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Towards a Hall effect magnetic tracking device for MRI*

J.-B. Schell, J.-B. Kammerer, L. Hébrard, *Member, IEEE*, E. Breton, D. Gounot, L. Cuvillon, M. de Mathelin, *Senior Member, IEEE*

***Abstract*— This paper presents the first prototype of a magnetic tracking device for Magnetic Resonance Imaging. The unique relationship between the space coordinates of a MRI scanner bore and the magnetic field gradients used in MRI allows building a localization system based on an accurate measurement of these gradients. These gradients are measured thanks to a 3D Hall device with a footprint of only 50µm^2, integrated with its specific conditioning circuit in a low cost, low voltage 0.35µm CMOS process. The first experimental results show that a sub-millimeter localization is possible. It opens the way to the development of MRI compatible magnetic tracking systems integrable in a surgical tool.**

I. Introduction

In contrast to Computed Tomography and X scanners, Magnetic Resonance Imaging (MRI) scanners have the advantage to provide high tissue contrast, to be no ionizing, and to offer a free image plane positioning technique. These features and the recent onset of wide and short-bore MRI scanners have lead to the development of MR-guided minimally-invasive surgery (Fig. 1). In order to ease the surgical procedure, the surgeon needs a way to track the surgical tool on the MR-image, from the entry point to the target point, e.g. a tumor. Passive tracking using markers filled with a contrast agent may be used but the tracking is slow. Optical localization systems could be used but they suffer from occlusion [1] which prevents their integration into the surgical tool, for instance a needle. The most commonly used active technique is based on miniature Magnetic Resonance MR-probes [2], now integrated on chip [3], which deliver a signal proportional to the precession frequency of the magnetization surrounding the probe. It means that the signal depends on the environment, which makes the device calibration a challenge, unless a specific body, e.g. a kind of contrast agent, is located close to the detection coil of the MR-probe [3]. This complicates the device fabrication. In addition, this coil needs to size at least one or two millimeter square so that the signal magnitude is not too weak.

Another way to locate the position relies directly on the unique relationship which exists between the MRI scanner bore space coordinates and the magnetic field gradients used in MRI. By providing the surgical tool with a small 3D magnetic sensor which measures these magnetic gradients, an easy to use localization system can be built. A tool based on this principle has been patented [4] and commercialized [5]. However, the 3D magnetometer is based on inductive coils which are difficult to deeply miniaturize, preventing its integration into a small surgical tool. The use of a 1D Hall effect magnetometer was proposed in 2004 [6] and even patented [7], but there was no development after that since no accurate and low cost monolithic 3D Hall device was available at that time. This paper fills this gap by presenting the first tracking device based on the use of a 3D Hall probe integrated in a low cost Complementary Metal Oxide Semiconductor (CMOS)-technology.

In the next section, the MRI environment and the localization principle are described. The 3D Hall probe is briefly presented in section III and the magnetic field gradient mapping of a 3T MRI scanner measured with this probe is given in section IV. Section V presents the tracking device with first experimental results before the conclusion and the outlooks of this work.

II. MRI environment and localization principle

Three different magnetic fields are used in MRI. First, a high static main magnetic field, B_0, typically 1.5T or 3T, generates the macroscopic magnetization, precessing at the Larmor frequency. Its direction is called the Z-axis (Fig. 1). Second, a small magnetic field $B_G(G_x, G_y, G_z)$, named magnetic gradient, is added to B_0 leading to a Z-oriented magnetic field given by $B_0 + B_{Gz} = B_0 + G_x{\cdot}x + G_y{\cdot}y + G_z{\cdot}z$. The magnetic gradient B_{Gz} locally modifies the magnitude of the Z-oriented main magnetic field, allowing a spatial encoding of the scanner bore. Third a RF magnetic field is applied perpendicularly to the Z-axis at the right Larmor frequency for image measurement.

Figure 1. MR-guided minimally-invasive surgery

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The authors are with the ICube laboratory, Université de Strasbourg - CNRS, 23 rue du Loess, 67037 Strasbourg, France (corresponding author: +33 388 106259; fax: +33 388 106548; e-mail: luc.hebrard@unistra.fr).

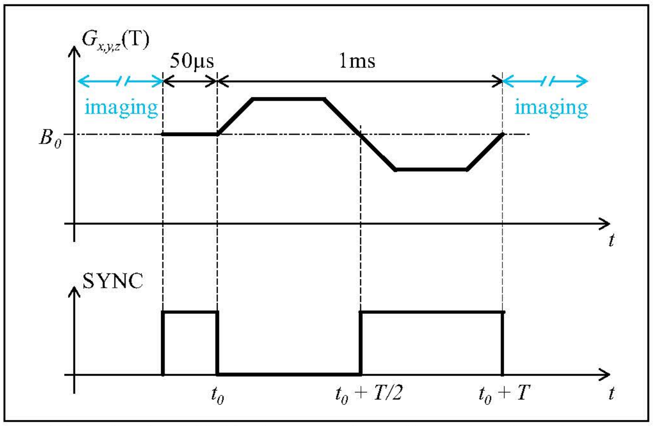

Figure 2. Sepicifc bipolar magnetic gradient pulse, $G_{x,y,z}(t)$, and synchronisation signal used for signal acquisition

The localization of the probe is based on the measurement of three specific magnetic gradients, the first one applied along the X-axis ($G_x \neq 0$, $G_y = 0$, $G_z = 0$), the second one along the Y-axis and the third one along the Z-axis. The big challenge is to accurately measure B_G in the presence of the high B_0 field. For this purpose, we apply successively bipolar B_G pulses as shown in Fig. 2, and we use a specific conditioning circuit (section III.B). These pulses may have a magnitude up to 40mT/m, a width of 1ms and a slew-rate around 200T/m/s.

For MR imaging, the B_G field should be ideally given by $B_G = B_{Gz} = G_x{\cdot}x + G_y{\cdot}y + G_z{\cdot}z$. Nevertheless, as the magnetic flux is conservative, e.g. $\nabla{\cdot}B = 0$, when a magnetic gradient is applied, magnetic lines cannot be perfectly aligned along the Z-axis. The resulting non-zero B_{Gx} and B_{Gy} components of B_G are known as the concomitant fields [8]. MRI gradient coils are used to be designed with a cylindrical geometry leading at first order to the following linear expression of B_G:

$$\begin{bmatrix} B_{Gx} \\ B_{Gy} \\ B_{Gz} \end{bmatrix} = \begin{bmatrix} -G_z/2 & 0 & G_x \\ 0 & -G_z/2 & G_y \\ G_x & G_y & G_z \end{bmatrix} \cdot \begin{bmatrix} x \\ y \\ z \end{bmatrix}$$

This expression is only valid inside the imaging area free from any artifact coming from no-linearities of the magnetic gradients, e.g. ±150mm or ±200mm around the bore center. Outside this area, no-linearities appear. A localization algorithm relying on the above linear relationship between the measurement of (B_{Gx}, B_{Gy}, B_{Gz}) and the space coordinates is easy to implement analytically. However, the surgeon needs to be able to track his medical tool from the entry point, which is close or even outside the linear imaging area, till the target, which is inside the linear imaging area. As a consequence, the localization and tracking algorithm generally relies on a look-up table and an accurate mapping of the magnetic field gradients of the scanner is required [4].

III. 3D Hall probe

The Hall probe uses a 3D Hall device integrated with a specific signal conditioning circuit. Both elements are now briefly discussed, more details being available elsewhere [9].

A. 3D Hall device

We use a 3D Hall device similar to the one we introduced in 2008 [10]. It combines two perpendicular Vertical Hall Devices (VHD), which sense both magnetic field components in the chip plane, and one conventional Horizontal Hall Device (HHD) sensitive to the magnetic field component perpendicular to the chip plane. Both VHD feature external sensing contacts making them compatible with a low cost low voltage CMOS technology [11]. The only difference with [10] and [11] is that the VH-Devices are twice as thick, i.e. $t = 6\mu m$, in order to lower their noise level and increase their resolution. Table 1 summarizes the main Hall devices characteristics.

TABLE I. Hall devices main characteristics

	HHD	VHD
Biasing current (mA)	1.2	3.1
Absolute sensitivity (mV/T)	101.1	19.2
Typical offset (mV)	2.5	0.8
Resolution over [5Hz – 1.6kHz] (µT)	20	50
Foot print (µm*µm)	16 * 16	6 * 25

B. Integrated signal conditioning

The probe chip features three signal conditioning chains, one per magnetic component sensed by the 3D Hall device. Each chain outputs an analog signal which is proportional to the magnitude of the applied gradient pulse. More details about the electronics are given in [9]. The measured sensitivity of the chain connected to the HHD is 148V/T, while it is 110V/T for both chains connected to both VHD.

The B_0 field can be seen as a high offset. To get rid of it, the conditioning circuit performs first a coarse offset cancellation whose right value is determined during the 50µs before applying the bipolar gradient pulse (Fig. 2). The pulse is then integrated with a reversing of the integrator input at half the pulse window. Thus, any DC field or offset are cancelled and the probe output is only proportional to the magnitude of the applied pulse [9]. Finally, the noise levels were measured to be 1.5mV rms at the output of the HHD chain and 4.5mV rms at the outputs of both VHD chains.

IV. Magnetic field mapping in a 3T scanner

A. Experimental setup

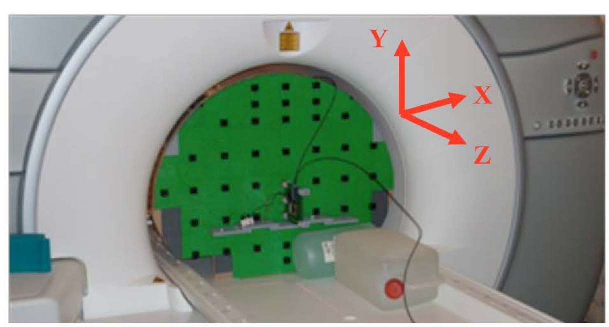

Figure 3. Experimental setup in a wide and short-bore 3T MRI scanner

The 3D Hall probe chip (2.3mm x 3.4mm) was designed in a low cost low voltage 0.35µm CMOS technology. It was mounted on a Printed Circuit Board (PCB) featuring three 14-bit ADC to convert the three analog outputs of the Hall probe. The ADCs are controlled by a micro-controller which also drives the optical transmission used to send serially the signals outside the scanner room, e.g. in the MRI control room. Here, the optical signal is converted back to an electrical signal which is sent to a computer thanks to an Ethernet link. An optical link was also used to send the synchronization signal from the MRI control room to the 3D Hall probe (Fig. 2). The choice of optical transmission was made because ferromagnetic materials are prohibited due to the presence of B_0, and because metallic cables are not recommended due to the magnetic gradients and RF signals which create Eddy currents leading to a warming of the cables. The time delay between the beginning of the analog to digital conversion and the computer acquisition is 950µs. The power supply is a non-magnetic standard lithium-manganese dioxide 9V battery with a 3.3V linear regulator.

A home-made plastic support, fabricated with green LEGO® plates and fasten vertically on the scanner movable table, was specifically designed to fit inside the scanner bore (Fig. 3). The PCB supporting the probe chip was provided with small LEGO® bricks in order to be able to plug it onto the small black LEGO® bricks placed every 80mm on the green plastic support. It permits to locate accurately the 3D Hall probe in the sagittal plane, e.g. the X-Y plane. The accurate location of the probe along the Z-axis is simply obtained by moving the scanner table. Thanks to this setup, we were able to measure the 3D map of the magnetic gradients inside a 3T MRI scanner bore (Fig. 3).

B. B_G map for a G_z magnetic gradient

We first applied a bipolar magnetic gradient along the Z-axis, e.g. $G_x = 0$, $G_y = 0$ and $G_z = 20$mT/m (magnitude of the pulse). The Hall probe, with its HHD chain aligned along the X-axis and its two VHD chains aligned respectively along the Y and Z-axis, was plugged in one of the 7 possible positions along the line y = -80mm on the plastic support, and the three components of B_G were measured for thirteen 80mm equally spaced positions of the scanner table along the Z-axis, e.g. between z = -480mm and z = 480mm. Position x = y = z = 0 corresponds to the isocenter of the scanner bore. This measurement was repeated for each of the 7 positions along the y = -80mm line leading to Fig. 4 where the components B_{Gz} is shown in the plane X-Z, e.g. y = -80mm. Complete measurements of B_{Gx} and B_{Gy} are available in [12].

Combining the measurements of B_{Gx}, B_{Gy}, B_{Gz}, and repeating them for each possible X-Z plane, e.g. y = -80mm to y = 160mm, by steps of 80mm, we get the 3D map of B_G for an applied magnetic gradient of $G_z = 20$mT/m, as shown in Fig. 5. Measurements for a magnetic gradient applied along X or Y, e.g. $Gx \neq 0$ or $Gy \neq 0$ are detailed in [12].

These results are in perfect agreement with the expression of the concomitant fields in the linear imaging area, as discussed in section II. Outside, as expected, the magnetic gradient fields are highly no-linear.

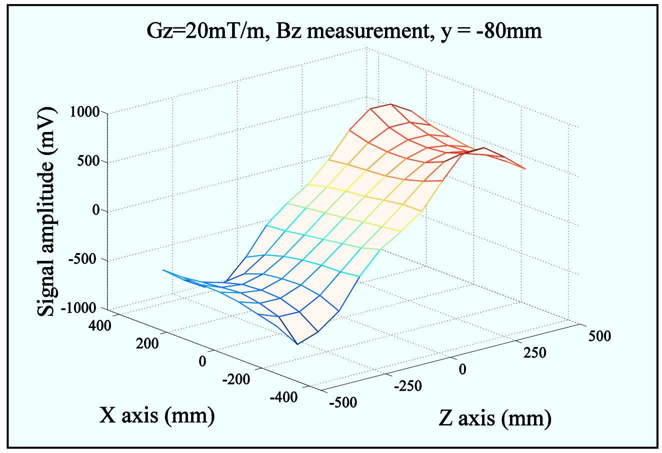

Figure 4. B_{Gz} measurement in the y = -80mm plane

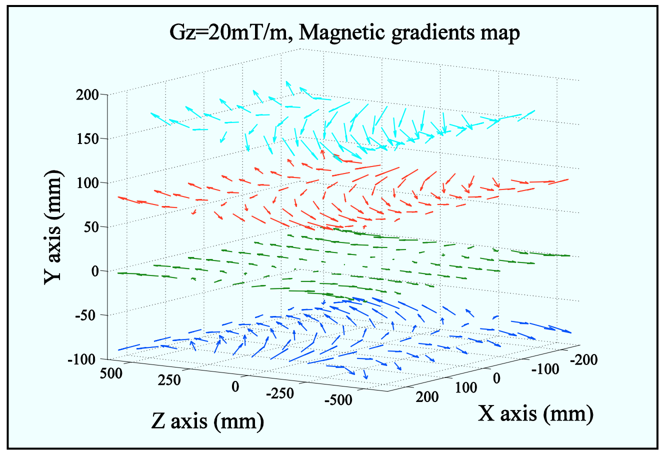

Figure 5. MRI gradient 3D map for G_z = 20mT/m

V. Tracking Device Prototype and First Experiments

Based on these results, a tracking device has been fabricated (Fig. 6 and 7). It features two 3D Hall probes with the second one, at its tip, tilt by 45°. However, only one probe is sufficient for localization and tracking. On the PCB located under the 9V battery, the tracking device features all the required electronics for signal conversion and transmission through an optical cable. The second optical cable is used to transmit the synchronization signal from the MRI control room to the 3D Hall probe (Fig. 2) as already discussed in section IV-A.

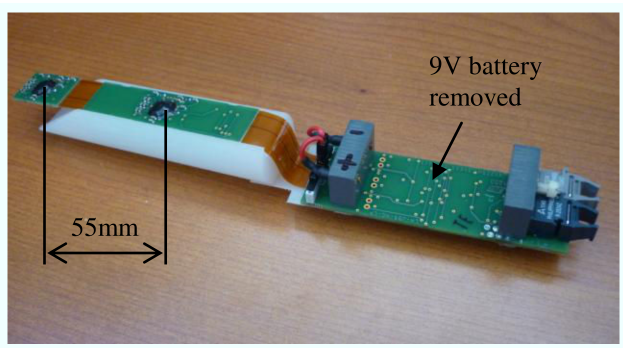

Figure 6. Inside the tracking device. Both 3D Hall probes, 55mm apart, are protected by a glob top epoxy.

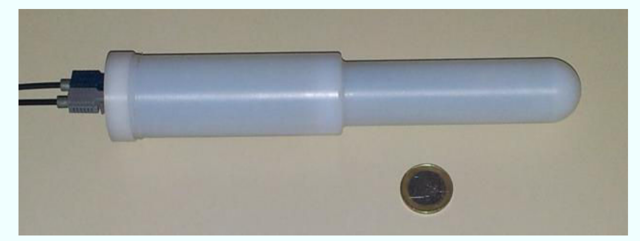

Figure 7. Tracking device prototype close to a 1€ coin

The tracking device was placed on the 3T MRI scanner movable table at x = y = 0, and was aligned along the Z-axis. Both 3D Hall probes, 55mm apart, were maintained horizontal, as shown in Fig. 6. In these conditions, the VHD are aligned along the X and Z-axis, while the HHD is aligned along the Y-axis (Fig. 8). The 3D Hall probe labeled "circuit A" in Fig. 8 was placed at the isocenter of the scanner thanks to a laser marker.

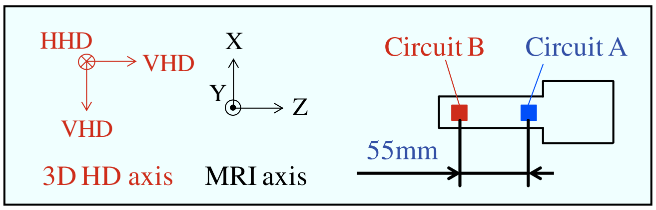

Figure 8. Hall devices alignement for tracking device test

A $G_x = G_y = 0$ and $G_z = 20$mT/m were applied and the B_{Gz} component was recorded, e.g. with a VHD, every 80mm while moving the table between $z = -480$mm to +480mm. From the concomitant field expression, we may say that in the linear imaging area, we have $B_{Gz} = G_z{\cdot}z$. Fig. 9 shows that, as expected, B_{Gz} is proportional to z in this area, while B_{Gx} and B_{Gy} outputs weak signals close to 0. These signals are not perfectly 0 because of possible misalignment of the 3D Hall probes. We can also easily see that both 3D Hall probes are 55mm apart, and that the VHD-chain sensitivity is 2.2mV/mm. Since the noise level measured at the output of the VHD-chain was 4.5mV rms (root-mean-square), it means that when the position of the probe is determined through a VHD, the localization resolution is 2.05mm rms. However this result depends on the magnitude of the bipolar gradient pulse. In this experiment, it was only 20mT/m, while twice this value is possible with most of the MRI scanners. In this case, the localization resolution becomes close to 1mm rms.

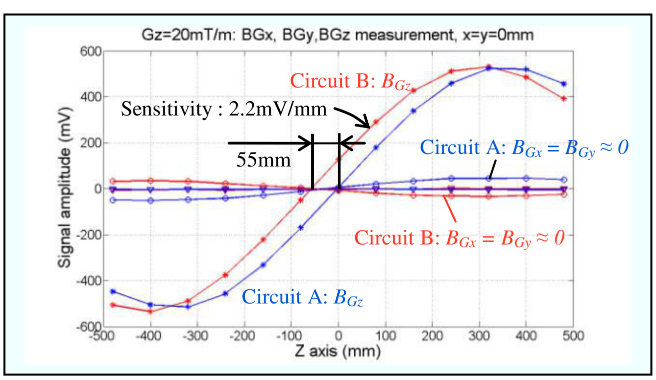

Figure 9. Outputs of circuits A and B for a Gz gradient

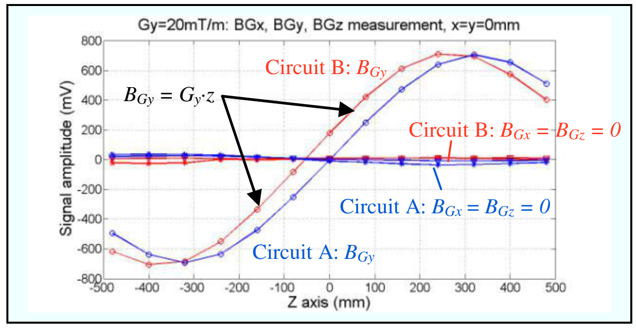

Figure 10. Outputs of circuits A and B for a Gy gradient

The same measurement was performed while applying a $G_y = 20$mT/m. Now, we have $B_{Gy} = G_y{\cdot}z$, which is sensed by the HHD-chain (Fig. 10). The measured HHD-chain sensitivity is 2.96mV/mm with a 1.5mV rms output noise level leading to a localization resolution in the linear imaging area of 0.5mm rms for a gradient of 20mT/m, and 0.25mm rms if the gradient is raised to 40mT/m.

VI. Conclusion and Outlooks

This paper has presented a first prototype of Hall effect magnetic tracking device for MRI scanners. The tracking algorithm is under progress yet. However, preliminary experimental results show that a sub-millimeter localization resolution is possible. The 3D Hall device footprint is only 50µm^2 and there is room to deeply miniaturize the integrated electronics, which opens the way for the development of a MRI compatible real time magnetic tracking system integrable in a surgical tool.

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