MR Imaging of Slow-flow Using a Flow Phantom

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Purpose : To find sensitivity of MRI imaging methods to slow flow phantom study was performed with conventional Spin-Echo, gradient echo based Phase Contrast, fast GRASS, and heavily T2-weighted Fast Spin Echo pulse sequences.

Materials and Methods : A siphon driven flow phantom was constructed with a ventriculo-peritoneal shunt catheter and a GE phantom to achieve continuous variable flow. Four different pulse sequences including Spin-Echo, Phase Contrast, GRASS and Heavily T2-weighted Fast Spin Echo were evaluated to depict slow flow in the range from 0.08 ml/min to 1.7 ml/min and to compare signal intensities between static fluid and flowing fluid.

Results : In the slow flow above 0.17 ml/min conventional Spin-Echo showed superior apparent contrast between static and flowing fluid while GRASS was more sensitive to the very slow flow below 0.17 ml/min. It was not accurate to calculate flow and velocity below 0.1 ml/min with a modified PC imaging.

Conclusion : Four different MR pulse sequences demonstrated different sensitivity to the range of slow flow from 0.08 ml/min to 1.7 ml/min. This finding may be clinically useful to measure CSF shunt flow or detecting CSF collection and thrombosis.

Index words : slow flow, GRASS, Phase contrast, Spine echo, flow imaging

Introduction

It is well established that fast flow motion produced high signal in gradient MR angiography for various clinical application (1). However there are couple of clinical environments which related to slow motion in biological fluids like cerebrospinal fluid (CSF) pulsation, CSF leakage (2), communication between the cyst and subarachnoid space (3), CSF in shunt tubing, in the urinary tract and fluid in the biliary tract. In particular flow inside CSF shunting could be varied according to patient pathological status. Heavily T2-weighted sequence usually depict well the very slow fluid motion in the MR urography or the magnetic resonance cholangio-pancreatography(MRCP) tech-

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nique. In the intermediate range of slow flow conventional T1-weighted spin-echo sequence has been known to generate high signal rather than signal void (4, 5).

Phase contrast (PC) imaging using gradient echo have been used successfully to detect the velocity and flow of fluid quantitatively. However it remains to evaluate precise velocity range of slow flow within which each MR sequence shows the greatest sensitivity to slow flow. In this study GRASS (Gradient recalled acquisition in the steady state), spin-echo, 2D-PC, heavily T2-weighted spin echo were evaluated to define the range of sensitivity to the slow flow using a siphon-driven flow phantom.

Materials and Methods

A flow phantom was constructed with 1.3 mm OSVII ventriculo-peritoneal shunt catheter (NMT neurosciences Implants S.A, France) as shown in figure 1. A single loop of tubing was positioned over a GE manufacture-provided MRI phantom along the axis of the main magnetic field such that flow was perpendicular to the transverse axial plane. The flow phantom was positioned in the center of the magnet, 1 meter from the entry surface. Isotonic saline was used instead of the body fluid such as CSF. A continuous variable, nonpulsatile flow was achieved by siphoning the flowing medium (physiological saline) from a reservoir which was hung over the stand. All measurements were taken relative to a stationary tube of saline adjacent to the tubes containing flowing fluid.

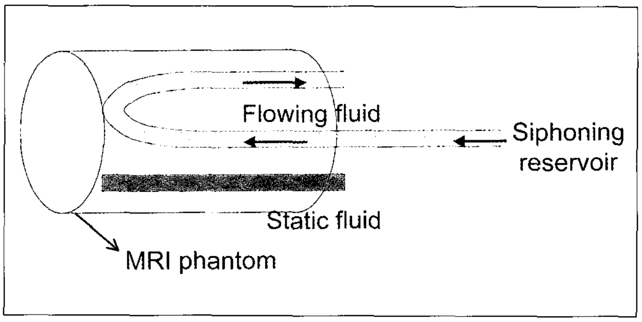


Fig. 1. Schematic diagram of flow phantom. Flowing fluid is con-nected with siphoning reservoir.

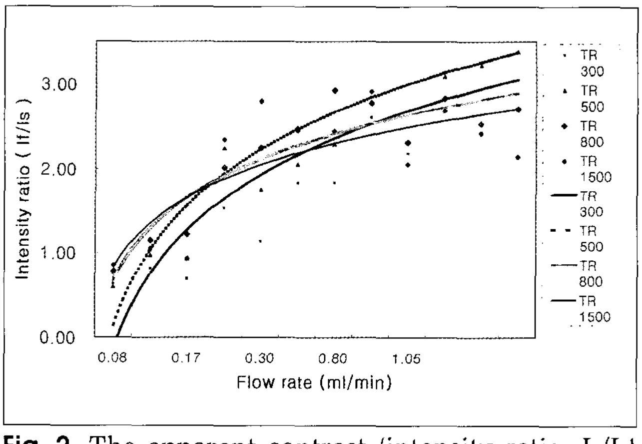


Fig. 2. The apparent contrast (intensity ratio, I_F/I_S) between static and flowing fluid is plotted against flow rate for various TR in Spin-Echo imaging.

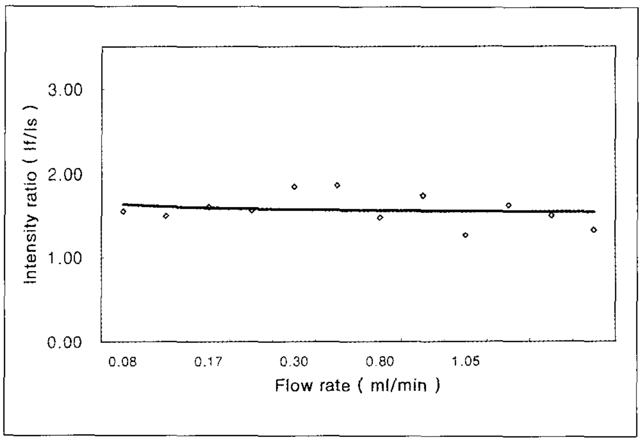


Fig. 3. The apparent contrast (intensity ratio, I_F/I_S) between static and flowing fluid is plotted against flow rate in driven-equilibrium GRASS imaging.

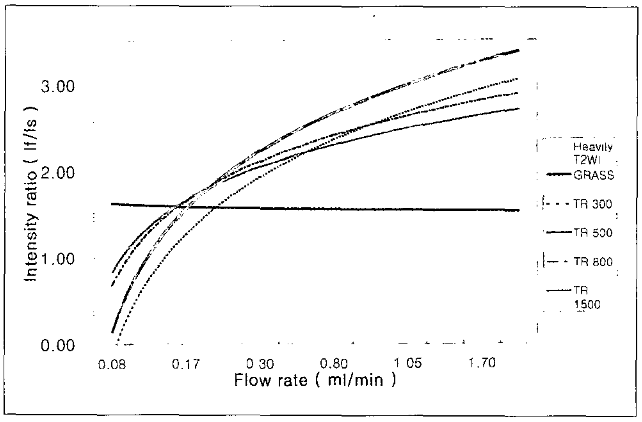


Fig. 4. Superimposed plot demonstrate differential sensitivi-ty of GRASS, Spin-Echo and heavily T2-weighted fast spin echo imaging to the slow flow. X-axis represent flow rate (ml/min). Slow flow sensitivities of two sequences were crossed around 0.17 ml/min

Timed drainage into a graduated cylinder provided a measure of volumetric flow rate. MR imaging was performed with a series of MRI sequences, on the flow phantom using a 1.5 T clinical MRI unit (GE Signa Horizon, Wisconsin, USA).

A heavily T2-weighted Fast Spin Echo (TR/TE = 3500/ 90ms) was acquired to find the presence of fluid inside shunt tubing. In T1-weighted Spin Echo sequence, increasing TR from 300 ms to 1500 ms at fixed TE, made it possible to monitor flow image according to flow inside tube. Subsequently flow was measured quantitatively using a phase contrast imaging. This sequence was modified such that minimum velocity encoding VENC was from 5 cm/sec to 0.5 cm/sec using EPIC (GE pulse sequence programming software) software. Imaging resolution was taken as 0.39 mm/pixel (FOV: 8 cm, 256×224 imaging matrix) to consider inner diameter of shunt tube, 1.3 mm. Finally, we tried to visualize flow-dependent contrast of static and flowed tubes using a driven equilibrium GRASS sequence with a TR of 17.9 ms, a TI of 60 ms, flip angle of 90° and bandwidth of 31.3 Khz. The imaging slice thickness was 5 mm in every performed pulse sequence.

The flow phantom was positioned over a GE phantom in the center of a 30-cm head coil to simulate conditions of clinical imaging. The intensity recorded for the intraluminal signal was averaged over 3 mm^2 from circular region of interest in the center of the tube at MRI console by placing a cursor into a ROI. These pixels were chosen specifically to avoid boundary layer effects at the tube surface. The intensities from the flowing and stationary fluid were measured for steady volumetric flow rates of 0.07 to 1.7 ml/min.

We calculated the average intensity of the flowing fluid and the static fluid inside each lumen. For convenience, we signify by IF the average intensity of the flowing fluid, by IS that of the static fluid, and by Cv ratio of two intensities (I_F/I_S). We tested flow velocity of 0.72, 0.97, 1.60, 1.89, 2.83, 4.90, 7.54, 8.49, 9.90, 10.56, 11.32, 16.03, 21.69 mm/sec (see Figure 2). The intensity of flow phantom relative to adjacent stationary fluid are plotted as function of flowing velocity (see Figure 4).

Results

Figure 1 shows the flow phantom with flowing and static fluid. In conventional Spin Echo, Flow-induced enhancement effect could be early seen from the flow rate as low as 0.07 ml/min where the flow induced signal loss was also noted with a GRASS sequence. In the Figure 2 Cv ratio(I_F/I_S) against the flow speed was plotted for conventional Spin Echo with various TR values while TE was set at 40 ms. The intensity curve showed a semi-logarithmic increase from 0 to 1.7 ml/min. Corresponding plot of Cv (I_F/I_S) ratio for a GRASS was shown in the Figure 3 which showed nearly constant value with respect to the flow rate range of 0 to 1.7 ml/min. Superimposed three Cv graphs including a data from heavily T2-weighted FSE (see Figure 4), demonstrated relatively different sensitivity of three pulse sequences to the slow flow with observation of linear, constant value of Cv for GRASS and increasing value for Cv of conventional Spin Echo. In the flow rate above 0.17 ml/min (1.6 mm/sec) conventional Spin Echo was superior to GRASS sequence for displaying slow flow whereas GRASS was more suitable below 0.17 ml/min (1.6 mm/sec) than Spin Echo sequence.

Table 1 showed the average of signal intensities on a heavily T2-weighted FSE sequence for each flow rate. The signal intensity of flowing fluid was measured higher than static fluid, and its ratio was calculated. Table 2 showed calculated flow (ml/min) and velocity (mm/sec) from a modified phase contrast imaging, compared to actual flow and velocity which was measured from the collecting volume of fluid within a

Table 1. Signal intensity of flowing and static fluids from heavily T2-weighted fast spin echo (arbitrary unit)

flow scale(arbitrary)	static fluid	flowing fluid	I_F/I_S
4	588	688	1.17
6	575	616	1.07
10	580	670	1.15
15	554	614	1.10
20	581	622	1.07
30	533	656	1.23
50	445	552	1.24
75	552	655	1.18
83	527	638	1.21
104	520	600	1.15

specified time. In the region of very slow flow (lower than 1mm/sec), flow data from PC imaging was not matched with the actual flow. As the flow velocity increased by a higher level, the calculated flow velocity from the PC imaging became closer to the actual flow velocity.

Discussion

In the conventional Spin-Echo sequence if the repetition time (TR) is on the order of tissue T1, the effect of slow flow would be to substitute partially saturated spins with fully magnetized spins. As a result, the obtained signal would be stronger. With increasing flow speed, more and more excited spins are lost from the refocusing region, resulting in the loss of signal. Therefore, the effect of flow on MRI signal would be the combination of the following two factors : substitu-

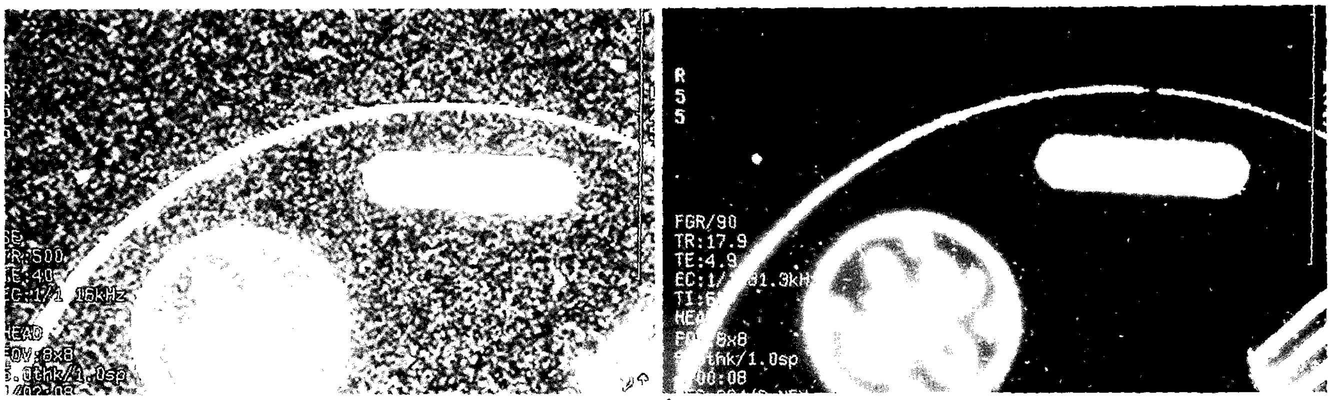
Table 2. Comparison the actual flow and calculated velocity from PC mapping

	Measured volumetric flow rate*		PC mapping	
flow scale	ml/min	mm/sec	ml/min	mm/sec
4	0.08	0.72	0.1	1
6	0.10	0.97	0.1	0
10	0.17	1.60	0.2	1
15	0.20	1.89	0.3	1
20	0.30	2.83	0.4	3
30	0.52	4.90	0.7	4
42	0.80	7.54	0.8	7
50	0.90	8.49	0.9	7
62	1.05	9.90	1	9
75	1.12	10.56	1.4	11
83	1.20	11.32	1.6	12
104	1.70	16.03		
150	2.30	21.69	2.3	18

*:manual measurement by calculating collected fluid in a given time duration.

a b

Fig. 5. Very slow flow(0.08 ml/min) imagings from Spin-Echo (**a**) and GRASS (**b**) showed differential apparent contrast and sensitivity between static and flowing fluid.



a b

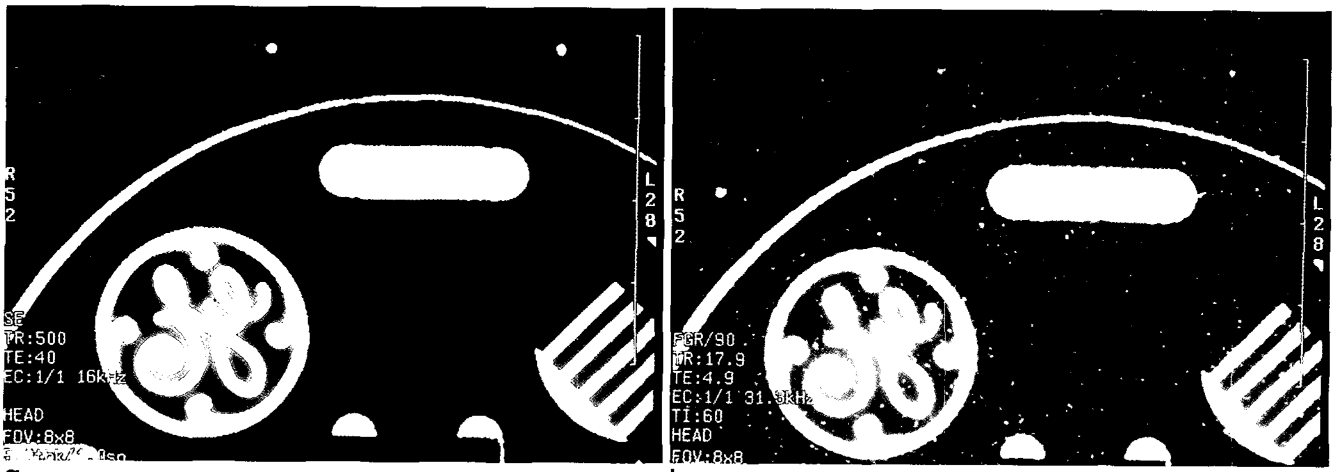
Fig. 6. Very slow flow(0.10 ml/min) imagings from Spin-Echo (**a**) and GRASS (**b**) showed better apparent contrast and sensitivity for GRASS imaging between static and flowing fluid.

tion of partially saturated spins with fully magnetized spins and loss of excited spins from the refocusing region. As a combination of these two effects, the obtained signal may either increase or decrease. The flow speed that gives the maximal enhancement is given by the next equation (6).

$v = d/TR$

where d is slice thickness (here d was 5 mm, TR = 300-500 ms, v = 1.7-1.0 cm/sec) As flow is getting slow, TR should be increased to see flow enhancement. But the intensity for static fluid was also increased due to increased proton density weighting factor which impede visual contrast between static and flowing fluid.

In this study with GRASS very short TRs, less than 100ms, was used. This results in a condition in which residual transverse magnetization has built up to the point that some coherent transverse magnetization, so called steady state, is also present. This steady state condition was emphasized with several factors including short TR(17.9 ms), large flip angle(90°) and tissue preparation time(60 ms) which make contrast more T2*-weighted. To preserve this residual transverse magnetization, left over from the previous



a b

Fig. 7. Slow flow(1.12 ml/min) imagings from Spin-Echo (**a**) and GRASS (**b**) showed much better apparent contrast and sensitivity for Spin-Echo imaging between static and flowing fluid.

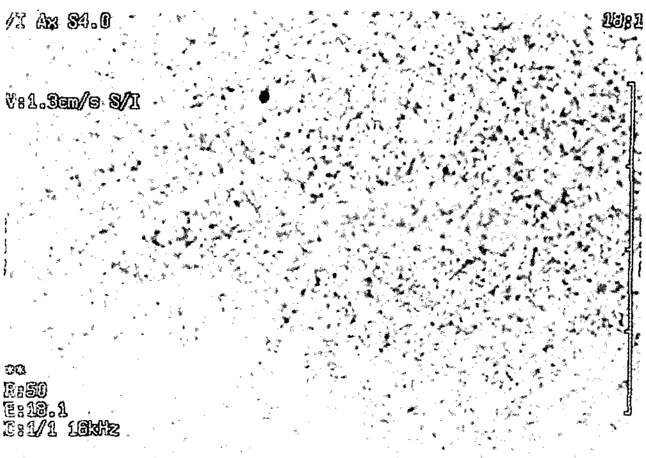


Fig. 8. Phase contrast imaging of a slow flow (0.4 ml/min) showed clear distinction of flow and static fluid, and in addition direction of flow (bright and dark phase map).

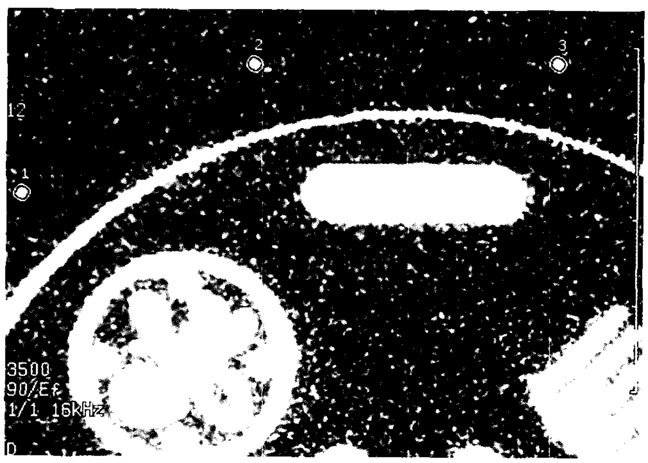


Fig. 9. Heavily T2-weighted fast spin echo imaging of slow flow was used to find position and size of flowing fluids for further calculation of intensities of flowing fluids from other imagings. 1 was static fluids, 2 was flowing fluids (I→S) and 3 was reverse-directional flowing fluids(S→I).

excitation pulses, a rewinding gradient pulse is applied. As a result, this steady-state magnetization makes some contribution to the signal. Tissues with longer T2 relaxation times may have larger steady-state components than tissues with shorter T2 relaxation times, which means that the rewinder gradient contributes some T2* weighting to the image. As the flip angle set to 90 degrees and is combined with a very short TR(17.9 ms), heavy saturation occurs due to insufficient time for Tl recovery between pulses, and a large steady-state component develops (7). This means T1 has little impact on the contrast and T2* has a greater impact due to the large steady-state component. A 90/180/90 RF Driven Equilibrium (DE) preparation pulse is designed to produce more T2-weighted contrast with Fast Gradient Echoes. The initial 90-degree pulse creates transverse magnetization, which immediately begins to dephase. The amount of dephasing is controlled by the preparation time. The longer the preparation time, the more dephasing and the more T2-weighting. The remaining transverse magnetization is rephased by the 180-degree RF pulse. In this fast GRASS sequence any process like flow which causes to interfere steady state give rise to signal decrease (7, 8). Other T2* dephasing processes, such as static fields inhomogeneities, intravoxel dephasing, chemical shift, and magnetic susceptibility artifacts are also contributing to signal decay.

In comparison of slow flow sensitivity the apparent contrast between static and flowing fluid (expressed as $Cv = I_F/I_S$) were obtained from Spin-Echo and GRASS as flow velocity were varied from 0.72 mm/sec to 1.6 cm/sec. As demonstrated in Figures 5-6 , apparent contrast of flowing fluid in GRASS was better than Spin-Echo at flow velocity below around 1.6 mm/sec. Spin-Echo sequence showed much better contrast to depict slow flowing fluid in the range of flow velocity from 2 mm/sec to 1.6 cm/sec(see Figure 7). SE and GRASS showed complementary visual contrast between static and flowing fluid in the range of slow flow.

Phase contrast techniques have widely used for quantitative velocity and flow measurements (9). In the case of the conventional method, such as gradient echo imaging, a bipolar gradient pulse encoding flow is implemented between each excitation pulse and echo signal. This has been already applied for diffusion weighted imaging. However, this method seems further attractive for the imaging of very slow flow, such as CSF flow, because the phase of each echo reflects the phase shifts generated during preparation period. In this study minimum VENC was modified to 0.5 cm/sec. In the range of very slow flow (velocity is less than 1 mm/sec) PC velocity mapping was not consistent with actual flow measured from collecting fluid during a specified time. The phase mapped due to very slow flow was not clearly displayed enough to calculate velocity by assigning ROI with curser. The relatively higher flow velocity (above 1 mm/sec) demonstrated clear phase maps of flow as shown in Figure 8 where brightest one and darkest region indicated forward and backward flow.

Heavily T2-weighted sequence depict well the very slow fluid motion. Signal intensity was higher in the flowing fluid than in the static fluid as shown in Table 2, whereas apparent contrast was not distinguishable at all between them(see Figure 9). However his imaging could be primarily used to find the position and the size of static and flowing fluid inside the tube.

Since the flow velocity inside CSF shunting tube was variable in measuring time and also depending on patient' s pathologic status, the range of flow rate was not known exactly. But, in our hands, it was in the range of 0.5 to 10 mm/sec considering our experience and other' s previous data. Thus It is desirable to apply different pulse sequence to measure flow rate of shunt tube depending on the CSF flow velocity inside tube as shown in this study.

Conclusion

Very slow flow under 1 cm/sec was well depicted using GRASS, Spin-Echo, 2D-PC, heavily T2-weighted Spin Echo sequences. Calculating velocity of slow flow above 1.6 mm/sec was relatively accurate using 2D-PC imaging while calculating very slow flow below 1.6 mm/sec was not consistent with measured actual flow. In the very slow flow below 1.6 mm/sec, GRASS showed superior apparent contrast to conventional Spin-Echo technique, while in the region of slow flow above 1.6 mm/sec, Spin-Echo was much better for depicting apparent contrast between static and flowing fluid. This method may provide a potentially useful slow flow imaging within CSF shunt or for detecting

CSF collection and thrombosis. In the slow flow above 1 cm/sec, heavily T2-weighted FSE imaging demonstrated a slight TOF enhancement compared to the static fluid since IF/IS was greater than average value of IF/IS, 1.16.

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유동모형을 이용한 저속유동의 자기공명영상

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목적 : 고식적 스핀에코, 위상 대조 경사에코, 고속 GRASS, 중T-2강조 고속스핀에코 연쇄를 각각 유동모형에 적용시켜, 저속유동에 대한 자기공명영상의 민감도를 찾아내고자 한다.

대상 및 방법 : 뇌실-복강 단락 도관과 GE 모형으로 이루어진 유동모형으로 싸이폰 효과에 의한 지속적인 가변 유속의 흐름을 내보내고, 각각의 유속에 위의 네 가지 자기공명영상연쇄를 적용시켰다. 0.08 ml/min 에서 1.7 ml/min 범위의 유량의 흐르는 액체와 정지된 액체에서 획득된 자기공명영상의 신호강도를 비교한 결과를 요약하면 다음과 같다.

결과 : 0.17 ml/min 이상의 느린 흐름에서는 고식적 스핀에코영상이 가장 우수한 정지-유동 액체사이의 걸보기대조를 보이나, 그 이하의 매우 느린 흐름에서는 GRASS 영상이 더 높은 민감도를 보였다.

결론 : 4가지 자기공명펄스연쇄는 0.08 ml/min에서 1.7 ml/min 범위의 유량의 흐름에서 서로 다른 민감도를 보였다. 이 결과는 임상적으로, 단락수술후의 뇌척수액의 흐름이나 혈관내의 혈전에 의한 흐름의 변이 등의 인체내의 저속유동을 검출하는 비침습적방법으로 유용하게 적용될 수 있을 것이다.

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