Technical Assessment of Artifact Production from Neuro Endovascular Coil At 3 Tesla MRI: An In Vitro Study

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Abstract- **Introduction: Magnetic resonance imaging (MRI) is an essential part of the diagnostic procedures in radiology. MRI 3 Tesla becomes more widespread due to high signal to noise ratio (SNR). The use of the neuro endovascular coil to overcome the neuro aneurysm can introduce the artifact in magnetic resonance imaging. Susceptibility artifacts and geometric distortions caused by magnetic field inhomogeneity- related signal loss is used to refer to an artifact in magnetic resonance images. It consists of a region of signal void with a surrounding area of an increased signal intensity that appears to be considerably larger than the actual size of the device causing the artifact. The objective of the study is to compare the size of the artifact on the MR image to the actual size of endovascular coils using a 3 Tesla magnetic resonance imaging system, in vitro study.**

Methods: The endovascular coils were made from detachable platinum and aneurysm models were constructed by using silicone tube. MRI 3 Tesla Philips Model Achieva with pulse sequence selections were: spin echo, fast spin echo, inversion recovery, fast gradient echo while additional parameters were echo time and turbo factor.

Results: Improved visualization of perianeurysmal soft tissues is best accomplished by spin echo for fast spin echo sequences, even better suited to reduce metal artifact. Furthermore, shorter turbo factor and shorter effective TE in the latter sequences are beneficial for the same reason as sequences having shorter TE. Sequences with a shorter TE are preferred because of less time for dephasing and frequency shifting. Imaging at gradient echo series increases susceptibility artifacts. In this in vitro study, some of the major characteristics related to MRI imaging of coil packs have been defined.

Discussion: Pulse sequence spin echo is the best sequence reducing the susceptibility artifact. Reducing the TE is the main factor in improving endovascular coil visualization on MRI images. The longer turbo factor of fast spin echo sequence, the echoes at the end of the echo train are more susceptibility artifact. The use of small field of view, high resolution matrix and thin slice can help reducing susceptibility artifacts.

Conclusions: Improved visualization of perianeurysmal soft tissue is best accomplished by selective Spin echo and Fast Spin echo sequence because of significant decreases in susceptibility artifacts. In any sequence, reducing the TE is the factor improving susceptibility artifact on MRI images.

Keywords- **MRI 3 Tesla, Endovascular coiling, Susceptibility artifact, Pulse sequence**

I. Introduction

Magnetic resonance imaging (MRI) at field strength of 3 Tesla has become frequently use with an increasing number of radiological sites in recent years. 3.0 Tesla MRI starts playing the same role for clinical imaging that were occupied by 1.5 Tesla systems in the past. The physical limitations of MRI 3.0 T relate to the higher field strength and the protocols transferred from 1.5 Tesla MRI which is not yet fully optimized. Furthermore, the B0 inhomogeneity and susceptibility effects, B1 inhomogeneity and wavelength effects, chemical shift effects also introduce larger artifacts for MRI 3.0T.

Endovascular coiling or endovascular embolization, is a procedure performed to block blood flow into an aneurysm (a weakened area in the wall of an artery). An aneurysm in the brain is called a cerebral aneurysm, a brain aneurysm, or an intracranial aneurysm.

Endovascular coil embolization has become a standard treatment option for managing intracranial aneurysms. Post procedure complications of endovascular coil embolization include refilling of blood into the aneurysmal lumen, enlargement of the aneurysm, and protrusion of detached coils into the parent artery. Frequent and detailed follow-up evaluation of the post procedure condition of coil embolized aneurysms, especially regarding their lumen, is essential. However, technical difficulties often limit such evaluation. A number of techniques such as catheter angiography, computerized tomographic (CT) imaging, and magnetic resonance angiography (MRA) can be employed for this purpose, among which MRA appears to be the most promising.

*B*0 inhomogeneity and susceptibility affect an extremely homogeneous static magnetic field, which is required around the isocenter of the magnet for magnetic resonance imaging. The homogeneity of the static magnetic field influences the distribution of the Larmor frequencies of the protons and also the linearity of the magnetic field gradients required for spatial encoding. If the static magnetic field is disturbed, different effects of reduced *B*0 homogeneity can be observed. Variations of the Larmor frequency within a single voxel result in dephasing of the spins and, thus, in signal loss in acquired gradient echoes. This effect is quantified by the relaxation time, *T*2 *, which becomes shorter with increasing microscopic field inhomogeneity

Metal and susceptibility artifacts from neuro endovascular coil can attenuate the intravascular signal and therefore reduce the apparent caliber of vessels or produce vessel wall irregularities. This typically occurs at the cerebral artery where signal drop-outs related to intravoxel dephasing (turbulence and acceleration due to complex vascular anatomy) may be further enhanced by susceptibility artifacts generated.

II.METRIALS AND METHODS

A. MRI system and endovascular coils

MRI system Philips Achieva 3 Tesla (Philips Healthcare, Best, Netherlands) was used in this study. A detachable coil used for the treatment of cerebral aneurysms (TruFill DCS Orbit Detachable coil, Cordis Neuro vascular, Inc., Miami Lakes, FL, USA) was assessed in this investigation. The outer diameter of these wires was 0.254 mm, the diameters of circular shape coil were 3 and 4 cm and the number of coils was 6.

B. Aneurysm model and coils pack density

Three aneurysm models were constructed by using silicone tube; the inner diameters were 2, 6 and 9.7 mm. assuming an aneurysm model of $10x10x10mm^3$. The volume of coil pack was calculated by using the following formula.

Volume of coil pack = $(4/3)\pi(a/2)(b/2)(c/2)\ mm^3$.

Where a, b and c represent three orthogonal dimensions. The length of coil could then be calculated by using a coil specific formula provided by the vendor:

Coil volume = $([\pi x \{coil\ OD\ mm\}^2]/4x$ coil length mm.

C .Image acquisition

MRI studies were performed by using the 8 channel sense head coil and four pulse sequences (Table 1). Variables were selected to investigate coil pack artifact. Variables included the echo time (TE), acquired voxel dimension, and turbo factor. For all sequence experiments, the following parameters were kept constant: repetition time (TR), FOV, rectangular FOV aspect ratio, flip angle, and phase encoding direction.

Table. 1 The scanning technique at MRI 3.0 T.

Parameter	Inversion Recovery	Spin Echo	Fast Spin Echo	Fast Gradient Echo
TR(ms)	11000	700	4000	107
TE(ms)	125	15	80	4.6
TSE Factor	-	-		-
TI(ms)	2800	-	-	-
FOV(mm)	180	180	180	180
Matrix	Varies	Varies	Varies	Varies
TH(mm)	Varies	Varies	Varies	Varies
Gap(mm)	0.0	0.0	0.0	0.0
BW(Hz)	Varies	Varies	Varies	Varies

TR: time repetition, TE: time echo, TSE Factor: turbo spin echo factor, TI: time inversion recovery, FOV: field of view, Matrix: matrix size, TH: thickness, Gab: slice gap, BW: bandwidth.

D. Data Analysis

The data was analyzed by using the Advantage Workstation 4.4 with high accuracy in routine study. The regions of interest were drawn to encompass the coil induced signal intensity loss. The areas were summed, providing a volume in cubic millimeters. Artifact on overestimation factor for volume measurements, $\alpha_{v,}$ was calculated on the basis of the measured artifact volume, v (MRI) with actual coil pack volume, v(coil pack) using the equation:

$$\alpha_v = v\ (MRI)/v(coil\ pack)$$

E. Statistical Analysis

The signal loss volume and the diameter were analyzed with an analysis of variance (ANOVA). The effects of parameter changes were analyzed. The null hypothesis was rejected at $P \leq 0.05$.

III. Results

Table. 2 Percent errors in pulse sequence of signal loss due to coil-induced susceptibility artifact, the actual size of coil pack maximum dimension 7.6x6.5x9.0 mm**.**

	% Error			
Matrix size/Slice thickness	Spin Echo	Fast Spin Echo	Inversion Recovery	Fast Gradient Echo
256x192/3	35.84	23.37	12.46	46.23
256x256/3	33.76	15.58	11.68	36.36
512x256/3	24.67	5.19	**2.98**	28.96
256x192/5	38.05	18.57	22.85	57.14
256x256/5	35.45	15.06	15.58	47.14
512x256/5	23.37	12.07	5.58	40.25
256x192/7	53.63	27.27	29.35	65.71
256x256/7	41.03	14.67	18.57	49.74
512x256/7	25.06	9.87	10.38	46.75

Table. 3 Volume overestimation factor (mm^3) in pulse sequence due to coil-induced susceptibility artifact actual size of coil volume (160 mm^3).

Matrix size/Slice thickness	Spin Echo	Fast Spin Echo	Inversion Recovery	Fast Gradient Echo
256x192/3	3.03	2.54	2.35	6.03
256x256/3	2.63	2.48	2.21	5.37
512x256/3	2.12	**1.9**	1.93	5.22
256x192/5	3.93	3.71	2.66	6.96
256x256/5	3.03	3.52	2.48	6.53
512x256/5	2.41	3.24	2.41	6.11
256x192/7	4.89	5.63	4.1	7.68
256x256/7	4.83	5.39	3.85	7.34
512x256/7	3.43	5.01	3.51	6.75

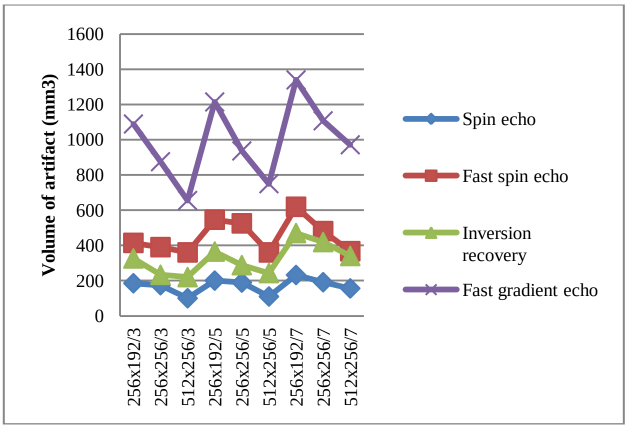

Fig.1 The signal loss volume (mm^3) versus pulse sequence, matrix size and slice thickness of volume of coil pack 160 mm^3**.**

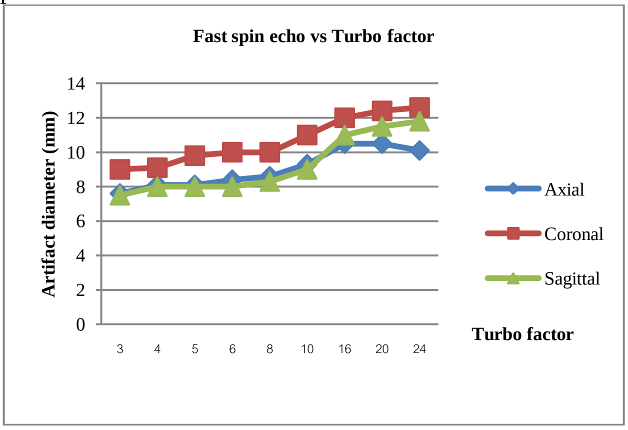

Fig.2 The artifact diameter of fast spin echo versus turbo factor in three orthogonal planes. Coil packs maximum dimension 5.6x5.6x9.0 mm.

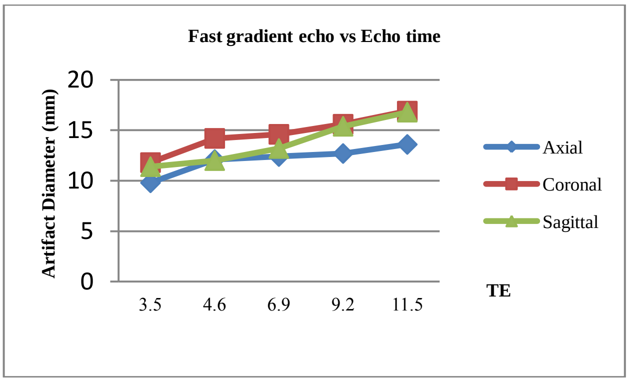

Fig.3 The artifact diameter of fast spin echo versus turbo factor in three orthogonal planes. Coil packs diameter 5.6x5.6x9.0 mm.

IV. Discussion

Platinum coils are non-ferromagnetic and paramagnetic materials with the potential to align along the direction of the lines of magnetic flux and display a force proportional to not only the gradient but also to the field strength [4]. However, one of the downsides of this flexibility is a greater complexity in terms of the choice of scanning parameters.

In this study, some of the major characteristic related to MRI imaging of coil packs have been defined. The main conclusions are 1.Spin echo (SE) and fast spin echo (FSE) with reduce turbo factor reduce the susceptibility artifact. A 180° refocusing pulse enables recovery of the transverse signal lost because of the inhomogeneity of the static magnetic field and bulk susceptibility differences. Substantial magnetic field inhomogeneity caused by bulk ferromagnetic materials result in additional local dephasing of the spins of hydrogen protons in randomly diffusing water molecules, and this diffusion-related dephasing is not recoverable with the application of a 180° refocusing pulse; 2.Susceptibility artifact increases with increased slice thickness; 3.Artifact increases with decreased matrix size; 4.Reducing the echo time (TE) is the main factor in improving artifact on fast spin echo and fast gradient sequence; 5.Artifact increases with increased coil packs density; and asymmetric artifact occurs in the frequency encoding direction on MRI images.

Walker et al [2] studied the platinum detachable coil packs at 1.5 T and 3.0 T with 3D TOF sequences and reported that imaging artifact were minimal when reducing the TE to approximately 3.5 ms and obtained high spatial resolution image.

There are several limitations in this study, including a static phantom, manual measurement, heterogeneity of the coil packs or artifact, and the acquisition time. Aneurysm model was a closed system with no flow: no estimation on the effects of flowing blood, particularly within the aneurysm, can be made. Also, without flow information, the effect of reducing the TE on the overall quality of the MRI and MRA images cannot be determined. The aneurysm models were measured manually, which can introduce error. Furthermore, the volume measurements were obtained by manual for each section once. The results described obtained by using specific image sequence as implemented on Philips Achieva equipment. Different results might be obtained with other equipment or vendors because of different gradient amplitude combinations employed in the sequence structure; however, the conclusions should remain the same.

V. Conclusions

Improved visualization of perianeurysmal soft tissue with endovascular coil is best accomplished by Spin echo sequences and are preferred significantly more ($P \leq 0.05$) than Fast spin echo, Inversion recovery and Fast gradient echo sequences.

Fast spin echo sequences are even better suited to reduce susceptibility artifact. Furthermore, shorter turbo factor and shorter effective TE in the latter sequences are beneficial for the same reason as sequences having shorter TE.

The rim of susceptibility artifact typically points in the frequency encoding direction (hyper signal). A smaller slice thickness and that through plane artifact decreased. Artifact would also be decreased as increasing the readout image matrix size.

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