A Monolithic 3D Magnetic Sensor in 65nm CMOS with <10µT Rms Noise and 14.8µW Power

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Magnetic sensors have become increasingly ubiquitous as they constitute an integral part of several fast-growing sectors such as automotive, navigation, robotics, medical devices and consumer electronics [1]-[6]. Due to their compatibility with the standard CMOS process, Hall magnetic sensors are widely used. However, one of the key challenges of CMOS-based Hall sensors is their relatively low sensitivity compared to the low Hall coefficient of Si. For better sensitivity, Hall sensors are biased at higher current levels which hinders their widespread use in low-power bioelectronics and other power-constrained applications. Another challenge is the difficulty in implementing high-sensitivity vertical Hall elements in planar CMOS processes for 3D sensing. This is often overcome by employing ferromagnetic materials that require additional and expensive steps during fabrication, thus increasing the cost [2][3].

Our sensor is composed of three orthogonal and highly dense on-chip magnetic coils that produce a magnetic field using an AC current, the same gradient can be generated with planar CMOS processes for 3D sensing. This is often overcome by employing ferromagnetic materials that are utilized for other on-chip components. The Z sensor coil is implemented in the plane of the chip as a multi-layer spiral (M1 to AP). R Coil results in a thermal noise floor of 364nV/√Hz and 300nV/√Hz for the X/Y and Z sensors respectively. The X/Y sensors generate 660nV of EMF in response to a 10µT field, which is the desired magnetic field resolution to achieve 500μm of mean localization accuracy. Since the X/Y sensors’ output is close to their thermal noise floor, it is imperative to design the front-end instrumentation amplifier (IA) with 8-10x lower input-referred noise (IRF) noise. The IA is implemented in a fully-differential closed-loop architecture and is capacitively coupled to the input to avoid DC offsets and low-frequency noise (Fig.2). Pseudo-resistors are used in the feedback path to provide a GΩ-level impedance, which is needed for the high-pass corner frequency to be within 10-100Hz. The Gm block is implemented as a cascade of two current-reuse stages. The next block is a bandpass filter (BPF) to eliminate the excessive out-of-band noise and improve the SNR (Fig.2). The programmable gain amplifier (PGA) amplifies the EMF signal for the ADC to process (Fig.2). It has a 3-bit tunable gain to accommodate the varying range of EMF throughout the FOV. To extract the peak magnitudes from the outputs of the PGA, a differential peak detector (PDH) circuit is implemented (Fig.2). The outputs of the PDH are fed to a 12-bit SAR ADC (Fig.3). EM simulations are conducted to study the effect of human tissue, frequency variation and other non-idealities on the mutual inductance M (Fig.3).

Measurement results of the sensor are shown in Fig.4 and Fig.5. The IA’s IRN is 40nV/√Hz around the EMF signal frequency. The output of the PGA is 0.5V for a typical mV-level input, resulting in 72dB SNR. The total integrated noise for the X/Y sensors is 8mVrms and for the Z sensor is 1mVrms, which translates to a raw magnetic noise of 64µT rms and 8µT rms respectively. 40 consecutive measurements are averaged to reduce the noise to 8.9µT rms (X/Y) and 1.2µT rms (Z), resulting in an update rate of 25Hz. The linearity error for both the sensor types is <0.2% in a range of ±10mT. To demonstrate tracking and navigation capabilities for clinical applications, the sensor is placed in the tip of a 12-french catheter (Fig.5). To reduce the fabrication cost of the prototype chips, Y sensor is realized using the X sensor of another identical chip by orthogonal placement on a flex PCB. Two such catheter-enclosed devices are submerged in a tissue phantom (placed on top of the gradient coils) to perform relative tracking. The tracking error is determined by the total noise of the 3D sensor (ΔB) and the local magnetic field gradient (G). The measured instrument resolution from a mean value of 350µm (high G FOV) to 600µm (low G FOV), while <1mm throughout the FOV (Fig.5). The comparison table (Fig.6) shows that our sensor is the first of its kind to achieve 3D magnetic sensing using three orthogonal on-chip coils in a standard CMOS process. By employing the technique of EMF induction in passive coils, the sensor does not require any power to operate. The processing circuitry consumes 14.8μW of total power which is orders of magnitude smaller than prior works [1]-[6].

References:
**Fig. 1:** 3D magnetic sensor x-ray view and DC vs AC gradients.

**Fig. 2:** Circuit schematics of the IA, Gm, BPF, PGA and PDH.

**Fig. 3:** Circuit schematic of the timing-block and ADC (top); sensor model and front-end (bottom-left); EM simulations for mutual inductance under different environment/frequency (bottom-right).

**Fig. 4:** Measurement results showing the IRN, frequency response, time-domain waveforms, sensor characterization and noise.

**Fig. 5:** Measurement results of ADC and 3D catheter localization.

**Fig. 6:** Comparison with state-of-the-art. Our CMOS-integrated 3D magnetic sensor has the lowest reported power (<10µT rms noise, ±10mT range, and immunity to DC offsets and low-F noise.)