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Optically pumped magnetic field sensors for magnetoencephalography and ultra-low field tomography

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Magnetoencephalography (MEG) is a technology for recording brain activity, characterized by high temporal resolution, and allowing to study fast neural processes without violating the integrity of tissues. Traditionally, in MEG tasks, SQUID sensors, collected in arrays, are used [1]. Despite the high sensitivity and compactness of the sensors themselves, these systems have limitations: the need to place SQUID sensors in a common Dewar vessel excludes the possibility of adapting the array to the anatomical features of the object and leads to a loss of the signal-to-noise ratio and spatial resolution. The stationarity and high cost of these systems are also obstacles to their widespread use.

Magnetic resonance imaging (MRI) is based on a different principle: a resonance signal is obtained by polarizing protons in a high field, followed by recording their free precession with induction coils. Since the 1990s, there has been a steady interest in recording MRI signals in an ultralow field (ULF) [2]. A serious argument in favor of the development of ULF MRI methods is the potential compatibility of the requirements for the parameters of the ULF MRI and MEG detecting systems – the latter are characterized by a sensitivity in units of fT/√Hz with a response speed of the order of 200-500 Hz. Registration of ULF MRI signals requires an expansion of the frequency band (e.g., in a 0.1 mT field - up to 4.2 kHz). At the same time, the requirements for the dynamic range of such a system are becoming more stringent, and especially for its ability to withstand pulsed fields used in ULF MRI.

An alternative to SQUID sensors is an optically pumped quantum magnetometer (OPM) based on optically detectable magnetic resonance (MR) in alkali metal atoms. The SERF magnetometers operating in the mode of suppressing the spin-exchange broadening of the MR line in zero field [3] demonstrate a record sensitivity among OPMs at the level of tenths of fT/√Hz. Their sensing element can have a volume of less than 1 cm^3, which makes it possible to place the sensors in the immediate vicinity of the object. Unfortunately, the spatial uniformity of the field required for MEG systems based on SERFs can only be ensured in magnetically shielded rooms. The use of SERFs in ULF MRI systems is possible only with the use of special circuits that transfer the MRI signal to the zero field region [4].

The aforementioned problems are prompting the development of highly sensitive magnetometers that would not require zero magnetic fields or ultra-low temperatures. These properties are characteristic of nonzero-field OPMs which demonstrated a sensitivity of 1.6 fT/√Hz almost 25 years ago [5]. However, this sensitivity was achieved mostly due to the large volume V of the cell (~1800 cm^3). A decrease in V results in sensitivity lost proportional to $\sqrt{V}$ [6]. An attempt to compensate it by increasing the atomic number density leads to an increase in the spin-exchange relaxation rate, which limits the sensitivity. Fortunately, in [7] the opposite effect was described (and later used in [8, 9]) – a decrease in the spin-exchange relaxation rate in in a "stretched state" which can be obtained at high laser intensities.

On this basis, we tried to develop a nonzero-field Cs sensor for MEG – ULF MRI systems. All our schemes were characterized by the following advantages: *i*) tuning the pumping beam into resonance with the $F = 3 \leftrightarrow F' = 3,4$ transition makes it possible to increase the signal by several times [9]; *ii*) the operation of the sensor in M_X mode provides a high response speed unattainable in the M_Z scheme; *iii*) the use of transverse linearly polarized beam for signal detection makes it possible to further increase the sensitivity, and *iv*) the transfer of the signal frequency to tens of kilohertz allows, in comparison with SERF schemes, to significantly reduce the effect of low-frequency laser flicker noise.

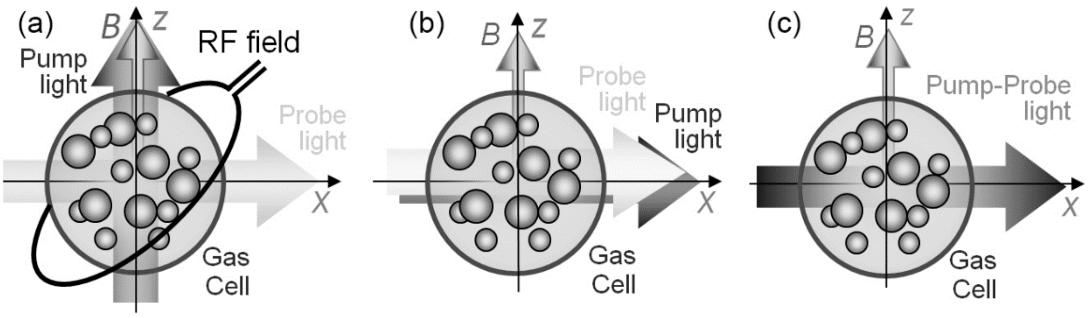

Fig. 1 (a) "classical" scheme; (b) Bell-Bloom scheme; (c) single-beam Bell-Bloom scheme.

As a basis for our nonzero field sensor, we used a slightly modified two-beam "classical" M_X scheme (Fig.1a). The ultimate sensitivity of the sensor in cell volume of 0.5 cm^3 was estimated (using the criteria of the ratio of the MR signal slope to the shot noise spectral density) to be better than 5 fT/√Hz, and the actual sensitivity measured directly in a gradiometric scheme reaches 18 fT/√Hz [10].

The next step was the transition to the Bell-Bloom scheme (Fig.1b), which eliminated the RF field from the sensor circuit [11]. As it turned out, the Bell-Bloom scheme with sinusoidal modulation of the pump radiation polarization does not allow achieving the same degree of squeezing the MR width as the "classical" scheme. Nevertheless, it allows obtaining a sensitivity of ~10 fT/√Hz at a speed of ~1 kHz, and, therefore, is capable of providing a sensitivity of ~40 fT/√Hz in a bandwidth of ~4 kHz, and can be used in ULF MRI schemes. On the other hand, in applications where high speed is not needed, the maximum line narrowing can be obtained using pulsed pumping [12].

The last step to date has been the transition from a two-beam Bell-Bloom scheme to a single-beam one (Fig.1c). As shown in [12], pumping in the Bell-Bloom scheme should be switched on at the moments when the collective spin of the system is directed along the pumping beam; on the contrary, optimum detection efficiency is achieved when the spin is oriented perpendicular to the detection beam. This allowed us to combine two beams into one, and to separate in time (within one precession period) pumping and detection without loss of sensitivity [13]. Thus, we have achieved the ultimate sensitivity better than 10 fT/√Hz in the same cell in a single-beam scheme. At present, we are investigating the applicability to this scheme of pulsed low-duty-cycle optical pumping, as well as the possibility of implementing a free spin precession sensor on its basis.

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