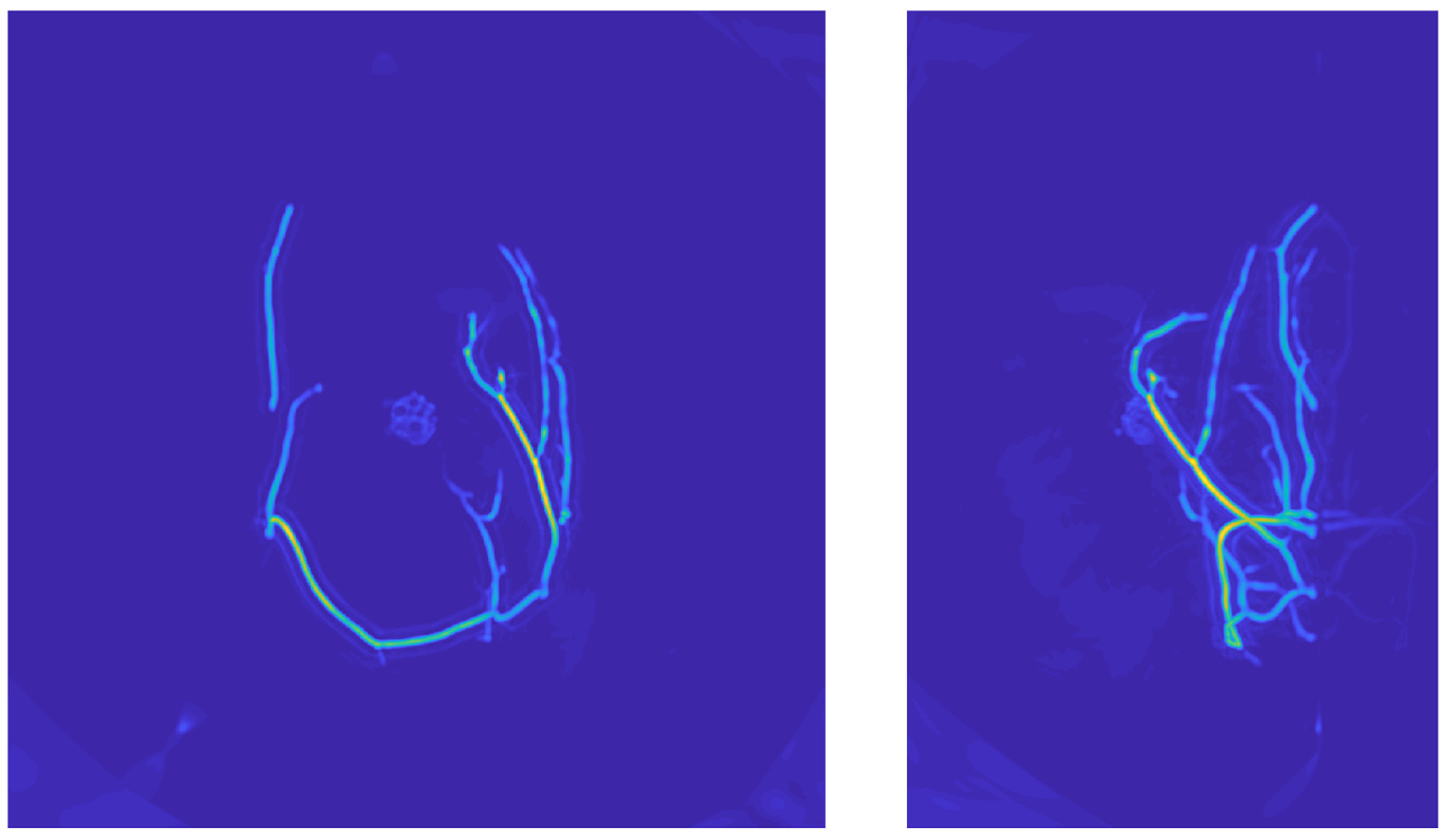
d_x	size	req. MPI	used MPI	nodes
0.2 mm	36.6 GB	19	32	16
0.4 mm	7.6 GB	4	16	8
0.8 mm	1.1 GB	1	8	4

Since MPI I/O is only able to manage up to 2 GB per process, the size of the biggest matrix dictates the minimal required number of MPI processes and the Salomon nodes.

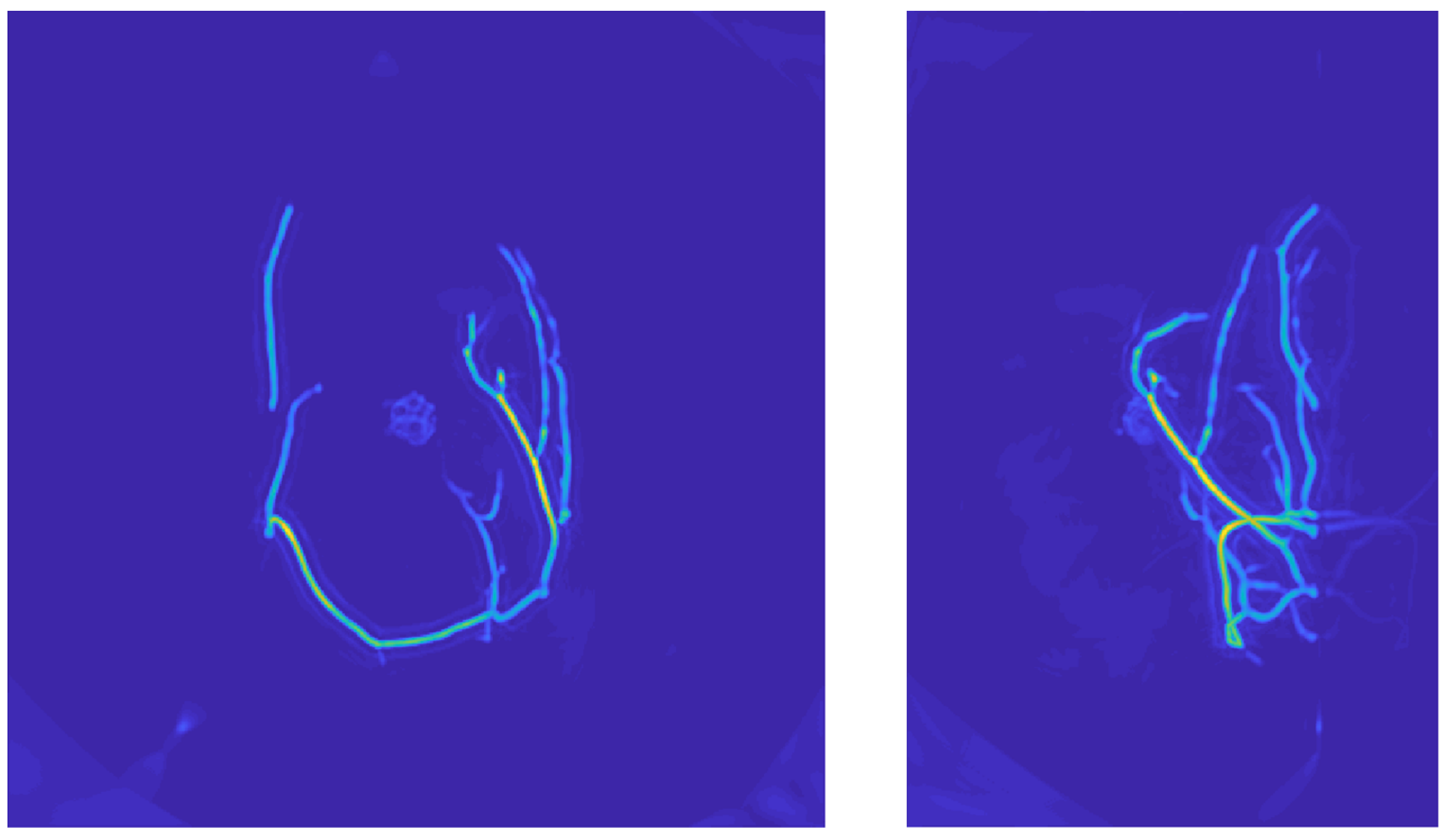
Results

The figure below shows a frontal (left) and lateral (right) view on the reconstructed image of breast phantom with noticeable vein structure.

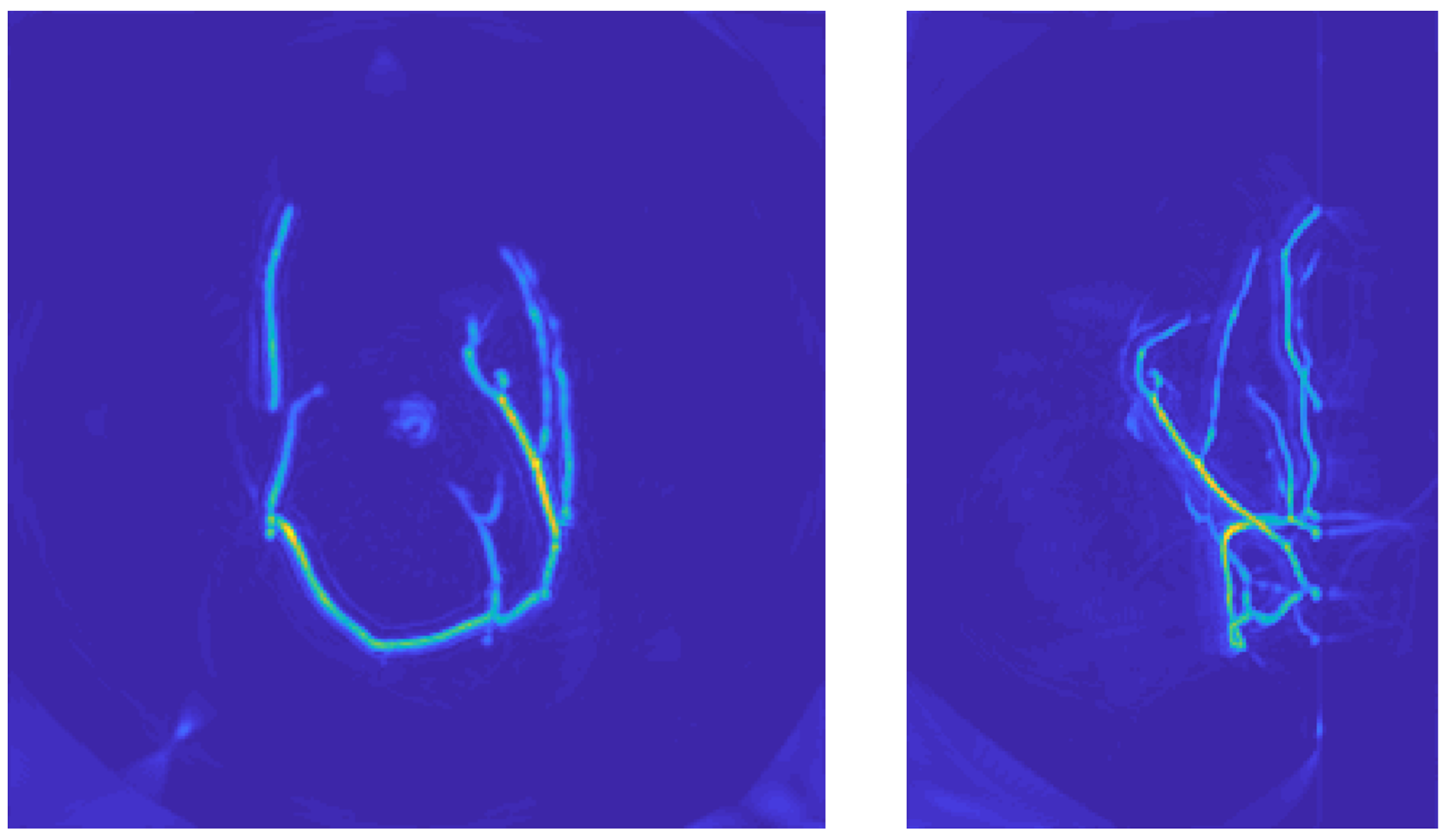
- $d_x = 0.2$ mm, $N_{x,y} = 1024$, $N_z = 672$, $N_t = 5220$
  nodes $= 16$, $t_{sim} = 4$ h 34 min



- $d_x = 0.4$ mm, $N_{x,y} = 528$, $N_z = 350$, $N_t = 2616$
  nodes $= 8$, $t_{sim} = 1$ h 22 min



- $d_x = 0.8$ mm, $N_{x,y} = 280$, $N_z = 192$, $N_t = 1320$
  nodes $= 8$, $t_{sim} = 9$ min



Conclusion

The worse result is provided by the last reconstruction which supported frequencies only up to 0.6 MHz. This was expected since the used ultrasound detectors with the central frequency of 0.5 MHz were used. The other reconstructions provided similar results and the difference in the image quality is almost negligible. The 0.2 mm reconstruction used **twice as many nodes** and ran **3.3 times longer**. The 0.4 mm reconstruction seems to be sufficient for 0.5 MHz sensors.

Even though the used signal was noise free, all reconstructed images contain artifacts, shadows, and vein-doubling, which are not desirable in medical imaging. A more robust iterative approach will have to be used to minimize the amount of these artifacts. Such an approach would require **up to 50 times more simulations**.

In the future, the iterative approach will be explored with a desire to use accelerators, more effective decomposition and adaptive resolution to reduce the cost of reconstruction.

k-Wave
A MATLAB toolbox for the time-domain simulation of acoustic wave fields

IT4Innovations national supercomputing center

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