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Analysis of Image Reconstruction Methods Employed in Photoacoustic Tomography using MATLAB with Different RAM Size in Computer

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Abstract: In this paper we have analyzed different image reconstruction methods employed in photoacoustic tomography. Methods analyzed are back projection method, pseudoinverse computation method and block coordinate descent method. All these methods are analyzed in MATLAB simulation tool for different sizes of RAM in computer. A blood vessel image is considered for the work. Our results clearly show the effect of RAM size on reconstruction of the image.
Keywords: Photoacoustic Tomography, K-wave, Transducer, Laser, Ultrasonic detection

I. INTRODUCTION

Photoacoustic effect was initially introduced in 1880 by Alexander Graham Bell. In 1938 photoacoustic effect started to be utilized for the purposes of analysis of gas. In 1960s the laser was invented. In 1970s – 1980s sensing applications started to evolve. Mid 1990s duration brought the starting of exploration of biomedical imaging. During early works progress was slow till the mid 2000s. From that instant biomedical imaging sector has witnessed large growth. Photoacoustic tomography is a hybrid technique which merges ultrasound and optics. It employs both mechanical and photonic energy, sound waves and non-ionizing ER waves respectively. In this technique non-ionizing waves are utilized to excite and produce ultrasonic emission.

The sector of photoacoustic tomography process has faced noticeable progress in the past decades. Although various pure optical imaging methods, comprising two-photon microscopy, confocal microscopy and optical coherence tomography have become highly beneficial, none of these methods can provide insertion above ~1 mm into scattering biological tissues, due to their dependency on ballistic and quasi-ballistic photons. Hence there exists a limit in high resolution optical imaging process above this insertion limit. Photoacoustic tomography, which merges high ultrasonic resolution and high optical contrast in a single sight, has eliminated this disadvantage and filled this vacancy. The sector of photoacoustic tomography (PAT) process has progressed with a great interest in the past decades. PAT is an imaging method working on the principle of photoacoustic effect. PAT merges high ultrasonic resolution and high optical contrast in a single sight, able to produce high-resolution functional, structural, and molecular imaging in optically scattering biological tissue at new penetration limits.

II. PHOTOACOUSTIC TOMOGRAPHY

PAT comprises three processes viz. optical irradiation, ultrasonic detection, and image formation. The tissue is generally irradiated by a laser beam with short pulse to generate acoustic and thermal impulse responses. Light absorbed in tissue is transformed into heat, which is then transformed into a pressure rise through thermo-elastic expansion of the tissue.

The beginning pressure rise measured by the local optical energy displacement also known as specific optical absorption (J/m^3) and other mechanical and thermal characteristics propagates in the tissue as an ultrasonic signal, which is called as a photoacoustic signal. The photoacoustic signal is analyzed by ultrasonic transducers kept outside the tissue, generating electric waves. The electric waves are then amplified, digitized, and transmitted to a computer, where an image is produced. PAT relies on any unscattered or scattered absorbed photons to generate photoacoustic waves until the photon excitation is retarded thermally.

Fig. 1 represents a schematic method to describe the photoacoustic effect principle, where a short pulse light is introduced to the photoacoustic source and afterwards the incident light wave is absorbed and transformed into acoustic signals through thermal expansion method. PAT is very much sensitive to optical absorption process. PAT works on unlighted background when there is absence of absorption; the background signal value is zero. This dark background permits sensitive detection. In the existence of absorption, the foreground signal is directly proportional to the absorption coefficient.

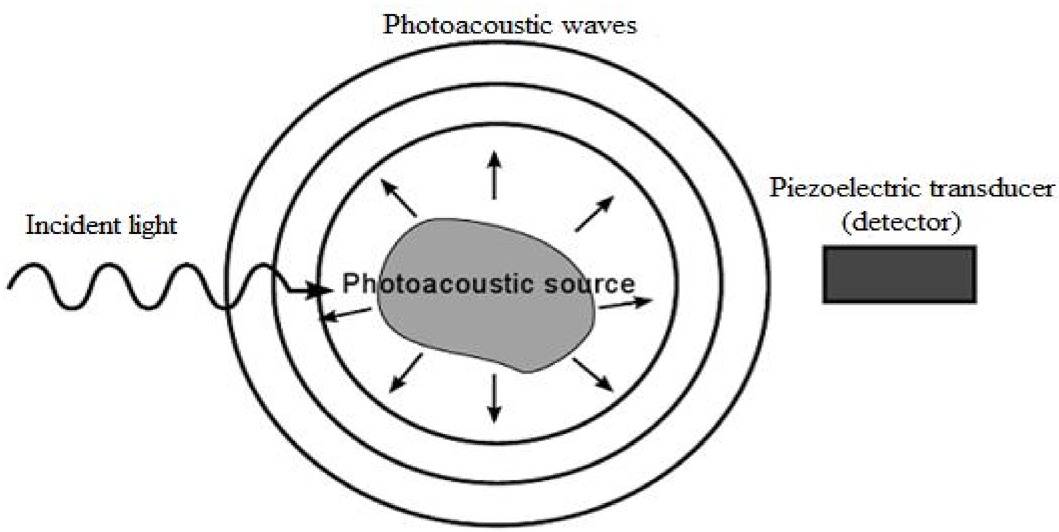


Fig. 1 Photoacoustic tomography principle

Any minor fractional variation in optical absorption coefficient results in an equal amount of fractional variation in PAT signal, which implies a relative sensitivity of 100%. The spatial resolution is developed from ultrasonic investigation in the photoacoustic emission process. Optical scattering is much stronger than ultrasonic scattering. The wavelength of detected photoacoustic signal is very short. Due to this, photoacoustic waves produce better resolution than optical signals beyond the soft penetration limit. Among other factors, the bandwidth and center frequency of ultrasonic detection system primarily measure the spatial resolution of PAT. The larger the center frequency and the broader the bandwidth, the enhanced the spatial resolution is.

III. IMAGE RECONSTRUCTION METHODS

The forward system equation of photoacoustic imaging process can be expressed in form of matrix equation as Ax = b where A denotes the system matrix. Suppose the dimension of A is 30,000 x 40401. The image to be recreated is denoted by the vector x. Vector x has the length of 40401. The measurement vector is denoted by b. It has a length of 30,000. Various techniques are employed to solve this equation some of them are illustrated below:

1) *Back Projection (BP) Method:* It is the simplest image recreation technique in which the expression to calculate the solution is given as

$$x = A^T b$$

Here A^T represents transpose of system matrix A. Multiplication of A^T with measurement vector b is equivalent to back projection. The back projection technique is computationally very much efficient because it is a non-iterative technique. Back-projection technique is known to produce only qualitative outputs, and may not be the correct choice for quantitative imaging calculations. Additionally, the measurement vector, b is usually corrupted by noise, so this kind of back-projection over amplifies the noise signal.

2) *Regularized Pseudoinverse Computation Based Method:* The best optimized solution for this technique is represented as

$$x = A^+ b$$

Where A^+ represents pseudoinverse of matrix A. A^+ is given as

$$A^+ = (A^T A)^{-1} A^T$$

This technique [39] is computationally very expensive. Computation of A^+ needs a lot of simulation time. Also, the solution x can contain noise signals. To enhance the solution we utilize a regularization parameter λ such that the regularized pseudoinverse becomes:

$$A_{reg}{}^+ = (A^T A + \lambda I)^{-1} A^T$$

And

$$x = A_{reg}{}^+ b$$

This equation is utilized to reconstruct the image which is denoted by the vector x.

3) *Block Coordinate Descent Method:* Block coordinate descent (BCD) technique optimizes the objective function over one segment (group of variables) x_j at each sub-iteration, while keeping all the other segments $x_i \neq x_j$ fixed. The global convergence of the BCD [40] iterates has been established for minimizing a convex non-differentiable function with certain separability and regularity properties. For the linear regression problem,

$$f(x) = \frac{1}{2}|b - Ax|^2$$

If we minimize above equation over x_i with all x_j fixed such that $i \neq j$,

$$0 = \nabla_i f(x) = A_i^T(Ax - b) = A_i^T(A_i x_i + A_{-i}x_{-i} - b)$$

i.e. we get,

$$x_i = \frac{A_i^T(b - A_{-i}x_{-i})}{A_i^T A_i}$$

The iterative method is utilized to estimate the vector x. Another method is to first establish the block-wise pseudoinverse matrix i.e.

$$A_i^+ = (A_i^T A_i)^{-1} A_i^T$$

Hence

$$x_i = A_i^+(b - A_{-i}x_{-i})$$

Further, with a use of regularization parameter λ above equation can be rewritten as

$$x_i = (A_i^T A_i + \lambda I)^{-1} A_i^T (b - A_{-i}x_{-i})$$

In this study, above equation is used for the reconstruction of the image represented by vector x.

IV. PROPERTIES OF TISSUE

Optical properties of the tissue and tumor are crucial parameters in a PA system affecting the sensitivity, dynamic range, and maximum detectivity of the instrument. The most important optical properties are the absorption coefficient μ_a (cm^{-1}), the scattering coefficient μ_s (cm^{-1}), and the anisotropy factor, g. The anisotropy factor is the mean cosine of the scattering angle, and demonstrates the directional probability of light scattering. The reduced scattering coefficient is defined as:

$$\mu_s' = \mu_s(1 - g)$$

where maximum value of g=1 indicates complete forward scattering and its minimum value g=-1 describes complete backward scattering. In human tissue, this parameter is typically in the range 0.7 to 0.9, enables the PA method to generate images from deep tissue where it is unreachable by pure optical methods. Tissue absorption and scattering are wavelength dependent. The popular wavelength range for PA imaging is IR and near-IR window due to low scatting in this range. In the body, there are four main absorbing components in this frequency window, water, lipid, deoxyhemoglobin (Hb) and oxyhemoglobin (HbO_2). The last two are the most important for diagnostic purposes. Hb components absorb the 755 nm irradiation 8 times stronger than HbO_2, and HbO_2 components absorb the 1064 nm irradiation 6 times stronger than Hb.The most important acoustic parameters of the tissue are the speed of sound and acoustic attenuation. The mean speed of sound in the tissue is ~1540 m/s with less than 10% deviation The acoustic attenuation describes the change of acoustic intensity as a function of depth. The acoustic attenuation coefficient is frequency dependent.

V. PROPOSED WORK

In this study a numerical blood vessel phantom is considered. This image is analyzed in MATLAB tool on the two configurations of computer first is having 8 GB RAM and second is having 64 GB RAM. For all the results presented here the regularization parameter was set to λ=0.018 as this was the best suited value observed for all imaging structures. Simulation results show that 64 GB RAM takes less time and produces better results than 8 GB RAM computer. Computer with 8 GB RAM does not support matrix dimension above 201 x 201. Computer with 64 GB RAM is used to analyze matrix dimension of 501 x 501.

Due to the high intrinsic contrast that blood cells give, photoacoustic imaging is widely used for visualizing internal blood vessel structures, both in the brain as well under the skin. The k-wave was used to generate the simulated PA data for 60 detector circular locations. 40 dB of noise was added to the simulated data. Fig. 2 shows the reconstructions obtained through back-projection method, reconstruction obtained through direct pseudo inverse computation through BCD method and line profile for 8 GB RAM and Fig. 3 shows results for 64 GB RAM.

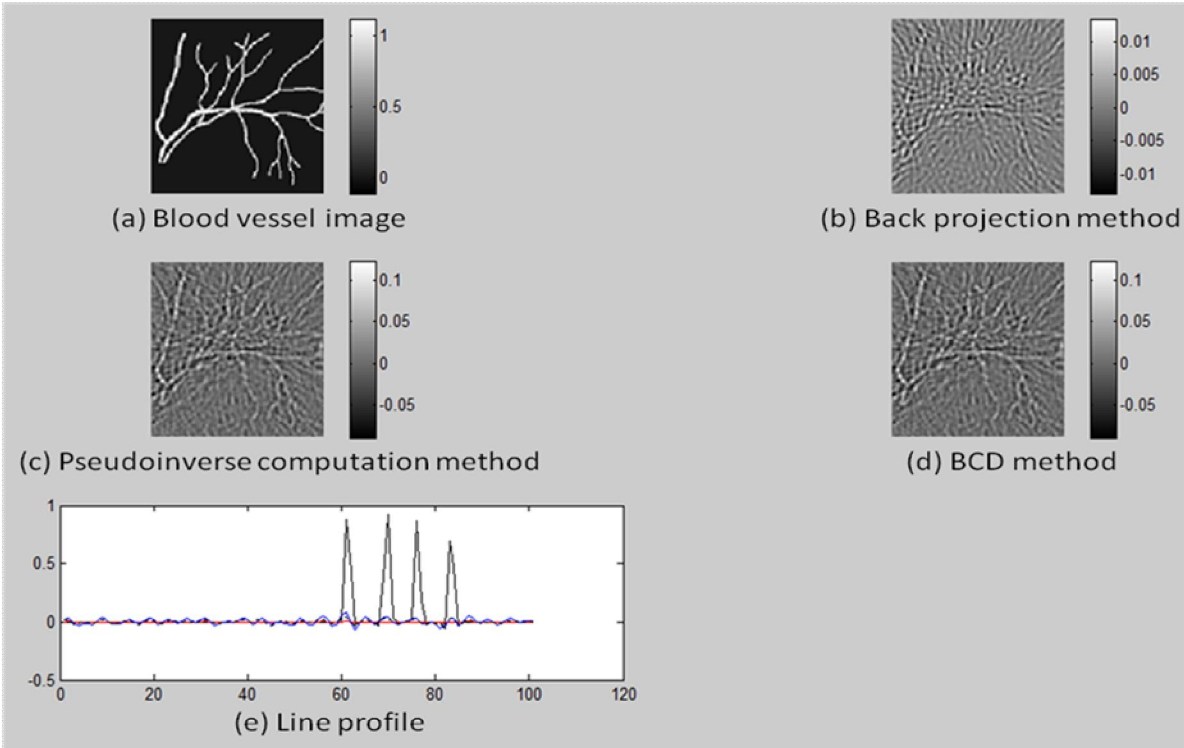
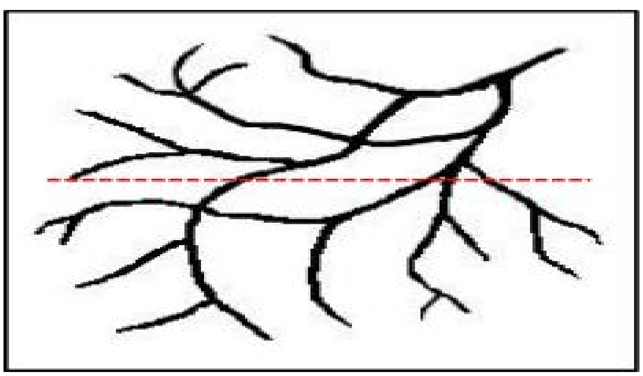
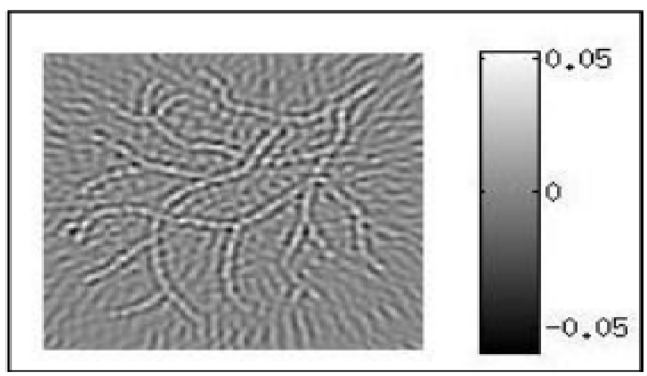


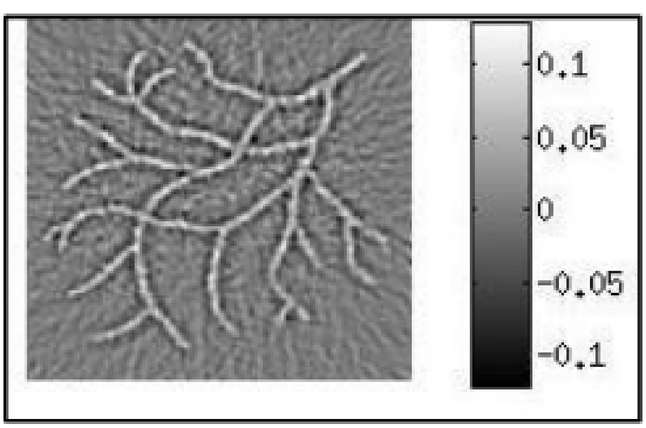
Fig. 2 Results for blood vessel image simulated in computer with 8 GB RAM



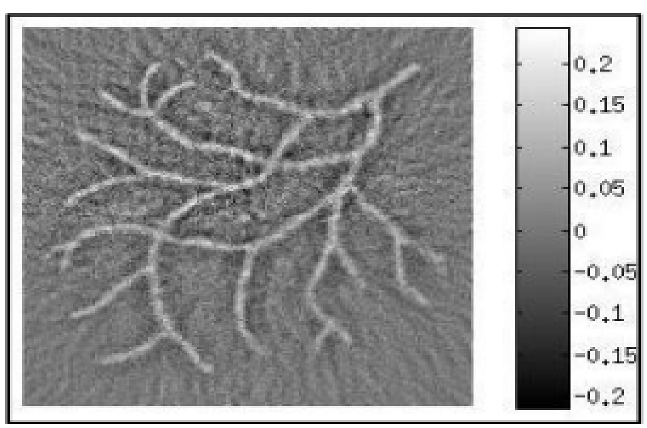
(a) Blood vessel image



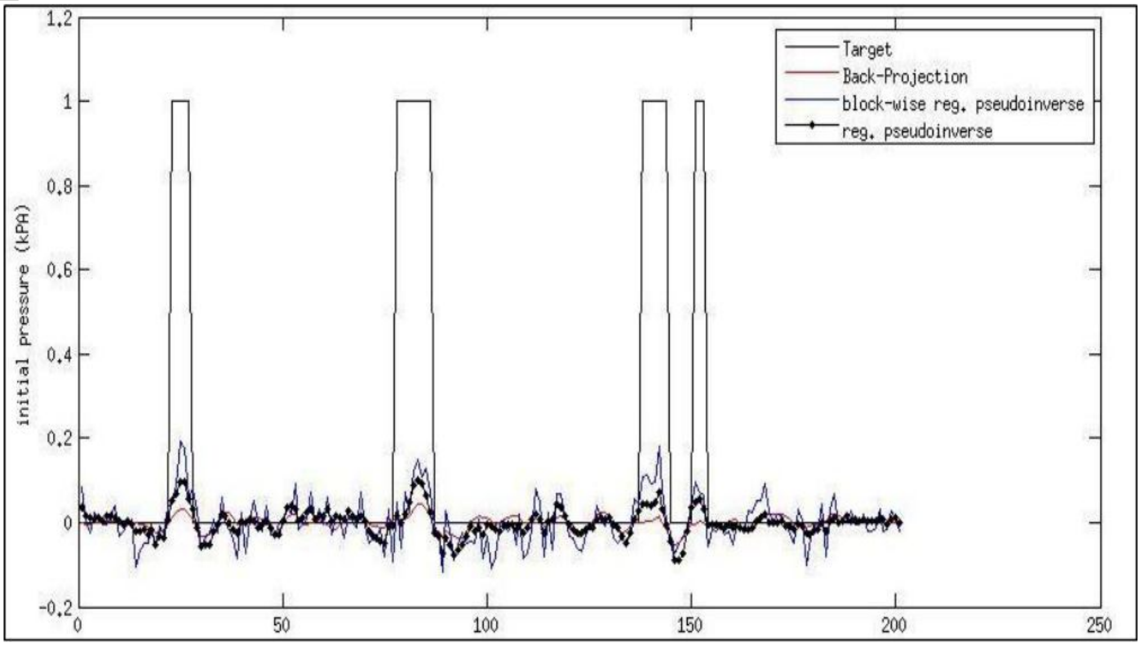
(b) Back projection method



(c) Pseudoinverse computation method



(d) BCD method



(e) Line profile

Fig. 3 Results for blood vessel image simulated in computer with 64 GB RAM

VI. CONCLUSION

In this work different methods are used for image reconstruction in photoacoustic tomography. It is concluded that BCD method takes much lesser time by dividing it into uniform size blocks. When the 201x201 grid size is divided into four blocks, this computation took one third of the time when computed directly. A division into nine blocks took almost one fifth times. The image reconstruction through BCD scheme gives quantitatively better result. These results show that increasing RAM size makes the images more clear and takes less time in computation.

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