# MR evaluation of internal gradients in porous systems: SE vs DDIF method.

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**Purpose:** When a magnetic susceptibility difference exists between a porous matrix and a saturating fluid, local magnetic field gradients (internal gradients, Gi) develop at the interface. Gi are due to the magnetic susceptibility difference between solid porous matrix and the fluid (usually water) which fills pores. Recently, two different methods were used to quantify effective Gi in porous systems such as rocks, cements and trabecular bone structures: the Spin-Echo (SE) method [1,2] and the Diffusion Decay Internal Field (DDIF) method [3-5]. Gi can be extracted by a fitting procedure from SE decay and ADC measurements ($S(TE)=S_0\exp\{(-TE/T_2^{true})-(\gamma^2G_i^2DTE^3/12)\}$). It can be also extracted from DDIF rate using corrections due to T1, gradient slice rate 1/Ts and ADC values ( $1/T_{DDIF}=(1/T_1-1/Ts)+\gamma^2G_i^2t_e^2D$ ).Aim of this study was to compare, SE and DDIF methods in evaluating Gi from water in packed micro beads as a function of their mean pore size.

**Materials and Methods:** A MR system operating at 9.4T and equipped with a micro-imaging probe with a maximum gradient strength of 1200 mT/m (rise time of 100 µs) was used to investigate seven different samples of packed polystyrene beads mono-dispersed in 8 mm NMR tube. Beads characterized by mean size of 6, 10, 15, 30, 40, 80 and 140 micrometers were investigated. The magnetic susceptibility difference between polystyrene micro-beads and water results to be in the same range of those in human tissues: $|\Delta\chi_m| = |\chi_m^{H2O}-\chi_m^{Polystyrene}| = 1.59*10^{-6}$ (in S.I. units). T1, T2, ADC parameters were measured and SE and DDIF decays were collected to extract Gi using both SE and DDIF method. MSME (Multi Slice Multi Echo) imaging sequences at different TEs (from 1,7 to 500 ms) and at different TRs (from 200 to 15000ms) were used to assess SE decay, T2 and T1 from selected slices (slice thickness =600 µm). Diffusion coefficient was measured by means of Pulsed Field Gradient Stimulated Echo (PGSTE) imaging sequences (TE/TR=40/3000 ms, diffusion gradient pulse delay Δ=80 ms, diffusion gradient pulse duration δ=2 ms and ten b-values ($b=\gamma^2g^2(\Delta-\delta/3)$ ranging from 1000 to 20000 s/mm$^2$). DDIF rate was measured by means of STE ( TE/TR=40/6000, with 12 diffusion time $t_D$ ranging from 15 to 700 ms). In the original version of DDIF [3-5] ADC of free water ($D=2.3*10^{-9}$m$^2$/s) was used, while in the modified version presented here (DDIF(M)), ADC of water measured in porous systems is used. Moreover we evaluate for each sample the characteristic diffusion length $l_d=(2D\Delta)^{1/2}$, the structural length scale $l_s=((3^{1/2}+1)/4)d$ and the dephasing length $l^*=(D/\gamma G_i)^{1/3}$ [6]. In particular we estimate two types of diffusion lengths: $\ell_d^{PGSTE}$ associated to SE and DDIF(M) (D=measured ADC) and $l_d^{H2O}$ associated to DDIF (D=ADC of free water). Similarly we evaluated three $l^*$ for each methods (Fig.2).

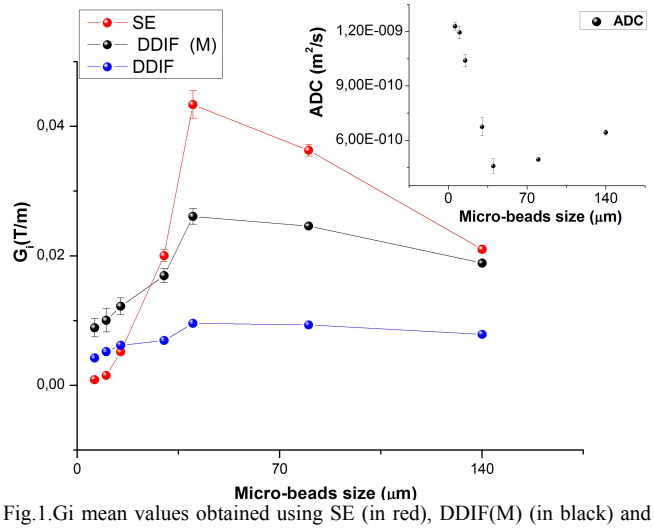


Fig.1.Gi mean values obtained using SE (in red), DDIF(M) (in black) and DDIF (in blue) method, as function of beads size. Graph on the right shows measured ADC as function of beads size.

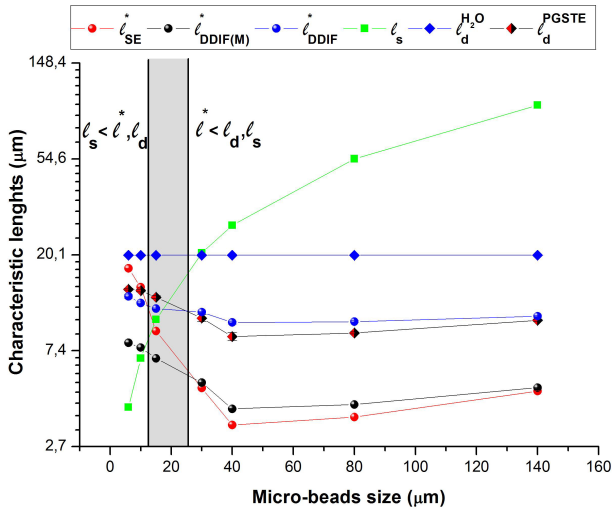


Fig.2. Characteristic lengths as a function of bead mean size. Three different regimes are identified: the fast-diffusion (FD) $l_s< l_d$, $l^*$ regime, the slow-diffusion (SD) $l^*< l_d$, $l_s$ regime and an intermediate regime which is a transition regime between FD and SD.

**Results and discussion:** Mean values of Gi obtained using SE, DDIF(M) and DDIF methods and ADC, as a function of beads size, are displayed in Fig.1. Characteristic diffusion lengths as a function of mean size are reported in Fig.2. Gi behavior as a function of beads size is closely related to characteristic lengths. In particular in Fig. 2 three regimes can be identified: the fast-diffusion (FD) $l_s< l_d$, $l^*$ regime, the slow-diffusion (SD) $l^*< l_d$, $l_s$ regime and an intermediate regime which is a transition regime between FD and SD [4]. Effective Gi is higher in SD than in FD region. Indeed in the former, spins explore the entire pore many times before the dephasing occurs. As a consequence effective Gi are almost totally averaged out by diffusion (Fig.1). In the latter regime spins near the wall pore experience an additional dephasing which is not averaged by dynamics because the $l_d$ is lower than the pore space (approximately $l_s$). In general the behavior of Gi extracting using SE, DDIF(M) and DDIF (Fig.1) as a function of beads sizes show similar trends. However, Gi values extracted from SE decay method showed higher values than those obtained using DDIF(M) and DDIF. Moreover Gi extracted from SE better discriminates between polystyrene beads samples as compared to Gi obtained from DDIF(M) and DDIF. There are two main reasons to explain the difference between Gi values measured by using SE and DDIF. First of all, effective Gi extracted from SE depends on both water diffusion regime and magnetic susceptibility difference. Conversely in DDIF the dynamics of water is assumed to be constant (equal to that of free water) in all samples and Gi value is characterized by a feeble dependence on the dynamics. Secondly, in order to quantify Gi, DDIF assesses deviations from $T_1$ relaxation rate, while SE evaluates deviations from $T_2$ relaxation rate.

**Conclusion:** Effective Gi measured by using SE, DDIF and DDIF(M) methods for discriminating porous systems characterized by different pores size, were compared. The behaviors of Gi extracted from SE, DDIF and DDIF(M) decays as a function of beads sizes, show similar trends. However, Gi values extracted from SE decay better discriminates between different porous systems when compared to Gi extracted from DDIF and DDIF(M) decay. Finally SE, unlike DDIF method, can be easily implemented on clinical scanners and requires less time for data acquisition and data processing than DDIF one.

**References:**[1] Callaghan PT, Oxford Science, 1995.[2] De Sanctis S. et al. Phys. Med. Biol. 55 (2010) 5767–5785. [3] Sigmund EE et al. Magn. Reson. Med. 59(1) (2008) 28–39. [4] Song Y.-Q. Concepts Magn Reson Part A 18A (2003) 97-110. [5] Song Y.-Q. et al. Nature (2000) 178-181 [6] Mitchell J. Et al. Phys. Rev. E81 (2010) 026101.