



Sub-Sm⁻¹ electromagnetic induction imaging with an unshielded atomic magnetometer

Cite as: Appl. Phys. Lett. **116**, 133501 (2020); <https://doi.org/10.1063/5.0002146>

Submitted: 22 January 2020 . Accepted: 02 March 2020 . Published Online: 31 March 2020

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ABSTRACT

Progress in electromagnetic induction imaging with atomic magnetometers has brought its domain to the edge of the regime useful for biomedical imaging. However, a demonstration of imaging below the required 1 Sm^{-1} level is still missing. In this Letter, we use an ^{87}Rb radio frequency atomic magnetometer operating near room temperature in an unshielded environment to image calibrated solutions mimicking the electric conductivity of live tissues. By combining the recently introduced near-resonant imaging technique with a dual radio frequency coil excitation scheme, we image 5 ml of solutions down to 0.9 Sm^{-1} . We measure a signal-to-noise ratio of 2.7 at 2 MHz for 0.9 Sm^{-1} , increased up to 7.2 with offline averaging. Our work is an improvement of 50 times on previous imaging results and demonstrates the sensitivity and stability in unshielded environments required for imaging biological tissues, in particular for the human heart.

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Recent years have seen a vast increase in the applications of quantum technologies and, in particular, atomic magnetometers¹ to the biomedical field. Notable examples include magnetocardiography^{2–5} and magnetoencephalography.^{4,6,7} Applications for monitoring the reactivity of the nervous system have also been reported.⁸ In all cases, the superior performance of the atomic magnetometers pushes existing technologies and diagnostic methods toward their fundamental limits.

However, mapping the electric conductivity of biological tissues—and in particular, of the human heart—is still an open issue. To date, no diagnostic tool is capable of non-invasively mapping the conductivity of cardiac tissue.⁹ Current investigations require the invasive recording of activation potentials via surgically introduced electrodes. This does not allow direct mapping of conductivity and presents issues due to the inconsistent adhesion of electrodes to the inner surface of the beating heart.¹⁰

Electromagnetic induction imaging—often referred to as magnetic induction tomography¹¹ to highlight its tomographic capabilities—has been proposed as a diagnostic tool for various conditions characterized by a variation or an anomaly in electric conduction.^{12–15} Using this technique, eddy currents are excited in the specimen under investigation by an AC magnetic field (primary field). The response, containing information about the electric conductivity, electric permittivity, and magnetic permeability of the object of interest, is detected via the magnetic field generated by the eddy currents (secondary field).

One of the main limitations of this approach is the limited sensitivity of the magnetic field sensors in use. Therefore, until recently, detection and imaging were limited to relatively large samples,¹⁶ in most cases well above useful volumes for medical applications. This issue was potentially solved by the demonstration of electromagnetic induction imaging with atomic magnetometers.^{17–19} The higher sensitivity of the core sensor paved the path for applications with small volumes of low-conductivity materials.^{20,21} Nevertheless, a direct demonstration of imaging of sub- Sm^{-1} specimens in unshielded environments, as required for applications in the biomedical field,²⁰ is still lacking.

In this Letter, we demonstrate electromagnetic induction imaging of sub- Sm^{-1} calibrated solutions using an ^{87}Rb radio frequency (RF) atomic magnetometer operating in an unshielded environment, near room temperature, without the need of acquiring background or reference images. The performance required for imaging samples mimicking the conductivity of the cardiac tissue^{22,23} ($\sigma_{\text{el}} \sim 0.7 \text{ Sm}^{-1} - 0.9 \text{ Sm}^{-1}$) is obtained by combining near-resonant imaging with dual RF coil driving. By using two RF coils in anti-phase, we enhance the sensitivity of the atomic magnetometer to changes in conductivity, without detrimental effects due to RF-induced power broadening. This represents an alternative to augment the detected phase change by increasing the magnetometer's operational frequency: a larger primary field produces a corresponding increase in the amplitude secondary field generated by the object of interest.^{11,12} To optimize the imaging performance, we minimize the measured phase noise using a redesigned approach to

data collection and programmable near-resonant imaging.²⁰ In this way, we are able to obtain a proof-of-principle demonstration exactly matching the requirements for imaging of cardiac tissue.⁹

The experimental setup is a targeted evolution of a previous realization.²⁰ The atomic magnetometer is operated in an unshielded environment, by optically pumping ⁸⁷Rb atoms in a 25 mm quartz cell (buffer gas: N₂, 40 Torr) to the $|5^2 S_{1/2} F=2, m_F = +2\rangle$ state via a σ^+ -polarized laser beam at 795 nm (80 MHz detuning, 330 μ W) and a collinear DC bias field. This field [B_{bias} , in Fig. 1(a)] is locked to the desired value, using a three-axis flux gate (Bartington MAG690), 46 mm away from the center of the vapor cell^{24,25} and a feedback loop. Passive compensation coils nullify DC stray magnetic fields and gradients. The readout of the atomic spin precession is obtained with a π -polarized probe beam, tuned 1300 MHz above the $|5^2 S_{1/2} F=2\rangle \rightarrow |5^2 P_{3/2} F=3\rangle$ transition Fig. 1(b). The probe's polarization plane is continuously monitored by a polarimeter, whose output is analyzed using a lock-in amplifier (Zurich Instruments HF2LI). Larmor precession of atomic spins is driven by an RF magnetic field [$B_1(\omega_{\text{RF}}) + B_2(\omega_{\text{RF}})$ in Fig. 1(a)]. The RF field is provided by a pair of 100 μ H coils, of diameter 7.8 mm. They are placed at ± 47 mm with respect to the center of the vapor cell, in an arrangement similar to those recently used in other atomic magnetometer setups,^{21,26} implementing the idea of Watson *et al.*²⁷ The coils are antiparallel along the same axis. This creates two AC fields oscillating at the same frequency and with a stable π phase difference (anti-phase). The object to be imaged is placed between the top coil and the sensor, 40 mm above its center [Fig. 1(a)].

The RF magnetic field— $B_1(\omega_{\text{RF}}) + B_2(\omega_{\text{RF}})$, the primary field—induces eddy currents in the object. These produce the secondary magnetic field [$B_{\text{EC}}(\omega_{\text{RF}})$ in Fig. 1(a)] oscillating at the same frequency and containing information about the properties of the object.¹¹ By moving the specimen, held in position by a motorized polylactide (PLA) and Nylon support, with respect to the sensor and measuring its response at the desired position, 2D images are obtained.

One issue with electromagnetic induction imaging and atomic magnetometers is the contrasting requirements for the amplitude of the RF field. Increasing the amplitude of the primary field at the object's position proportionally increases the secondary field to be detected.¹² In contrast, increasing the amplitude of the driving field causes RF-induced power broadening and a consequent reduction in the magnetometer's sensitivity.²⁴ The dual RF coil, driven in parallel by the local oscillator of the lock-in amplifier, allows us to increase the amplitude of the RF field by more than 15 times at the object's level (the driving voltage for the RF coils is increased from $V_{\text{RF}} = 0.6$ V to

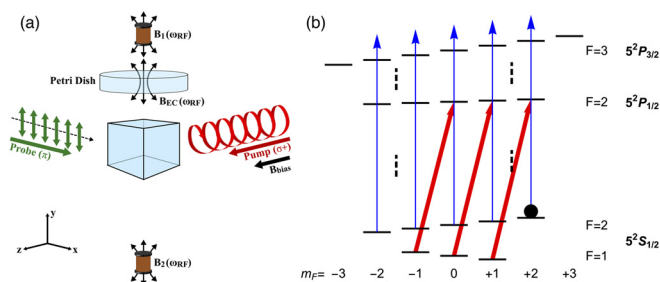


FIG. 1. (a) Schematic of the imaging setup with two RF coils. (b) Energy levels involved in the atomic magnetometer operation.

$V_{\text{RF}} = 10$ V), without broadening the atomic resonances (Fig. 2). In this way, extremely low conductivity specimens can be imaged in unshielded environments, without the need to increase the frequency.²⁰ We did not completely cancel the response of the magnetometer (Fig. 2), instead created a small imbalance in the currents flowing in the two branches of the circuit (split ratio $\sim 51\%:49\%$). By maintaining a residual RF field, we ensure the continuous and consistent operation of the magnetometer in unshielded environments and also in the absence of a field generated by the eddy currents in samples. This contributes to the stability of the measurements, while performing RF sweeps. A different approach to long-term stability for electromagnetic induction imaging was recently demonstrated with a self-oscillating spin maser.²⁸

Calibrated conductivity solutions, contained in Petri dishes (Star Lab CytoOne Dish, diameter 35 mm, height 10 mm), are used. The range of conductivities was chosen to mimic live biological tissues. The size and the volume of the samples were chosen to represent a fraction of the size and volume of an adult human atrium,²⁹ where conductivity anomalies are expected to be found. The specimens were independently assessed using a Jenway 4510 conductivity meter and a Jenway 027 213 Epoxy Bodied probe, at the temperature and time of imaging. We present results using three levels of conductivity: (1) $(9.1 \pm 0.1) \text{ Sm}^{-1}$ at $(20.1 \pm 0.1)^\circ\text{C}$, Reagecon CSKC100M (batch CS100M19H1); (2) $(4.5 \pm 0.1) \text{ Sm}^{-1}$ at $(20.0 \pm 0.1)^\circ\text{C}$, obtained with an NaCl (Sigma-Aldrich Redi-Dri reagent, $\geq 99\%$ purity, 746398–500G) and de-ionized water (RS PRO De-ionized water, 251–3687) solution; and (3) $(0.91 \pm 0.01) \text{ Sm}^{-1}$ at $(20.2 \pm 0.1)^\circ\text{C}$, Reagecon CSKC10M (batch CS10M19B2). Other tests were conducted with NaCl solutions at different concentrations. No noticeable difference in the response of different solutions with the same conductivity is detected. This suggests that the imaging process is independent of the chemical details of the solution and of the ionic carrier. The Petri dishes are filled with 5.0 ± 0.2 ml of solution and then sealed to prevent evaporation. For imaging, we operate the magnetometer at around 2 MHz. This frequency was chosen to allow direct comparison with previous results²⁰ and to highlight the advantages of the configuration presented in this Letter. To

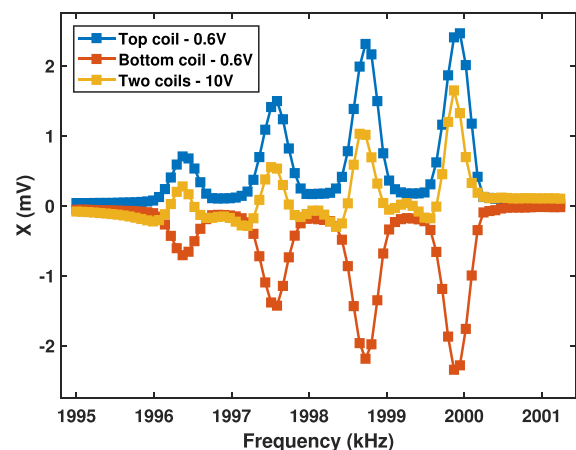


FIG. 2. Comparison of the in-phase (X) response with the top, bottom, and dual coils driving the atomic magnetometer for $V_{\text{RF}} = 0.6$ V, 0.6 V, and 10 V, respectively.

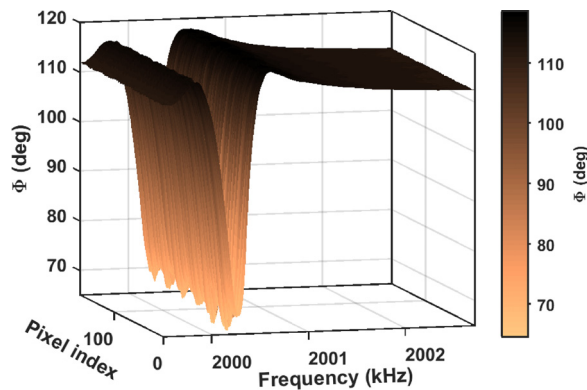


FIG. 3. Phase (Φ) spectra as a function of the RF frequency collected for a typical image. A complete image acquisition is constituted by X, Y, R, and Φ spectra per each pixel of the 2D imaging area.

maximize the amount of the retained information, the system automatically acquires a full RF spectrum at each position of the image (i.e., for each pixel), by sweeping the RF field frequency across the atomic resonance, between 1.995 MHz and 2.001 MHz. Five arrays, containing the swept frequency, the in-phase response (X), the quadrature response (Y), the amplitude of the response (R), and its phase lag (Φ), are collected during the scan. For an $n \times m$ pixel scan and a k -point long frequency scan, five matrices are thus stored, each formed by k columns and $(n \times m)$ rows. As an example, Fig. 3 illustrates the typical raw Φ data obtained by the system. In this case, each pixel is represented by a Φ spectrum of the atomic resonance recorded at that position. In this way, images are obtained for all parameters. The small modulation appearing at the minima of Fig. 3 is attributed to residual magnetic inhomogeneities of the imaging volume, probably caused by the motorized translational stage in use. This approach allows the automatic implementation of near-resonant imaging²⁰ at a programmable detuning that minimizes the impact of magnetic noise in the parameter of interest. In the case of the Φ maps presented here, images can be obtained at the near-resonant frequency exhibiting the minimum slope of the phase response. We found that the optimum performance at 2 MHz was $+0.7$ kHz from the nominal resonance of the system. Multi-frequency imaging is also automatically attainable. In contrast to our previous demonstration,²⁰ here the optimization can be performed exclusively based on image quality without any intervention on the

hardware. In addition, the complete set of raw data is fully retained for further analysis or performance enhancement.

In Fig. 4, we show the Φ maps of three Petri dishes containing 5 ml of calibrated solutions, at 9.1 Sm^{-1} (a), 4.5 Sm^{-1} (b), and 0.91 Sm^{-1} (c). Images in Fig. 4 are obtained by weighted average of 43, 18, and 51 images [for (a), (b), and (c), respectively]. Datafiles were collected consecutively, at random intervals, under the same environmental conditions and with the same settings of the imaging systems. For a given conductivity, the weights are obtained with a normal distribution fit to the set of images (see Fig. 5 for further details). The obtained image is then interpolated using a cubic piecewise algorithm. A nearest-neighbor Gaussian filter (radius 2 pixels) is also applied to aid visualization.

The specimens under investigation are correctly detected, imaged, and located in the scene, within the 4 mm step-size of the translational stage in use. Minor distortions appear in the 0.91 Sm^{-1} sample [Fig. 4(c)]. These are attributed to residual magnetic fluctuations of the background, which are not entirely eliminated by the active stabilization of the bias field and the near-resonant imaging.

This result demonstrates electromagnetic induction imaging with an atomic magnetometer of small volumes of sub- Sm^{-1} specimens. The solutions reproduce the ionic conductivity typical of live tissues, where electrolytes move in the intra- and intercellular spaces.²² Figure 4(c) mimics imaging of the heart by non-invasive electromagnetic induction imaging with atomic magnetometers: the electric conductivity of the healthy cardiac tissue in the MHz range varies^{22,23} between 0.7 Sm^{-1} and 0.9 Sm^{-1} . This represents the background conductivity on which anomalous structures supporting atrial fibrillation, and speculated to be characterized by an increase in conductivity,^{9,30} emerge. The demonstrated performance of our instrument is thus sufficient for their detection. We recall that, given the current lack of diagnostic tools dedicated to atrial fibrillation, quantitative information on its supporting structures is not available.⁹ For this reason, an imaging system capable of mapping the conductivity, even at a low resolution, of the cardiac tissue would be of primary importance. In this view, the results of Fig. 4(c), together with the unshielded and automated operation, allow one to conclude that the feasibility of electromagnetic induction imaging of biological tissues is now fully demonstrated. If required, the spatial resolution could be enhanced via machine learning localization and classification algorithms.³¹

The minimum imaged conductivity is improved by more than 50 times with respect to previous results²⁰ and by a factor 4 with respect to the recently reported detection of low-conductivity solutions in a

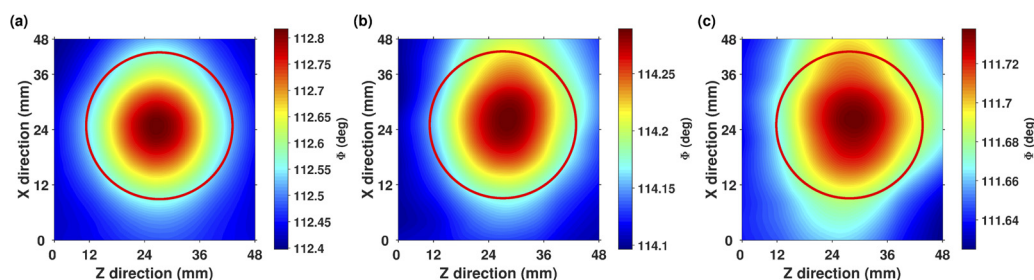


FIG. 4. Imaging of low-conductivity 5 ml calibrated solutions. (a) Phase image of the $(9.1 \pm 0.1) \text{ Sm}^{-1}$ sample. (b) Phase image of the $(4.5 \pm 0.1) \text{ Sm}^{-1}$ sample. (c) Phase image of the $(0.91 \pm 0.01) \text{ Sm}^{-1}$ sample. The red circles mark the position and the extension of the Petri dishes.

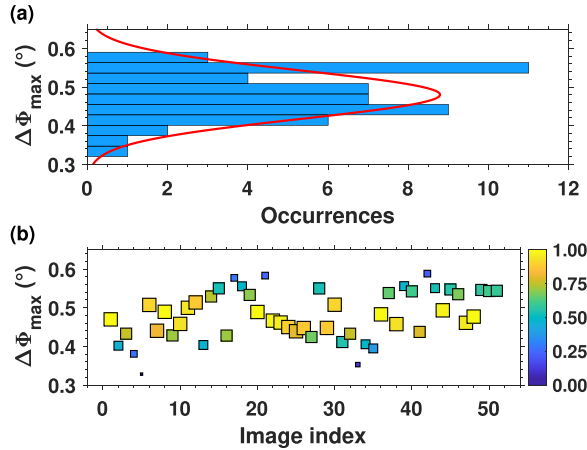


FIG. 5. Stability of sub- Sm^{-1} imaging. (a) Distribution of the maximum recorded phase change ($\Delta\Phi_{\max}$) across 51 images of the 0.91 Sm^{-1} sample. The red line is the fitting normal distribution. (b) Corresponding scatterplot. The size and color are proportional to the weight used for averaging. The colorbar indicates the weights $[0, 1]$ attributed to the images.

shielded environment.²¹ However, when considering the sensitivity of electromagnetic induction imaging instrumentation, the effective volume supporting eddy currents and thus contributing to the generation of the secondary field is also relevant. The larger the volume exhibiting a given conductivity, the bigger the signal produced. By assuming the most favorable conditions—namely, when the skin depth is larger than the thickness of the object and the transverse size of the primary field at the specimen's surface is of the same order of the object's size—one can simply take into consideration its physical volume (V). Following this reasoning, we introduce a figure-of-merit to characterize the absolute conductivity sensitivity of systems performing electromagnetic induction imaging,

$$\mathcal{S} \equiv \sigma_{\text{el}} \cdot V [\text{Sm}^2], \quad (1)$$

where σ_{el} is the electric conductivity of the object of interest. Based on Eq. (1), our record performance with the $\sigma_{\text{el}} = 0.91 \text{ Sm}^{-1}$ sample of $V = 5.0 \text{ ml}$ yields $\mathcal{S} = 4.55 \times 10^{-6} \text{ Sm}^2$, an improvement of more than two orders of magnitude over our previous work²⁰ ($\mathcal{S} = 3.13 \times 10^{-4} \text{ Sm}^2$) and of an order of magnitude with respect to recent detection reports²¹ ($\mathcal{S} \approx 3.0 \times 10^{-5} \text{ Sm}^2$). The $\sigma_{\text{el}} = 4.5 \text{ Sm}^{-1}$ sample corresponds to $\mathcal{S} = 2.25 \times 10^{-5} \text{ Sm}^2$, $\sim 26\%$ better than previous works.

\mathcal{S} also highlights the potential of unshielded electromagnetic induction imaging with an RF atomic magnetometer with respect to other approaches. Our goal here is to compare bare performance with the state-of-the-art of electromagnetic induction imaging. Detection of conductivities as low as 0.5 Sm^{-1} has been reported with standard coils.¹⁶ However, the comparatively large volumes ($2.3 \times 10^3 \text{ ml}$) of the samples give $\mathcal{S} = 1.13 \times 10^{-3} \text{ Sm}^2$, three orders of magnitude worse than the level demonstrated here. In a different context, current nitrogen-vacancy (NV) center magnetometers performing electromagnetic induction imaging³² showed $\mathcal{S} \approx 1.8 \times 10^{-1} \text{ Sm}^2$. However, we note that assessing the smallest detectable object was not the scope of that work,³² of previous ones,^{16,20,21} or of this Letter.

We have also investigated the performance with various images of identical samples acquired across several hours. The single-image signal-to-noise ratio (SNR) is well above unity in all cases. For the 9.1 Sm^{-1} sample, we measured a $\text{SNR} = 13.4$. For the 4.5 Sm^{-1} sample, $\text{SNR} = 8.0$. Finally, the 0.91 Sm^{-1} sample's images exhibit a single-image $\text{SNR} = 2.74$.

We have also explored the direct averaging of the images, to improve the signal-to-noise ratio and readability. This would not be necessary when computer algorithms or machine learning are used.³¹ In this way, long-term performance as well as stability and repeatability of imaging—all critical features for imaging relying on unshielded atomic magnetometers—can be tested and quantified. Averaging several images improves such numbers, as expected: the averaged signal-to-noise ratio, $\langle \text{SNR} \rangle$, is 35.5, 16.2, and 7.23 from the three conductivities used, with an improvement of around 2.6 times.

As an illustration of the stability of the imaging for the 0.91 Sm^{-1} sample, we show the distribution of the maximum phase difference measured across an image $\Delta\Phi_{\max}$ for 51 unfiltered and non-interpolated repetitions [Fig. 5(a)],

$$\Delta\Phi_{\max} = (\Phi_{\max} - \Phi_{\min})_{\text{image}}. \quad (2)$$

This allows one to immediately identify issues with the instrument or of the imaging process. The dataset is well-fitted by a normal distribution, with mean value $\langle \Delta\Phi_{\max} \rangle = 0.48^\circ$ with 95% confidence interval $[0.46^\circ, 0.50^\circ]_{0.95}$ and standard deviation $\sigma = 0.06^\circ$ ($[0.05^\circ, 0.08^\circ]_{0.95}$). In Fig. 5(b), the effect of the weighted average for Fig. 4(c) is illustrated. Once the distribution of the image dataset is identified, it is used for weighting the contribution of the single images to the average. Although the details of the mechanisms leading to atrial fibrillation are yet to be confirmed,⁹ one can claim that the phase sensitivity and stability demonstrated in Fig. 5 are sufficient to detect small anomalies of conductivity in the heart. For example, in the case of a factor 2 increase in conductivity in a 5 ml volume, we expect to detect a phase change around 0.2° with the instrument presented here. This is safely above the detection threshold and intrinsic variability of our imaging system ($[0.05^\circ, 0.08^\circ]_{0.95}$).

In conclusion, we have presented electromagnetic induction images of calibrated solutions of conductivities below sub- Sm^{-1} with an unshielded RF atomic magnetometer. The simultaneous use of near-resonance imaging and of a dual RF coil excitation provided the necessary sensitivity and stability to allow us to image 5 ml of 0.91 Sm^{-1} solution, in magnetically unshielded environments. Our atomic magnetometer-based imaging system operates with sufficient stability to guarantee consistent imaging performance across several days. Based on our results, the feasibility of practical use of electromagnetic induction imaging with atomic magnetometers in the biomedical field is finally fully demonstrated. In particular, our instrument meets the requirements for direct, non-invasive imaging of the cardiac tissue—for example, for the diagnosis of atrial fibrillation⁹—which now appears to be technically feasible.

This work was supported by the UK Quantum Technology Hub in Sensing and Metrology, Engineering and Physical Sciences Research Council (EPSRC) (No. EP/M013294/1). We thank Dr. Helena Wong and Professor Andrea Sella (UCL Department of Chemistry) for the loan of the conductivity meter and guidance on its operation.

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