Coil combine for conductivity mapping of breast cancer

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Introduction: Phase-based electrical property tomography (EPT)[1] was recently proposed which uses only RF (radio frequency) transceive phase ($\varphi_{\pm}$) to reconstruct electrical conductivity map. For phase-based EPT, the spatial variation of magnitude of transmit field (B_1^+) have to be negligible level. To guarantee this assumption, quadrature body coil (QBC) and/or single channel head transmit coil was recommended[1,2]. In general, separate coils can be used for transmission and reception[3]. However, to use a different coil for reception such as breast imaging, the spatial variation of the magnitude of receive field (B_1^-) also have to be negligible. Here, we propose a coil combine method which minimizes the spatial variation of the combined B_1^-. A zero-order phase ($\varphi_{0,i}$) is selected for each i-th coil such that $\nabla B_1^- / B_1^-$ is minimized. The $\varphi_{0,i}$ was selected from a homogeneous phantom data and in-vivo breast conductivity imaging was performed.

Theory: Using the Helmholtz equation for complex transmit field ($\boldsymbol{B_1^+}$) and receive field ($\boldsymbol{B_1^-}$), individual conductivity map, σ^+ and σ^- can be reconstructed identically as in Eq.1[1]. The average value of σ^+ and σ^- can be decomposed as in Eq.2. When the term **a** and **b** are negligible, Eq.2 can be simplified using transceive phase ($\varphi_{\pm}$) as in Eq.3.

$$\sigma^+ = \mathrm{Im}\left\{\nabla^2 \boldsymbol{B}_1^+ / \boldsymbol{B}_1^+\right\} / \mu_0\omega \qquad \sigma^- = \mathrm{Im}\left\{\nabla^2 \boldsymbol{B}_1^- / \boldsymbol{B}_1^-\right\} / \mu_0\omega \quad (1)$$

$$(\sigma^+ + \sigma^-)/2 = \mathrm{Im}\left\{\underbrace{2\nabla B_1^+ \cdot \nabla e^{i\varphi_+} / \boldsymbol{B}_1^+}_{a} + \nabla^2 e^{i\varphi_+} / e^{i\varphi_+} + \underbrace{2\nabla B_1^- \cdot \nabla e^{i\varphi_-} / \boldsymbol{B}_1^-}_{b} + \nabla^2 e^{i\varphi_-} / e^{i\varphi_-}\right\} / 2\mu_0\omega \quad (2)$$

$$\approx \mathrm{Im}\left\{\nabla^2 e^{i\varphi_+} / e^{i\varphi_+} + \nabla^2 e^{i\varphi_-} / e^{i\varphi_-}\right\} / 2\mu_0\omega = \mathrm{Im}\left\{\nabla^2 e^{i\varphi\pm} / e^{i\varphi\pm}\right\} / 2\mu_0\omega \quad (3)$$

To guarantee the result of phase-based EPT, the term ***b*** is minimized using a novel coil combine method. The term ***a*** is already known to be negligible due to the usage of QBC for transmit[1,2]. Directly acquiring B_1^- is hard for conventional MRI. In this study, a homogeneous reference phantom was used to evaluate $\varphi_{0,i}$. For the phantom, the tissue signal can be regarded as constant so the spatial variation of signal magnitude (S) is only dependent on B_1^+ and B_1^-. Using this characteristic, the spatial variation of S can be decomposed as

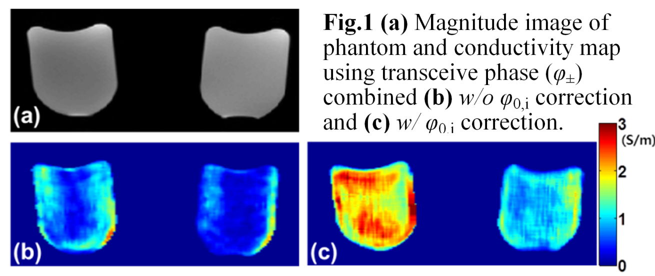

Fig.1 (a) Magnitude image of phantom and conductivity map using transceive phase ($\varphi_{\pm}$) combined **(b)** *w/o* $\varphi_{0,i}$ correction and **(c)** *w/* $\varphi_{0,i}$ correction.

$$\underset{\varphi_0}{\operatorname{argmin}} \left\| \frac{\nabla S}{S} \right\|_2 \approx \underset{\varphi_0}{\operatorname{argmin}} \left\| \frac{\nabla f(B_1^+)}{f(B_1^+)} + \frac{\nabla B_1^-}{B_1^-} \right\|_2 = \underset{\varphi_0}{\operatorname{argmin}} \left\| \frac{\nabla B_1^-}{B_1^-} \right\|_2 \quad (4)$$

where $f(\cdot)$ is B_1^+ related function in MR signal. Therefore, we find $\varphi_{0,i}$ which minimizes the left side of (4) which corresponds to the $\varphi_{0,i}$ that minimizes the right side of (4). Note that the term related to B_1^+ in (4) is constant for varying $\varphi_{0,i}$.

Methods: As shown in Fig 1a, NaCl solutions with 2.0 (Left) and 1.1 (Right) S/m conductivity phantom were used. Using the determined $\varphi_{0,i}$ values from homogeneous phantom, in-vivo breast conductivity imaging was performed under the assumption that the $\varphi_{0,i}$ of phantom is similar to the $\varphi_{0,i}$ of human breast. Phantom and in-vivo imaging from a patient with known malignant breast cancer was performed in a 3T clinical scanner (MR750, GE Healthcare, Waukesha, WI) with a 8-channel breast coil using 2D T2-weighted fast spin echo (FSE) sequence (TR/ TE_{eff}=4420/102ms, voxel size=0.81×1.3×3 mm^3). A modified bilateral filter and mean filter was used for image quality improvement.

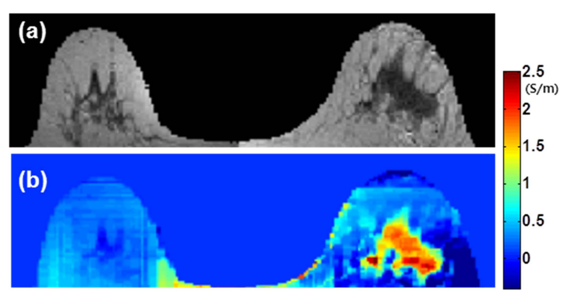

Fig. 2 (a) Magnitude image of breast and resulting conductivity map **(b)** using $\varphi_{\pm}$ combined with determined $\varphi_{0,i}$

Results & Conclusion: As shown in Fig 1b, conductivity reconstruction contained error due to non-negligible the spatial variation of B_1^- when no phasing was used. However, by combining each coil data with $\varphi_{0,i}$, error in the conductivity map reduced (Fig 1c) and the average conductivity values closely corresponded to the measurement value (Table 1). When the determined $\varphi_{0,i}$ was used on patient data, the resulting conductivity value of tumor was increased which closely corresponded to conductivity value from ref.[4]. In conclusion, a coil-combine process was developed which minimizes the spatial variation of B_1^-, and this approach was used to determine conductivity of malignant breast tumor.

Phantom	**Left**	**Right**
Meas.	2.1 S/m	1.0 S/m
w/o $\boldsymbol{\varphi_{0,i}}$	0.62 (±0.41)	0.23 (±0.28)
w/ $\boldsymbol{\varphi_{0,i}}$	2.19 (±0.17)	0.98 (±0.12)
In-vivo	**Fat**	**Tumor**
ref.[4] **at 100MHz**	0.04(at 37°C)	1.4 (at 37°C)
w/ $\boldsymbol{\varphi_{0,i}}$	0.14 (±0.50)	1.50 (±0.47)

Table.1 The resulting conductivity value (mean ± standard deviation) of phantom and in-vivo data

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