Numerical methods and machine learning algorithms for solution of Inverse problems

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Microwave medical imaging in monitoring of hyperthermia: the course project and code

- Presentation of the course project *"Regularized algorithms for detection of tumours in microwave medical imaging"*. Matlab code (data and programs, zip file) with an example see in CANVAS, “Computer Projects”:

  `http://www.math.chalmers.se/Math/Grundutb/CTH/tma265/2021/IPcourse/Project_Hyperthermi.pdf`

- Matlab code: in CANVAS as well as in

  `https://github.com/ProjectWaves24/MicrowaveHyperMatlab`

- Advanced C++/PETSc computations and visualization in paraview: `https://github.com/ProjectWaves24/MESH`. Needs account in github, contact me.

- Advanced C++/PETSc computations using adaptive FEM:

  `https://github.com/ProjectWaves24/MicrowaveHypAFEM2`

  Needs account in github, contact me.

Statement of an ill-posed problem

Let $\Omega \subset \mathbb{R}^n, n = 2, 3$ which is a bounded domain with the boundary $\partial\Omega$. Our goal is to solve a Fredholm integral equation of the first kind

$$\int_\Omega \rho(x, y) z(x) dx = u(y) \quad y \in \Omega, \tag{1}$$

where $u(y) \in L_2(\Omega)$, $z(x) \in H$, $\rho(x, y) \in C^k(\Omega \times \Omega), k \geq 0$ is the kernel of the integral equation. We can rewrite (1) in an operator form as

$$A(z) = u \tag{2}$$

with an operator $A : H \to L_2(\Omega)$ defined as

$$A(z) := \int_\Omega \rho(x, y) z(x) dx. \tag{3}$$

The Problem (P).
Let $z(x) \in H$ in

$$\int_\Omega \rho(x, y) z(x) dx = u(y) \quad y \in \Omega, \tag{4}$$

be unknown in Ω. Determine $z(x) \in H$ for $x \in \Omega$ assuming that functions $\rho(x, y) \in C^k(\Omega \times \Omega), k \geq 0$ and $u(y) \in L_2(\Omega)$ in (4) are known.

The Tikhonov functional

Let W_1, W_2, Q be three Hilbert spaces, $Q \subseteq W_1$ as a set. We denote scalar products and norms in these spaces as

$$(\cdot,\cdot), \|\cdot\| \text{ for } W_1,$$

$$(\cdot,\cdot)_2, \|\cdot\|_2 \text{ for } W_2$$

$$\text{and } [\cdot,\cdot], [\cdot] \text{ for } Q.$$

Let $A : W_1 \to W_2$ be a bounded linear operator. Our goal is to find the function $z \in Q$ which minimizes the Tikhonov functional

$$J_\alpha(z) = \frac{1}{2}\|Az - u\|_2^2 + \frac{\alpha}{2}[z]^2, u \in W_2; z \in Q, \tag{5}$$

where $\alpha > 0$ is a regularization parameter. We search for a stationary point of the above functional with respect to z satisfying $\forall b \in Q$

$$J'_\alpha(z)(b) = 0, \tag{6}$$

where $J'_\alpha(z)$ is the Fréchet derivative of the functional (5).

The Tikhonov functional

When the operator $A : L_2 \to L_2$ the following Lemma is valid:
Lemma 1a [BKS] *Let $A : L_2 \to L_2$ be a bounded linear operator. Then the Fréchet derivative of the functional (5) is*

$$J'_\alpha(z)(b) = (A^*Az - A^*u, b) + \alpha [z, b], \forall b \in Q. \tag{7}$$

In particular, for the integral operator (4) we have

$$J'_\alpha(z)(b) = \int_\Omega b(s) \left[\int_\Omega z(y) \left(\int_\Omega \rho(x,y) \rho(x,s)\, dx \right) dy - \int_\Omega \rho(x,s) u(x)\, dx \right] ds$$
$$+\alpha [z, b], \forall b \in Q. \tag{8}$$

[BKS] A. B. Bakushinsky, M. Y. Kokurin, A. Smirnova, *Iterative methods for ill-posed problems*, Walter de Gruyter GmbH&Co., 2011.

The Tikhonov functional

When the operator $A : H^1 \to L_2$ the following Lemma is valid:
Lemma 1b [BGN] Let $A : H^1(\Omega) \to L_2(\Omega_\kappa)$ be a bounded linear operator. Then the Fréchet derivative of the functional

$$M_\alpha(f) = \frac{1}{2}\|Af - u\|^2_{L_2(\Omega_\kappa)} + \frac{\alpha}{2}\|\,|\nabla f|\,\|^2_{L^2(\Omega)}, \qquad (9)$$

is

$$M'_\alpha(f)(b) = (A^*Af - A^*u, b) + \alpha(|\nabla f|, |\nabla b|),\ \forall b \in H^1(\Omega), \qquad (10)$$

with a convex growth factor b, i.e., $|\nabla b| < b$

[BGN] L. Beilina, G. Guillot, K. Niinimäki„ The Finite Element Method and Balancing Principle for Magnetic Resonance Imaging, Springer Proceedings in Mathematics and Statistics, vol 328. Springer, Cham (2020).

Lemma 2 is also well known since $A : W_1 \to W_2$ is a bounded linear operator.

Lemma 2 [TGSY] *Let the operator $A : W_1 \to W_2$ be a bounded linear operator which has the Fréchet derivative of the functional (5). Then the functional $J_\alpha(z)$ is strongly convex on the space Q and*

$$\left(J'_\alpha(x) - J'_\alpha(z), x - z\right) \geq \alpha [x - z]^2, \forall x, z \in Q.$$

It is known from the theory of convex optimization that Lemma 2 implies existence and uniqueness of the global minimizer $z_\alpha \in Q$ of the functional J_α such that

$$J_\alpha(z_\alpha) = \inf_{z \in Q} J_\alpha(z).$$

[TGSY] A.N. Tikhonov, A.V. Goncharsky, V.V. Stepanov and A.G. Yagola, *Numerical Methods for the Solution of Ill-Posed Problems*, London: Kluwer, London, 1995.

Balancing principle to find regularization parameter

$$M_\alpha(f) = \frac{1}{2}\|Af - u\|^2_{L_2(\Omega_k)} + \alpha\frac{1}{2}\|f\|^2_{H^1(\Omega)} = \varphi(f) + \alpha\psi(f). \qquad (11)$$

For the functional (11) the value function $F(\alpha) : \mathbb{C} \to \mathbb{C}$ is defined as

$$F(\alpha) = \inf_f M_\alpha(f). \qquad (12)$$

If there exists derivative $F'(\alpha)$ at $\alpha > 0$ then from (11) and (12) follows that

$$F(\alpha) = \inf_f M_\alpha(f) = \underbrace{\varphi'(f)}_{\bar{\varphi}(\alpha)} + \alpha \underbrace{\psi'(f)}_{\bar{\psi}(\alpha)}. \qquad (13)$$

Since $F'_\alpha(\alpha) = \psi'(f) = \bar{\psi}(\alpha)$ then from (13) we get

$$\bar{\psi}(\alpha) = F'(\alpha), \;\; \bar{\varphi}(\alpha) = F(\alpha) - \alpha F'(\alpha). \qquad (14)$$

For the functional (11) balancing principle (or Lepskii) finds $\alpha > 0$ such that the following expression is fulfilled

$$\bar{\varphi}(\alpha) = \gamma\alpha\bar{\psi}(\alpha), \qquad (15)$$

K. Ito, B. Jin, Inverse Problems: Tikhonov theory and algorithms, Series on Applied Mathematics, V.22, World Scientific, 2015.

Balancing principle

When $\gamma = 1$ the method is called zero crossing method. The balancing rule (15) finds optimal $\alpha > 0$ minimizing the balancing function

$$\Phi_\gamma(\alpha) = \frac{F^{1+\gamma}(\alpha)}{\alpha}. \tag{16}$$

From conditions (14) it follows that

$$0 = \bar{\varphi}(\alpha) - \gamma\alpha\bar{\psi}(\alpha) = F(\alpha) - \alpha F'(\alpha) - \gamma\alpha F'(\alpha) = F(\alpha) - \alpha F'(\alpha)(1+\gamma),$$

which can be rewritten as

$$F(\alpha) = \alpha F'(\alpha)(1+\gamma). \tag{17}$$

We can check that the minimum of $\Phi_\gamma(\alpha)$ is achieved at

$$0 = (\Phi_\gamma(\alpha))'_\alpha = \frac{(1+\gamma)F'(\alpha)F^\gamma(\alpha)\alpha - F^{1+\gamma}(\alpha)}{\alpha^2}.$$

From the above equation we get

$$(1+\gamma)F'(\alpha)F^\gamma(\alpha)\alpha = F^{1+\gamma}(\alpha) \to (1+\gamma)F'(\alpha)\alpha = F(\alpha).$$

This equation is the same as the equation (17) which gives the balancing principle.

Fixed point algorithm: constant value of *alpha*

- Step 0. Start with the initial approximations α_0 and compute the sequence of α_k in the following steps.
- Step 1. Compute the value function $F(\alpha_k) = \inf_f M_{\alpha_k}(f)$ for (11) and get reconstruction f_{α_k}.
- Step 2. Update the regularization parameter $\alpha := \alpha_{k+1}$ as

$$\alpha_{k+1} = \frac{||\bar{\varphi}(\alpha_k)||_2}{||\bar{\psi}(\alpha_k)||_2}$$

- Step 3. Choose tolerance $0 < \theta < 1$. Stop computing regularization parameters α_k if computed α_k are stabilized, i.e., if $|\alpha_k - \alpha_{k-1}| \leq \theta$. Otherwise, set $k := k + 1$ and go to Step 1.

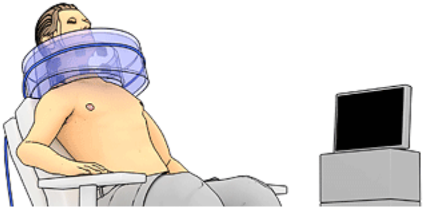
Fixed point algorithm: vector of parameters *alpha*

- Step 0. Start with the initial approximations α_0 and compute the sequence of α_k in the following steps.
- Step 1. Compute the value function $F(\alpha_k) = \inf_f M_{\alpha_k}(f)$ for (11) and get reconstruction f_{α_k}.
- Step 2. Update the regularization vector of parameters $\alpha := \alpha_{k+1}$ as

$$\alpha_{k+1} = \frac{\bar{\varphi}(\alpha_k)}{\bar{\psi}(\alpha_k)}$$

- Step 3. Choose tolerance $0 < \theta < 1$. Stop computing regularization parameters α_k if computed α_k are stabilized. Otherwise, set $k := k + 1$ and go to Step 1.

Microwave medical imaging in monitoring of hyperthermia



- Joint work with the group of Biomedical Imaging at the Department of Electrical Engineering at CTH, Chalmers.
- Microwave hyperthermia is used for cancer therapies: it increases the tumour temperature to $40 - 44^\circ C$ keeping healthy tissue at the normal temperature.
- Thermal dose monitoring is critical for treatment. Thus, robust real-time methods for localization of the focal point in the target are needed.
- AFEM with combination of least squares method is applied in microwave thermometry for non-invasive monitoring of hyperthermia [1].

[1] M. G. Aram, L. Beilina, H. Dobsicek Trefna, Microwave Thermometry with Potential Application in Non-invasive Monitoring of Hyperthermia, *Journal of Inverse and Ill-posed problems*, https://doi.org/10.1515/jiip-2020-0102, 2020.

Consider a region of space that has no electric or magnetic current sources, but may have materials that absorb electric or magnetic field energy. Then, using MKS units, the time-dependent Maxwell's equations are given in differential and integral form by *Faraday's law* :

$$\frac{\partial \mathbf{B}}{\partial t} = -\nabla \times \mathbf{E} - \mathbf{M} \quad (18a)$$

$$\frac{\partial}{\partial t} \iint_A \mathbf{B} \cdot \mathbf{dA} = -\oint_L \mathbf{E} \cdot \mathbf{dL} - \iint_A \mathbf{M} \cdot \mathbf{dA} \quad (18b)$$

The MKS system of units is a physical system of units that expresses any given measurement using fundamental units of the metre, kilogram, and/or second (MKS))

A. Taflove, S. C. Hagness, Computational Electromagnetics. The finite-difference time-domain method, 3rd edition, *Artech House Publishers*, 2005.

Maxwell's equations

Ampere's law :

$$\frac{\partial \mathbf{D}}{\partial t} = \nabla \times \mathbf{H} - \mathbf{J} \tag{19a}$$

$$\frac{\partial}{\partial t} \iint_A \mathbf{D} \cdot \mathbf{dA} = \oint_L \mathbf{H} \cdot \mathbf{dL} - \iint_A \mathbf{J} \cdot \mathbf{dA} \tag{19b}$$

Gauss' law for the electric field :

$$\nabla \cdot \mathbf{D} = 0 \tag{20a}$$

$$\oiint_A \mathbf{D} \cdot \mathbf{dA} = 0 \tag{20b}$$

Gauss' law for the magnetic field :

$$\nabla \cdot \mathbf{B} = 0 \tag{21a}$$

$$\oiint_A \mathbf{B} \cdot \mathbf{dA} = 0 \tag{21b}$$

Maxwell's equations

In (18) to (21), the following symbols (and their MKS units) are defined:

$\mathbf{E}$	:	electric field (volts/meter)
$\mathbf{D}$	:	electric flux density (coulombs/meter^2)
$\mathbf{H}$	:	magnetic field (amperes/meter)
$\mathbf{B}$	:	magnetic flux density (webers/meter^2)
A	:	arbitrary three-dimensional surface
$\mathbf{dA}$	:	differential normal vector that characterizes surface A (meter^2)
L	:	closed contour that bounds surface A (volts/meter)
$\mathbf{dL}$	:	differential length vector that characterizes contour L (meters)
$\mathbf{J}$	:	electric current density (amperes/meter^2)
$\mathbf{M}$	:	equivalent magnetic current density (volts/meter^2)

Maxwell's equations

In linear, isotropic, nondispersive materials (i.e. materials having field-independent, direction-independent, and frequency-independent electric and magnetic properties), we can relate **D** to **E** and **B** to **H** using simple proportions:

$$\mathbf{D} = \varepsilon \mathbf{E} = \varepsilon_r \varepsilon_0 \mathbf{E}; \quad \mathbf{B} = \mu \mathbf{H} = \mu_r \mu_0 \mathbf{H} \tag{22}$$

where

ε	:	electrical permittivity (farads/meter)
ε_r	:	relative permittivity (dimensionless scalar)
ε_0	:	free-space permittivity (8.854×10^{-12} farads/meter)
μ	:	magnetic permeability (henrys/meter)
μ_r	:	relative permeability (dimensionless scalar)
μ_0	:	free-space permeability ($4\pi \times 10^{-7}$ henrys/meter)

Note that **J** and **M** can act as *independent sources* of E- and H-field energy, $\mathbf{J}_{source}$ and $\mathbf{M}_{source}$.

We also allow for materials with isotropic, nondispersive electric and magnetic losses that attenuate E- and H-fields via conversion to heat energy. This yields

$$\mathbf{J} = \mathbf{J}_{source} + \sigma \mathbf{E}; \quad \mathbf{M} = \mathbf{M}_{source} + \sigma^* \mathbf{H} \tag{23}$$

where
σ : electric conductivity (siemens/meter)
σ^* : equivalent magnetic loss (ohms/meter)

Finally, we substitute (22) and (23) into (18a) and (19a). This yields Maxwell's curl equations in linear, isotropic, nondispersive, lossy materials:

$$\frac{\partial \mathbf{H}}{\partial t} = -\frac{1}{\mu} \nabla \times \mathbf{E} - \frac{1}{\mu} \left(\mathbf{M}_{source} + \sigma^* \mathbf{H} \right) \tag{24}$$

$$\frac{\partial \mathbf{E}}{\partial t} = \frac{1}{\varepsilon} \nabla \times \mathbf{H} - \frac{1}{\varepsilon} \left(\mathbf{J}_{source} + \sigma \mathbf{E} \right) \tag{25}$$

Write now Maxwell's curl equations in linear, isotropic, nondispersive, lossy materials with $\sigma^* = 0, \mathbf{M}_{source} = 0$:

$$\frac{\partial \mathbf{H}}{\partial t} = -\frac{1}{\mu} \nabla \times \mathbf{E} \tag{26}$$

$$\frac{\partial \mathbf{E}}{\partial t} = \frac{1}{\varepsilon} \nabla \times \mathbf{H} - \frac{1}{\varepsilon} \sigma \mathbf{E} - \frac{1}{\varepsilon} \mathbf{J}_{source} \tag{27}$$

Taking now $\frac{\partial}{\partial t}$ from (27) and multiplying by ε, and then taking $\nabla \times$ from (26), we have:

$$\nabla \times \frac{\partial \mathbf{H}}{\partial t} = -\nabla \times \frac{1}{\mu} \nabla \times \mathbf{E} \tag{28}$$

$$\varepsilon \frac{\partial^2 \mathbf{E}}{\partial t^2} = \frac{\partial}{\partial t} \nabla \times \mathbf{H} - \sigma \frac{\partial}{\partial t} \mathbf{E} - \frac{\partial}{\partial t} \mathbf{J}_{source} \tag{29}$$

CIPs for electric wave propagation

Substitude the right hand side of (28) into (29) instead of $\frac{\partial}{\partial t}\nabla \times \mathbf{H}$ to obtain Maxwell's equations for electric field $\mathbf{E} = (E_1, E_2, E_3)$. Let us consider now Cauchy problem for the Maxwell's equations for electric field $\mathbf{E}$ in the domain $\Omega_T = \Omega \times [0, T]$:

$$
\begin{aligned}
\varepsilon \frac{\partial^2 \mathbf{E}}{\partial t^2} + \nabla \times \frac{1}{\mu} \nabla \times \mathbf{E} = -\sigma \frac{\partial}{\partial t}\mathbf{E} - \frac{\partial}{\partial t}\mathbf{J}_{source} \text{ in } \Omega_T, \\
\nabla \cdot (\varepsilon \mathbf{E}) = 0, \\
\mathbf{E}(\mathbf{x}, \mathbf{0}) = f_0(x), \quad \mathbf{E_t}(\mathbf{x}, \mathbf{0}) = f_1(x) \text{ in } \Omega,
\end{aligned}
\tag{30}
$$

- Let $\Omega \subset \mathbb{R}^3$ be a convex bounded domain with the boundary $\partial\Omega \in C^3$ and specify time variable $t \in [0, T]$. Next, we supply the Cauchy problem by the appropriate b.c.
- $\varepsilon(x)$ and $\sigma(x)$ are dielectric permittivity and electric conductivity functions, respectively of the domain Ω. In (30), $\varepsilon(x) = \varepsilon_r(x)\varepsilon_0, \mu = \mu_r\mu_0$ and $\sigma(x)$ are dielectric permittivity, permeability and electric conductivity functions, respectively, ε_0, μ_0 are dielectric permittivity and permeability of free space, respectively.

CIPs for electric wave propagation

Ω

$\varepsilon_r(x) = ?$

$\sigma = 0, \mu_r = 1$

$E(x,t) = g(x,t)$ on $\partial\Omega$

Ω

$\varepsilon_r(x) = ?$
$\sigma(x) = ?$

$\mu_r \approx 1$

$E(x,t) = g(x,t)$ on $\partial\Omega$

Inverse Problem (EIP1) *Determine the relative dielectric permittivity function $\varepsilon_r(x)$ in Ω for $x \in \Omega$ in nonconductive ($\sigma(x) = 0$) and nonmagnetic ($\mu_r = 1$) media when the measured function $g(x,t)$ s.t.*

$$\mathbf{E}(x,t) = g(x,t), \forall (x,t) \in \partial\Omega \times (0,T].$$

is known in Ω.

Inverse Problem (EIP2) *Determine the functions $\epsilon(x), \sigma(x)$ in Ω for $x \in \Omega$ for $\mu_r \approx 1$ in water assuming that $g(x,t)$ is known in $\partial\Omega \times (0,T]$.*

Maxwell's equations in frequency domain

Assuming $\mathbf{E}(\mathbf{x}, \mathbf{t}) = \widehat{E}(x, \omega) \cdot e^{-i\omega t}$ and $\mathbf{J}_{source} = \widehat{J}(x, \omega) \cdot e^{-i\omega t}$ and applying this to (30) with $\mu_r = 1$ we obtain the following vector wave equation:

$$\nabla \times \nabla \times \widehat{E}(x, \omega) - \omega^2 \left(\frac{\varepsilon_r(x)}{c^2} + i\mu_0 \frac{\sigma(x)}{\omega} \right) \widehat{E}(x, \omega) = i\omega\mu_0 \widehat{J}(x, \omega). \quad (31)$$

We introduce the spatially distributed complex dielectric function $\varepsilon'(x)$:

$$\varepsilon'(x) = \varepsilon_r(x) \frac{1}{c^2} + i\mu_0 \frac{\sigma(x)}{\omega}, \quad (32)$$

where ω is the angular frequency. Then the equation (31) transforms to the equation

$$\nabla \times \nabla \times \widehat{E}(x, \omega) - \omega^2 \varepsilon'(x) \widehat{E}(x, \omega) = i\omega\mu_0 \widehat{J}(x, \omega). \quad (33)$$

which should be supplied by appropriate boundary conditions.

Applying $\nabla \times \nabla \times \widehat{E} = \nabla(\nabla \cdot \widehat{E}) - \nabla \cdot (\nabla \widehat{E})$ and in case of $\mathbf{E}(\mathbf{x}, \mathbf{t}) = \widehat{E}(x, \omega) \cdot e^{i\omega t}$ we obtain inhomogeneous Helmholtz equation

$$\triangle \widehat{E} + k^2 \widehat{E} = i\omega\mu_0 \widehat{J}, \tag{34}$$

where $k^2 = \omega^2 \varepsilon'$. This equation can be rewritten for the solution $\widehat{E} = E(r)$ in cylindrical coordinates and in transverse electric (TE) mode as a Bessel equation

$$\left(\frac{1}{r} \frac{\partial}{\partial r} (r \frac{\partial}{\partial r}) + k^2 \right) E = i\omega\mu_0 J. \tag{35}$$

The general solution to this equation is in the form

$$E(r) = AJ_0(kr) + BN_0(kr), \tag{36}$$

where J_o and N_0 are zero-order Bessel's functions of the first and second order, respectively. The time-harmonic solution of the equation (35) is given by

$$E(r, \omega) := E(r) = -\frac{\omega\mu_r}{4} \int_S J H_0^{(2)}(kR)\, dS, \tag{37}$$

for a generalized source initialized at r_0 and $R = |r - r_0| = \sqrt{r^2 + {r_0}^2 - 2rr_0 cos(\varphi - \varphi_0)}$.

[BE] L. Beilina and A. Eriksson, Reconstruction of dielectric constants in a cylindrical waveguide, *Inverse Problems and Applications, Springer Proceedings in Mathematics & Statistics*, Vol. 120, 2015.

Microwave Imaging: Differential Image Reconstruction

Let we have a bi-static pair $(i,\ j)$ of antennas located on the scan line Γ, i.e. $\mathbf{r_i},\ \mathbf{r_j}\ \in\ \Gamma$.
Using Lorentz reciprocity theorem and under Born approximation, the scattered electric field between the pair of antennas at angular frequency of ω can be written as

$$\mathbf{E}_{ji}^{s} \simeq i\omega\mu_0 k_b^2 I(\omega) \int_{\Omega} \overline{\mathbf{G}}(\mathbf{r_j},\mathbf{r}',\omega)\cdot \epsilon'(\mathbf{r}',\omega)\overline{\mathbf{G}}(\mathbf{r_i},\mathbf{r}',\omega)dv' \qquad (38)$$

where Ω is the imaging domain, $I(\omega)$ is the excitation current of the transmitter, k_b is the lossless background wavenumber, $\overline{\mathbf{G}}$ is the dyadic Green's function and ε' is defined as in (32).

Next, scattered fields $\mathbf{E}_{ji}^{s}$ are replaced with their corresponding S-parameters, as well as input power and characteristic impedance of the ports. Then (38) is transformed to the equation

$$S_{ji}^{sca}(\omega) \simeq C \int_{\Omega} \mathbf{E}_{inc,j}^{CST}(\mathbf{r}',\omega)\cdot \Delta O(\mathbf{r}',\omega)\mathbf{E}_{inc,i}^{CST}(\mathbf{r}',\omega)dv' \qquad (39)$$

where $C = -k_b^2/(4i\omega\mu)$ and $\mathbf{E}_{inc,i}^{CST}$ is the exported E-field from CST under irradiation of the i^{th} antenna. Here, $\Delta O = \varepsilon'(\mathbf{r}) - \varepsilon_b'(\mathbf{r})$, $\varepsilon_b'(\mathbf{r})$ is baseline.

Equation (39) is the standard Fredholm integral equation of the first kind, and thus, it is an ill-posed problem. It can be solved for an linear operator A by minimizing the Tikhonov regularization functional

$$F(\varepsilon') = \frac{1}{2} \left\| A\varepsilon' - d \right\|^2_{L_2(\Omega)} + \frac{\lambda}{2} \left\| \varepsilon' \right\|^2_{L_2(\Omega)}. \tag{40}$$

where $d = S^{sca}$, λ is the regularization parameter. The optimal value will be:

$$F'(\varepsilon') = A^* A\varepsilon' - A^* d + \lambda\varepsilon' = 0. \tag{41}$$

Discretizing operator A, we get the matrix $\mathbf{A}$ and the problem (41) will be rewritten as the system of normal equations

$$\varepsilon' = (\mathbf{A}^T\mathbf{A} + \lambda I)^{-1}\mathbf{A}^T d. \tag{42}$$

Applying SVD of $\mathbf{A} = U\Sigma V^T$ in we get the equation to reconstruct ε' :

$$\varepsilon' = V(\Sigma^2 + \lambda I)^{-1}\Sigma U^T d. \tag{43}$$

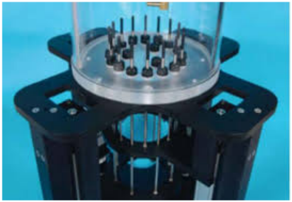
Applying SVD of $\mathbf{A} = U\Sigma V^T$ in we get the equation to reconstruct ε' :

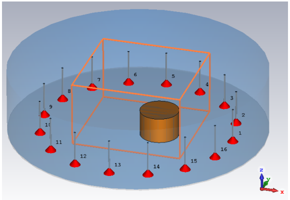
$$\varepsilon' = V(\Sigma^2 + \lambda I)^{-1}\Sigma U^T d. \tag{44}$$

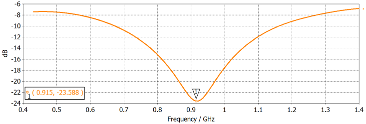
Proof: Since $\mathbf{A} = U\Sigma V^T$ then $\mathbf{A}^T = (U\Sigma V^T)^T = V\Sigma U^T$, then equation (25) can be written as:

$$\varepsilon' = (\mathbf{A}^T\mathbf{A}+\lambda I)^{-1}\mathbf{A}^T d = (V\Sigma U^T U\Sigma V^T+\lambda I)^{-1} V\Sigma U^T d = V(\Sigma^2+\lambda I)^{-1}\Sigma U^T d. \tag{45}$$

Reconstruction of heated target







Timeline (min)	$\varepsilon_r(t)$	$\sigma(t)$ (S/m)
t=0 (Baseline)	26	0.12
t=2	28	0.15
t=4	29	0.19
t=6	29.8	0.21
t=8	30.5	0.23
t=10	31	0.24
Coupling fluid 20:7 isopropyl : water ($\varepsilon_r = 24.5, \sigma = 0.46$)		

Microwave imaging for breast cancer detection. Top left: setup of the representation and actual photograph of the data acquisition platform for breast cancer detection used at CTH and Medfield Diagnostics AB: Assembled antenna hardware. Top right: schematic 3-D representation of 16 monopole antennas in a matching liquid tank, in CST(`http://www.cst.com`*); Bottom left: Return loss* S_{11} *of the designed antenna for the frequency band* 915 MHz. *Bottom right: permittivity and conductivity of the target as it starts to cool down from 55° C to 29° C over a ten-minute window of time.*

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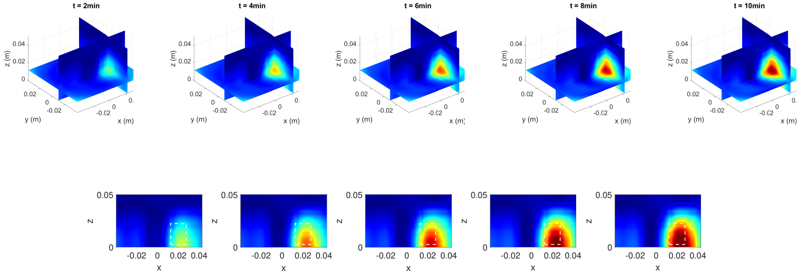
Reconstruction of heated target: least squares solution

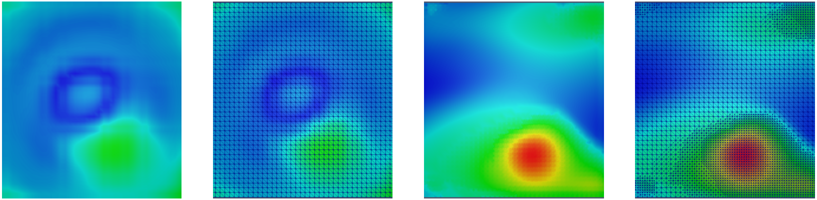
Geometry with $nno = 40 \times 42 \times 26 = 43680$.
Solution is obtained via the formula

$$\varepsilon' = (\mathbf{A}^T\mathbf{A}+\lambda I)^{-1}\mathbf{A}^T d = (V\Sigma U^T U\Sigma V^T+\lambda I)^{-1} V\Sigma U^T d = V(\Sigma^2+\lambda I)^{-1}\Sigma U^T d$$

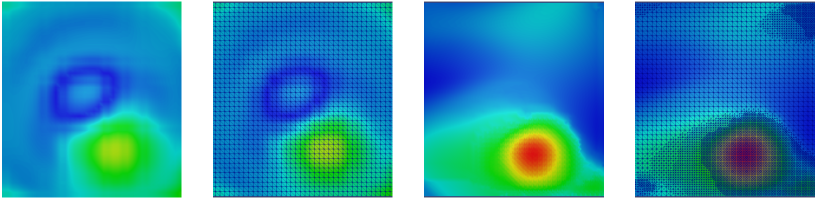
with $\lambda = 1$.



Reconstruction: Least Squares + AFEM, *xy*-plane

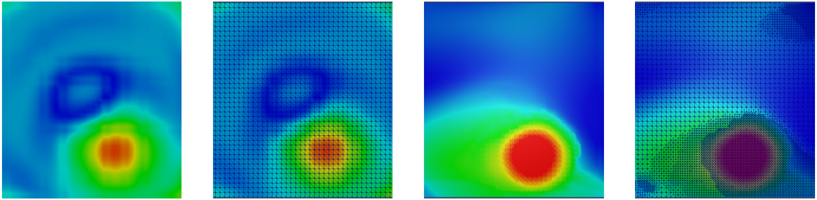


t = 2 min

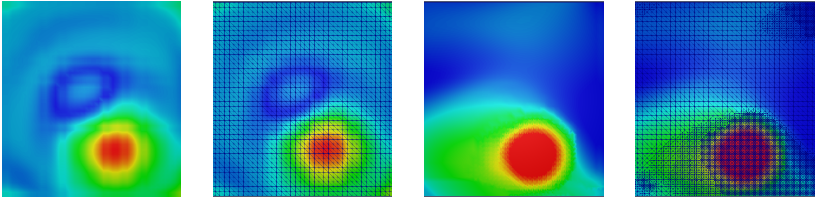


t = 4 min

Reconstruction: Least Squares + AFEM, *xy* plane

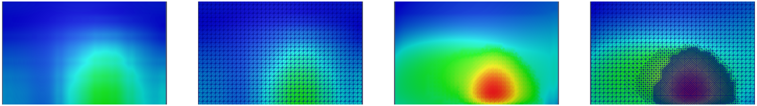


t = 8 min

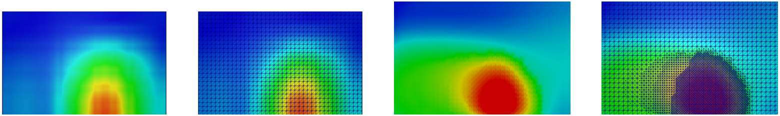


t = 10 min

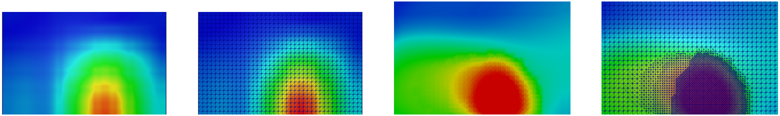
Reconstruction: Least Squares + AFEM, *zx* plane



t = 2 min



t = 8 min



t = 10 min

Convergence of fixed point algorithm and AFEM

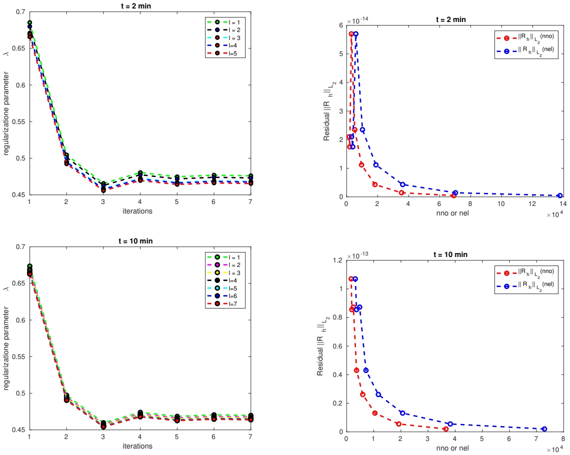


Figure: Left figures: convergence of fixed point algorithm. Here, *l* is the number of mesh refinement. Right figures: convergence of AFEM on adaptive locally refined meshes.

Project: Regularized algorithms for detection of tumours in microwave medical imaging

- In this project we will study different regularization strategies for detection of tumours using microwaves. This problem is a typical Coefficient Inverse Problem (CIP) for determination of complex dielectric permittivity function in Helmholtz equation from scattered electric field in frequency domain.
- Alternatively, the dielectric permittivity function can be determined from the solution of a Fredholm integral equation of the first kind which is an ill-posed problem.
- The goal of the current project is further development of mathematical methods presented in the recent paper [ABD] for real-life applications in microwave medical imaging.

M. G. Aram, L. Beilina, H. Dobsicek Trefna, Microwave Thermometry with Potential Application in Non-invasive Monitoring of Hyperthermia, *Journal of Inverse and Ill-posed problems*, 2020. https://doi.org/10.1515/jiip-2020-0102

Project: Regularized algorithms for detection of tumours in microwave medical imaging

More precisely, in this project students will:

- Study different regularized formulations of the reconstruction problem presented in the paper [ABD] which can be downloaded from the link

  https://doi.org/10.1515/jiip-2020-0102

- Determine the dielectric permittivity function by solving the regularized linear system of equations (LSE) in 3D by modifying existing Matlab code used for computations in the paper [ABD] available for download at

  http://www.math.chalmers.se/Math/Grundutb/CTH/tma265/2021/IPcourse/MatlabCode_MicrowaveImaging.zip.

  https://github.com/ProjectWaves24/MicrowaveHyperMatlab

Project: Regularized algorithms for detection of tumours in microwave medical imaging

- Test different regularization strategies (Morozov's discrepancy principle, Balancing principle) for choosing the regularization parameter λ.
- Test reconstructions with choosing different regularization terms, i.e. try to minimize

$$F(\varepsilon') = \frac{1}{2} \left\| A\varepsilon' - d \right\|^2_{L_2(\Omega)} + \frac{\lambda}{2} \left\| \nabla \varepsilon' \right\|^2_{L_2(\Omega)} . \tag{46}$$

or minimize

$$F(\varepsilon') = \frac{1}{2} \left\| A\varepsilon' - d \right\|^2_{L_2(\Omega)} + \frac{\lambda}{2} \left\| \varepsilon' + \nabla \varepsilon' \right\|^2_{L_2(\Omega)} . \tag{47}$$