Single-Beam Double-Pass Miniaturized Atomic Magnetometer for Biomagnetic Imaging Systems

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Abstract—Miniaturized atomic magnetometers (AMs), particularly spin-exchange relaxation-free (SERF) AMs, have been emerging in clinical imaging applications such as magnetocardiography (MCG) and magnetoencephalography (MEG). Miniaturization, portability, and low cost are primary development targets for biomagnetic imaging technologies, as well as high sensitivity, temporal, and spatial resolution. In this article, we propose a low-cost solution for a biomagnetic imaging system based on AMs, in which one laser source is used for a multichannel AM sensor array. A novel design is demonstrated for a miniaturized SERF AM, consisting of a single-beam double-pass configuration based on an optical fiber circulator. The effects of temperature and laser power on the zero-field magnetic resonance linewidth are characterized, and the experimental results show that the present design achieves better performance, compared with a traditional single-beam single-pass configuration. The magnetic field noise spectrum shows that such single-beam double-pass miniaturized AMs reach a sensitivity of approximately $86 \text{ fT/Hz}^{1/2}$ at a frequency of $10 \text{ Hz}$ when operating in the single-channel mode and a sensitivity of approximately $40 \text{ fT/Hz}^{1/2}$ when operating in magnetic differential mode with two sensors sharing one laser source and a 6-mm baseline. A typical adult MCG signal with all the visible P, QRS-complex, and T waves is recorded by the magnetometer in the single-channel mode. This design is especially suitable for AMs operating in arrays as a basic building element for low-cost biomagnetic imaging systems.

Index Terms—Atomic magnetometer (AM), biomagnetic imaging, optical fiber circulator, spin-exchange relaxation-free (SERF).

I. INTRODUCTION

MAGNETIC fields produced by biological organisms have recently received much attention owing to the valuable information contained in these fields regarding underlying physiological processes and their pathologies [1], [2], [3]. Over the past 40 years, these weak biomagnetic fields have generally been detected by superconducting quantum interference devices (SQUIDs) [4], [5], [6], which are commercially available but require high-cost cryogenic condition.

Rapid progress has been made in the field of atomic magnetometers (AMs), which have recently achieved sensitivities comparable to those of SQUIDs [7], [8]. Among the AMs with various principles, the spin-exchange relaxation-free (SERF) magnetometer has the maximum sensitivity for detecting weak magnetic fields [8], [9], [10]. In contrast to SQUIDs, AMs can simultaneously realize miniaturization, low power consumption, excellent temporal and spatial resolution, portability, and ultrahigh sensitivity [11], [12], [13].

Miniaturization, portability, and low cost are primary development targets for biomagnetic imaging technologies based on AMs or SQUIDs. The National Institute of Standards and Technology (NIST) demonstrated a chip-scale AM using a microelectromechanical system approach in 2004 [14], [15]. QuSpin, Inc., presented a commercial, compact, high-performance zero-field AM (QuSpin QZFM) in 2013 [16], and related biomagnetic imaging tests for magnetoencephalography (MEG) and magnetocardiography (MCG) are underway [17], [18]. In biomagnetic imaging applications, multiple miniaturized AM sensors are needed to form a magnetic field measurement array. The sensors are usually separate and independent, with each sensor consisting of a vertical-cavity surface-emitting laser (VCSEL), a vapor cell with alkali metal, and a photodetector (PD) [14], [19]. When operating in array mode, differential mode noise is introduced owing to the nonuniformity of VCSEL laser sources, vapor cells, and PDs.
which deteriorates the imaging resolution of the magnetic field. Differential mode noises induced by the inconsistency between different sensors tend to cause magnetic image distortion, which becomes severe as the number of sensors increases. Moreover, the integration of a laser source and PD into a sensor head not only increases the sensor size and the cost per channel but also inevitably introduces electromagnetic noise, which increases the noise floor of an AM and reduces its sensitivity.

In this article, we propose a novel AM design for biomagnetic imaging systems, consisting of a double-pass optical path configuration based on a mirror and an optical fiber circulator. Without a VCSEL laser source and PD integrated in the sensor head, each sensor channel shares one common laser source. When arrayed to build a multichannel biomagnetic imaging system, these sensors may have good uniformity of common-mode noise, which allows high sensitivity and spatial resolution for imaging biomagnetic fields.

II. PRINCIPLE

A classical AM consists of three crucial elements: a pumping laser, an atomic ensemble (usually consisting of alkali metal), and a detection system [20]. The atomic ensemble is optically pumped by the circularly polarized pumping laser, resulting in a high steady-state spin polarization along the direction of laser beam propagation. In the presence of a transverse external magnetic field, the spin polarization of the atomic ensemble precesses along the magnetic field and finally reaches a new steady-state along a new direction between the longitudinal and transverse axes. Then, the longitudinal component of the spin polarization is measured by the detection system, and thus, the magnetic field is also obtained [11], [21], [22], [23]. The atomic shot-noise-limited sensitivity of an AM is determined by the lifetime of the spin coherent precession as follows [7]:

\[ \delta B = \frac{1}{\gamma \sqrt{N} \cdot \tau \cdot t_{\text{int}}} \]  

(1)

where \( \gamma \) is the alkali metal gyromagnetic ratio, \( N \) is the number of atoms being polarized and detected, \( \tau \) is the spin coherence lifetime, and \( t_{\text{int}} \) is the integration time.

While ensuring a long lifetime for the atomic spin coherence precession, any perturbation breaking this coherence can be detected sensitively [24], [25]. However, this spin coherence is disrupted by many undesired factors, such as an inhomogeneous magnetic field around the atoms, atomic collisions with the walls, or other gases in the vapor cell. Herein, the dominating factor is the spin-exchange process [25], [26], [27].

When the spin-exchange collision rate of the atoms is much larger than the precession frequency (Larmor frequency), the relaxation due to spin exchange is greatly suppressed, corresponding to the SERF regime demonstrated by Allred et al. [7] and Dang et al. [10].

In the SERF regime, a coordinate system is defined where the laser beam propagates along \( \hat{z} \) and the external magnetic field along \( \hat{y} \). The behavior of the spin polarization \( S \) is described as [20], [26], [28]

\[ \frac{dS}{dt} = \frac{1}{q} \left[ \gamma e B \times S + R_{\text{op}} \left( \frac{1}{2} \hat{z} \cdot \vec{s} - S \right) - R_{\text{rel}} S \right] \]  

(2)

where \( \gamma e \) is the electron gyromagnetic ratio, \( B \) is the external magnetic field, \( R_{\text{op}} \) is the optical pumping rate, \( s \) is the photon polarization, \( R_{\text{rel}} \) is the sum of all the relaxation rates except the optical pumping rate, and \( q \) is the nuclear slowing-down factor, which is equal to 6 for a small atomic polarization [7].

Operated in a high-density condition, the linewidth of the atomic transition becomes significantly broader than that of the laser. \( R_{\text{op}} \) is approximately equal to the average rate at which a nonpolarized alkali metal atom absorbs a photon of the pumping laser, shown as [29], [30], [31], [32]

\[ R_{\text{op}} \approx r e f D_{1} \phi \frac{\Gamma/2}{(v - \nu_{0})^{2} + (\Gamma/2)^{2}} \]  

(3)

where \( r_{e} \) is the classical electron radius, \( c \) is the velocity of laser, \( f D_{1} \) is the oscillation intensity of the \( ^{87}\text{Rb} \) D1 line, \( \phi \) is the photon flux, \( \Gamma \) is the pressure broadening of alkali metal atoms with the quenching gas and buffer gas, \( v \) is the pumping laser center frequency, and \( \nu_{0} \) is the center frequency of the D1 line absorption. \( R_{\text{op}} \) is proportional to the pumping laser power and determines the atomic spin polarization with the relaxation rate and the external magnetic field simultaneously.

The atomic spin polarization along the laser beam is approximately a Lorentzian curve in the presence of a dc magnetic field \( B_{y} \) transverse to the laser beam, given by

\[ S_{z - dc}(t) = S_{0} \frac{1}{1 + (\gamma B_{y} \tau)^{2}} \]  

(4)

where \( S_{0} = R_{\text{op}}/(R_{\text{op}} + R_{\text{rel}}) \) is the effective polarization, \( \tau = q/(R_{\text{op}} + R_{\text{rel}}) \) is the spin coherent lifetime, and \( \gamma = \gamma e / q \) is the gyromagnetic ratio. With the addition of an oscillating magnetic field \( B_{\text{mod}} \cos \omega t \) applied along \( \hat{y} \), the first-harmonic component of the polarization is approximately given by [20], [28], [33]

\[ S_{z}(t) = S_{0} J_{0} \left( \frac{\gamma B_{\text{mod}}}{\omega} \right) J_{1} \left( \frac{\gamma B_{\text{mod}}}{\omega} \right) \left( S_{z}(0) - \frac{\gamma B_{y} \tau}{1 + (\gamma B_{y} \tau)^{2}} \right) \sin \omega t \]  

(5)

where \( J_{n}(n = 1, 2) \) are the Bessel functions of the first kind.

The power \( I \) of the transmitted laser beam through the atoms is proportional to the spin polarization, shown as [26]

\[ \frac{dI}{dz} = -n \sigma(v) I(z) (1 - 2 \langle S_{z} \rangle) \]  

(6)

\[ I_{\text{out}} = I_{0} e^{-(1 - 2 \langle S_{z} \rangle) OD} \]  

(7)

where \( n \) is the atom density, \( \sigma(v) \) is an absorption coefficient related to the laser frequency \( v \), and \( I_{0} \) and \( I_{\text{out}} \) are the incident and transmitted powers of the laser beams, respectively. OD = \( n \sigma(v) l \) is the optical depth (OD), where \( l \) is the interaction length between the laser beam and the atoms.

Equations (2)–(7) constitute control equations of an SERF AM system with magnetic field modulation and lock-in detection. Fig. 1 shows a typical resonance curve in our experiment.
SERF AMs is shown in Fig. 2. The SERF linewidth (FWHM) is approximately 34 nT for a temperature of 140 °C and an incident laser power of 80 µW.

III. EXPERIMENTAL SETUP

The schematic design of a biomagnetic imaging system with SERF AMs is shown in Fig. 2. The biomagnetic imaging system consists of a laser source, multichannel fiber power beam-splitter, a set of single-mode polarization-maintaining fibers, PD rack, and control and signal-processing unit. One 795-nm external cavity diode laser (ECDL, DL Pro, TOPTICA) is used for the laser pumping and detecting source for multiple AM sensor heads [34], [35]. The laser is tuned to the resonance transition of the 87Rb D1 line with a red detuning of approximately 12 GHz. After being coupled into a fiber coupler and distributed into multiple channels by the fiber power beam-splitter, the laser beam exiting each channel enters a sensor head for laser–atom interaction and comes out for detection by an optical fiber circulator and an AM sensor with a single-beam double-pass configuration [12], as shown in the inset in Fig. 2. The optical fiber circulator we used is a three-port optical fiber device designed such that the laser beam entering port 1 is emitted from port 2. However, if some of the emitted beams are reflected back to the circulator via port 2, they do not come out of port 1 but instead exit from port 3.

The input laser for the sensor head comes from port 2 of the optical fiber circulator and is collimated with a titanium collimator (60FC-4-M10-02, Schafter + Kirchhoff). After passing through quarter-wave plate and vapor cell, the laser is retro-reflected by a mirror, double-passed through the vapor cell, and coupled into the same collimator for detection. The detection laser is finally collected by a PD (PDA36A, THORLABS) in the PD rack via port 3 of the optical fiber circulator. Due to the presence of the quarter-wave plate, the polarization direction of the reflected laser is rotated approximately 90° relative to the incident laser, which is why the optical fiber circulator is designed to propagate the laser in both the axes. A set of PCB solenoid coils are placed to produce a modulating and compensating magnetic field along the direction perpendicular to the laser beam axis, which corresponds to the sensor measurement axis. The vapor cell is a 4 × 4 × 4 mm cubic cell (wall thickness is 0.5 mm), containing a droplet of Rb, 100 torr of quenching gas N2, and 700 torr of buffer gas 4He. The vapor cell is placed in a miniaturized oven fixed in the sensor head and is heated to approximately 150 °C by two heating films with a 100-kHz ac heating current. The distance from the vapor cell center to the outside of the sensor is approximately 8 mm. The oven is made by polytetrafluoroethylene (PTFE) and is supported by an aluminum frame, which also holds the mirror and the collimator to maintain structural stability in the retro-reflected configuration. The magnetic noise induced by the aluminum is ignorable in this part of the sensor after a measurement with a commercial AM (QZFM2.0, QuSpin). During assembly of the sensor head, the position of the mirror is carefully adjusted by a precision fiber optic alignment stage to ensure that the laser is coupled back into the optical fiber through the collimator.

All the components were carefully designed with non-magnetic materials, including wave plates, atomic vapor cells, heater films, mirrors, coils, and supporting structures. The fibers used in the system are all single-mode polarization-maintaining fibers.

IV. RESULTS AND DISCUSSION

In our experiments, a zero-field resonance signal is provided by a lock-in amplifier (LIA, MFLI, Zurich Instruments) and then is sent to a digital proportional–integral–derivative (PID) controller for a closed-loop locking. In closed-loop mode, the signal from the PID controller is fed back to the PCB solenoid coils for real-time signal compensation, with the added modulation signal. The maximum measurement sensitivity is realized by locking the zero-crossing of the magnetic resonance signal.

The double-pass magnetometer sensor is placed in the center of a totally enclosed magnetic shield tube with a shielding factor larger than 10^5, which provides a zero-field environment, where the remnant magnetic field is much smaller than the linewidth of the zero-field resonance curve.

When a magnetic field orthogonal to the pumping axis is scanned in terms of amplitude near a zero magnetic field, the transparency of the atomic cell exhibits a Lorentzian curve, as shown in Fig. 1. The blue Lorentzian curve measured with the magnetometer has a linewidth [fullwidth at half-maximum (FWHM)] of approximately 34 nT. A small modulation in the magnetic field at approximately 1 kHz is applied by the internal PCB coils. With a phase-sensitive LIA, the PD output is demodulated to produce a dispersion curve, as shown by the red line. The amplitude and linewidth of the absorption and dispersion curve determine the sensitivity of the magnetometer, with the latter being related to the relaxation rate of atoms from their pumped state via physical processes, where spin-exchange collisions provide the dominant effect. By heating the atoms, the AM can work in the SERF regime, resulting in a narrow linewidth resonance.

Without the optical fiber circulator and mirror, the laser in the single-pass sensor passes through the cell once and gets absorbed directly by a PD integrated in the sensor.
head. Compared with the traditional single-pass configuration [16], the present retro-reflected configuration has a longer interaction OD between the laser and atoms because the laser passes through the cell twice. We experimentally measured and compared the performance difference between two configurations in terms of the linewidth of the SERF resonance curve, considering effects of the cell temperature and laser power.

Fig. 3 shows the effect of cell temperature on the magnetic resonant linewidth for two configurations. The power of the incident laser is maintained at 40 µW for both the cases. As the cell temperature rises above 90 °C, spin-exchange relaxation is suppressed, and the linewidth of the resonance curve decreases for both the configurations. For the double-pass configuration, the resonance linewidth tends to have a minimum of approximately 36 nT at temperature close to 140 °C. As the cell temperature increases above 140 °C, the resonance linewidth increases owing to the influence of spin destruction relaxation [7], [8]. This finding confirms that the AM is working in the SERF regime. As shown in Fig. 3, the resonance linewidth of the double-pass sensor is much narrower than that of the single-pass sensor, since the laser interacts with the atoms longer. The data shown here are the averages of triplicate measurements performed under the same experimental conditions, and the corresponding errors are also shown in the figure.

Fig. 4 shows the effect of pumping laser power on the magnetic resonance linewidth for two configurations. In the experiment, the power of the pumping laser was varied from 20 to 200 µW, and the linewidth of the resonance curve was measured while the cell temperature was maintained...
at 140 °C. The pumping power of the sensor channel was measured at the fiber output port before the collimator using another channel, which was calibrated to ensure the output power ratio between the two channels. The signal-to-noise ratio becomes too low to measure the linewidth when the optical power is below 20 µW. The resonance linewidths tend to decrease with increasing laser power for low laser powers, reaching a minimum of approximately 34 nT at 80 µW for the double-pass configuration. The linewidths then increase with laser power for both the configurations. The atomic polarization is induced by the rates balance between the atomic relaxation rate and the optical pumping. For a low pumping condition with weak laser power, the optical pumping rate is small compared with the relaxation rate, and the polarization increases as the laser power rises. Meanwhile, for a high pumping condition, most of the atoms are pumped and polarized in a saturation condition, and thus cannot be in the SERF regime. In this case, the resonance linewidth rises with increasing pumping laser power. Because the laser passes through the cell twice in the double-pass configuration and its pumping time and polarization efficiency are higher, the linewidths are lower than those in the single-pass configuration for a given laser power. Similar to our previous experiment, the data shown herein are the averages of triplicate measurements performed under the same experimental conditions.

In Figs. 3 and 4, the magnetic resonant linewidths in the double-pass configuration are smaller than those in the single-pass configuration because of the larger OD in the former. The double-pass configuration cannot reduce the linewidth of atomic spin polarization by increasing the spin coherent lifetime, but influences the absorption detection process by increasing the interaction length of the laser beam and atoms, i.e., the OD of the laser beam transmission, as in (7). A simulation is performed according to (4) and (7), as shown in Fig. 5. The FWHM of the atomic spin polarization from (4) is calculated to be 48.4 nT with a Lorentzian profile. Considering the transmitted beam absorption with a definite OD from (7), the FWHM in the single-pass case is approximately 38.8 nT, while that in the double-pass case decreases to approximately 31.7 nT when the OD of the laser beam transmission doubles. This method is suitable for the miniaturized vapor cell because the large cell needs a higher laser intensity for an effective signal-to-noise ratio, which may broaden the response linewidth.

Frequency responses of the sensor are evaluated by applying a series of calibration sinusoidal magnetic fields with a constant calibration peak at different frequencies along the sensitive axis of the magnetometer sensor, as shown in Fig. 6. The measured −3 dB bandwidth of the sensor in open-loop mode is approximately 86 Hz. The bandwidth in closed-loop mode is expanded to 350 Hz.

A proof-of-concept system with a two-channel magnetometer array is characterized for its magnetic field measurement sensitivity. The two sensors are placed in the magnetic shield with a distance of 6 mm after being adjusted consistently with a temperature of 140 °C and a laser power of 100 µW. The coefficients of two channels’ response to magnetic field are calibrated with a set of calibration Helmholtz coils, which have been calibrated with the QuSpin AM. Sensors’ voltage
outputs can be transformed into magnetic field intensity in the magnetic field modulation axis. The magnetometer response is linearly fit around 0 nT, with the nonlinear error of less than 1%. Then we shut off the currents of calibration Helmholtz coils and observe noise in two different channels with a sample frequency at 10 kHz. The corresponding magnetic field noise is transformed from the experimental voltage noise from AM response.

Fig. 7 shows the time-domain measurement noises of the two channels, \( B_1 \) and \( B_2 \), represented with red and blue lines, respectively, as well as their differential \( B_1 - B_2 \) represented with green line. It is obvious that the differential signal has a smaller noise level under 4 pT, compared with those single-channel signals. In our case, two sensors share the same laser source and have better uniformity of common-mode noise. And systematic magnetic field noises induced by fluctuations of laser power and polarization can be suppressed, compared with the single-channel operation mode.

Fig. 8 shows the magnetic field noise spectrums normalized by dividing their corresponding frequency responses for characterizing the sensitivity of the two-channel system. The peak at a frequency of 50 Hz that exists in the figure corresponds to the utility frequency noise. Operating in the single-channel mode, both the channels are measured to be of the sensitivities of 86 fT/Hz\(^{1/2}\) at 10 Hz. Their differential signal is measured to be of the sensitivity of approximately 40 fT/Hz\(^{1/2}\) at 10 Hz after the suppression of the common-mode noise with a 6-mm baseline, which is half of that of single sensors.

Table I lists various sources of noise for one single-channel sensor in the system, as well as the corresponding estimation of noise magnitudes at a frequency of 10 Hz. Magnetic noise is commonly generated by any noise sources located within the sensitive frequency band, including fluctuating noise from the ambient magnetic field, magnetic noise from the circuits in the sensor, electrical noise in the detecting and signal-processing circuits, and so on. In the magnetic shield tube, the residual magnetic field background is under 10 fT/Hz\(^{1/2}\) measured with the QuSpin AM. The probe waist and the length of the cell containing atoms are both approximately 3 mm. The cell is heated at a temperature of 140 °C where the atomic density is approximately \( n = 1 \times 10^{14} \) cm\(^{-3}\).

With the experimental parameters, the spin-projection noise is calculated to be 0.9 fT/Hz\(^{1/2}\) and photon shot noise to be 1.1 fT/Hz\(^{1/2}\). The magnetic noise induced by the heating circuits is calculated with the measured current noise of the heating circuits and the response relationship between the magnetic field and the driving current in the heating film, which is measured using the QuSpin AM. The technical noise in the current SERF-AM setup is evaluated to be 68.7 fT/Hz\(^{1/2}\) by measuring the laser noise induced by fluctuations of the optical intensity and polarization rotation and the noise of the detection circuits [36]. These are the primary factor for limiting the sensitivity of a single AM sensor in the present system. As demonstrated above, the differential detection mode in a multichannel system is helpful for suppressing this kind of noise. Further performance improvement of the SERF-AM sensors can be made by optimizing the heater design, the stabilization method of the laser parameters, and the protection method of the optical fibers, as well as the signal detection system.

To measure the biological magnetic field, 2-m-long polarization-maintaining fibers are used to send the laser beam from an optical table into the sensor heads in another large semi-enclosed magnetic shield tube with a set of three-axis nulling coils and gradient compensating coils inside, as shown in Fig. 7.

**Table I**

| Type          | Source                      | Level (fT/Hz\(^{1/2}\)) |
|---------------|-----------------------------|--------------------------|
| Magnetic      | Residual magnetic field     | <10                      |
|               | Noise from cell heaters     | 10.6                     |
| Quantum       | Spin-projection noise       | 0.9                      |
|               | Photon shot noise           | 1.1                      |
| Technical     | Laser noise & current noise | 68.7                     |

Fig. 7. Time-domain measurement noises of the two channels and their differential mode. Noises of channels 1 and 2 are represented with red and blue lines, respectively, and their differential mode with green line.

Fig. 8. Magnetic field noise spectrum density of the zero-field resonance signal. The sensitivity of this SERF AM is measured to be approximately 86 fT/Hz\(^{1/2}\) and the differential system to be approximately 40 fT/Hz\(^{1/2}\) at 10 Hz.
in Fig. 9. The noise of the residual magnetic field background is approximately 14.8 fT/Hz$^{1/2}$ as measured with the QuSpin AM. A typical MCG measurement was demonstrated using the single-channel mode magnetometer. The sensor head is fixed on the chest skin for improving the signal-to-noise ratio of MCG as much as possible. MCG peaks from the heartbeat of the subject are observed periodically in time, from the sensor working in the single-channel mode, as shown in Fig. 10(a). Then, with two 0.1-Hz bandwidth bandstop filters, centered at 50 Hz and 1 kHz, respectively, an adult MCG signal with all the visible P, QRS-complex, and T waves is recorded with a 10-kHz sample frequency in Fig. 10(b).

During the measurement, the adult subject tried to keep his breathing stable for preventing the position fluctuation of the sensor. The magnitude of the QRS-complex is approximately 120 pT. This experiment demonstrates the ability of our SERF AMs for the MCG measurement system. After further optimization in magnetic field noise, especially the laser noise and cell heaters’ noise, our system has the potential for more biomagnetic imaging applications.

V. CONCLUSION

In summary, we have proposed a new design for a biomagnetic imaging system, based on closed-loop SERF AMs with a single-beam, double-pass configuration. A double-pass SERF sensor head with an all-optical-fiber structure has been experimentally demonstrated and characterized in comparison to the traditional single-pass configuration. The effects of cell temperature and laser power on the zero-field magnetic resonance linewidth were investigated. The typical resonance linewidth was measured and calculated to be approximately 34 nT at a laser power of 80 µW and a cell temperature of 140 °C. The magnetic field noise spectrum of the closed-loop SERF magnetometer shows that the sensitivity of the magnetic field reaches approximately 86 fT/Hz$^{1/2}$ and the sensitivity of differential system to be approximately 40 fT/Hz$^{1/2}$, both at 10 Hz. A typical adult MCG signal with all the visible P, QRS-complex, and T waves is recorded by the magnetometer in the single-channel mode.

The experimental results show that the double-pass configuration can achieve better performance compared with the traditional single-pass configuration. What is more, the all-optical-fiber design allows for a small, compact sensor head and enables a sensor array layout with high spatial density, which is important for biological magnetic field measurements, such as MEG and MCG.

This design also presents a low-cost solution for biomagnetic field imaging applications, with one laser source used for a multichannel AM sensor array. The use of one laser source for multiple sensors is helpful for suppressing common-mode noise, improving the sensitivity of the sensor, and enhancing the flexibility of the biomagnetic field image. Future work will focus on improving the sensitivity of the AM sensor and developing an imaging system based on a multichannel sensor array for biological applications.

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