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Phase Dependent Image Contrast Enhancement in MRI

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요 약 : 본 연구에서는 MRI 영상에서 위상을 조절하여 영상 대조를 증가 시키는 방법을 제안 하였다. 영상하고자 하는 물체 자체가 갖는 위상에 따라 영상의 대조가 변하기 때문에 본 방법은 자화율 영상이나 유속 영상에 유용하게 쓰일 수가 있다. 본 논문에서는 위상 분포에 따른 영상의 대조를 증가 시키기 위하여 RF 펄스를 복셀 내에서 위상을 갖도록 디자인 하였다. 따라서 복셀에서의 신호의 크기는 물체자체의 위상과 RF 펄스에 의하여 가해준 위상이 결합에 의하여 결정된다. 외부위상 변화에 따른 신호의 변화를 분석하였고 그때 디자인된 RF 펄스를 이용하여 자화율만의 영상과 유속만의 영상을 얻었다. 컴퓨터 시뮬레이션의 결과는 제안된 알고리즘이 복셀내에 위상을 갖는 물체의 영상에 유용하고 그런 물체만을 영상하는데 유용함을 보였다.

Abstract : An enhancement technique for phase dependent image contrast in MRI(Magnetic Resonance Imaging) is proposed. Because the method can enhance inherent phase contrast it is suited for susceptibility imaging and flow imaging where intravoxel phase is a source of image contrast. In this paper, applying external phase in the voxel enhances phase contrast. The external phase is generated by a tailored RF pulse so that one can control the phase contrast and even produces phase only contrast. Signal intensity due to both inherent phase and external phase is analyzed and the proposed technique is applied to a susceptibility effect only imaging and a flow effect only imaging. To verify the proposed technique, computer simulations are performed and their results are given.

Key words : Phase contrast in MRI, RF phase, Flow imaging, Susceptibility effect imaging

INTRODUCTION

Magnetic resonance imaging (MRI) as a medical diagnostic system provides various imaging modalities and thereby one can obtain various image contrasts depending on the imaging techniques[1]. Using the various image contrast a physiological information and even functional information as well as anatomical information of human body can be obtained in recent magnetic resonance imaging system[2-4]. Among image contrasts of MRI, T1 and T2 weighted contrast are well known and used in general diagnostic routine. The T1 and T2 contrast are dependent on the property of materials in the object and they are enhanced by pulse sequences designed for T1 imaging and T2 imaging [5,6].

Moreover magnetic field inhomogeneity and moving material in the object affect image contrasts. In magnetic field inhomogeneity, especially local field inhomogeneity due to susceptibility differences, the image contrast acquired is T2* weighted contrast[7]. A moving material such as blood flow also affects image contrast. Using time of flight or phase contrast techniques flow contrast is measured and enhanced[7-11].

Both local field inhomogeneity and the flow affect the phase distribution in a voxel of image. Besides of object-induced phase, a gradient pulse and a RF pulse affect the phase in a voxel in MRI. Normally the gradient pulse provides a linear phase in a voxel. To generate an arbitrary phase in a voxel, RF pulse is known to be more suitable [13].

The RF phase has been used for 2D slice-selection of 3D object in many imaging techniques in magnetic resonance

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imaging. For example, a linear phase is used for multi-slice imaging[14] and a nonlinear phase for the reduction of dynamic range of received signal in phase scrambling technique[15]. Recently, non-linear phase such as a quadratic phase is used to reduce the susceptibility effect. In gradient echo imaging, phase in an image voxel is affected by strong field inhomogeneity such as susceptibility effect[13]. By modifying the phase in a voxel one can get the susceptibility contrast imaging[13]. The built-in phase or inherent phase due to strong field inhomogeneity can be modified by the phase generated by the RF pulse. Namely by adding the phase in a voxel, the distribution of phase in the voxel is modified and thereby susceptibility inducing image contrast can be obtained[13].

In this paper, we propose a contrast enhancement technique using a RF pulse tailored to control phase distribution in a voxel. The usefulness of phase information in MRI and several imaging techniques using phase contrast are described. We focuses on enhancing image contrast of object where phase is main key to produce the contrast. The susceptibility imaging and flow imaging are shown as examples for the phase enhanced contrast techniques.

INTRAVOXEL PHASE IN MR IMAGE

An image obtained in MRI, in general, represents magnitude intensity showing proton spin density or T1/T2 weighted intensity. Signal intensity in the image is known to be a vector sum of spins in the voxel. Spins in the voxel distribute with different phase. If spins have same phase the vector sum of spins is maximized, i.e., the voxel produce maximum signal intensity. The phase term of spins in MRI is usually caused by a phase error arising from electronic system, magnetic field inhomogeneity and flow effect.

An intravoxel signal intensity in MRI can be written as

$$S(x,y,z) = \int_{voxel} m(x,y,z)\exp[j\phi(x,y,z)], \qquad (1)$$

where $m(x,y,z)$ is spins' magnetization distribution function and $\phi(x,y,z)$ shows spins' phase distribution. We assume that a voxel position is (x_i,y_j,z_k) and its size is Δx, Δy and Δz in x, y and z direction, respectively. Then the signal from a voxel can be rewritten as

$$S(x_i,\ y_j,\ z_k) = \int_{x_i-\Delta x/2}^{x_i+\Delta x/2} \int_{y_j-\Delta x/2}^{y_j+\Delta x/2} \int_{z_k-\Delta x/2}^{z_k+\Delta x/2} m(x,y,z) \exp[j\phi(x,y,z)]dxdydz, \qquad (2)$$

where we assume x-y is transverse plane and z is slice direction.

The system is assumed to be stable so that electronic phase error is negligible. Then the phase, $\phi(x,y,z)$ in equation (2) is mainly due to the object-induced term such as flow and susceptibility effects. This phase term gives a useful information of object, i.e., one can measure the susceptibility effect or the flow effect for an image. Normally it causes image artifacts such as geometrical position shift and deformation or signal degradation. Among those, the signal degradation due to the phase distribution in a voxel affects image contrast, i.e., the phase distribution is not constant so that spins in a voxel cancel each other by different their phases. The phase distribution in a voxel is affected by external phase by gradient pulses and RF pulses as well as the object induced phase such as field inhomogeneity and flow effect. In the following, image contrast due to intravoxel phase and its enhancement by RF pulses applied externally are described.

PHASE CONTRAST ENHANCEMENT

Adding a phase distribution externally can change the signal intensity in equation (1). By inserting the phase of $\phi_c(x,y,z)$, equation (2) can be rewritten as

$$S(x_i,\ y_j,\ z_k)\,|_{\phi_c} = \int_{x_i-\Delta x/2}^{x_i+\Delta x/2} \int_{y_j-\Delta x/2}^{y_j+\Delta x/2} \int_{z_k-\Delta x/2}^{z_k+\Delta x/2} m(x,y,z) \exp[j\phi(x,y,z)+j\phi_c(x,y,z)]dxdydz, \qquad (3)$$

The phase distribution in a voxel is changed to $\phi(x,y,z)+\phi_c(x,y,z)$. The changed phase distribution leads to the change of intravoxel signal intensity.

Normally 2D image in MRI has a slice thickness to improve SNR. The transverse resolution is typically 1 mm while slice thickness is 10 mm. The spatial resolution in transverse direction is, therefore, so small that phase variation within the transverse pixel is negligible. Moreover the spins' magnetization distribution function, $m(x,y,z)$ is assumed to be constant within a voxel because voxel size is so small that magnetization variation within a voxel is negligible. Therefore the intravoxel signal intensity could be mostly affected by phase variation in slice thickness. So equation (3) can be simplified as one-dimensional case, e.g., only slice selection direction of z-direction. Then the signal intensity can be rewritten as

$$S(x_i,\ y_j,\ z_k) = \int_{z_k - \Delta z/2}^{z_k + \Delta z/2} m(z; x_i, y_j)\exp[j\phi(z)]dz, \qquad (4)$$

where $m(z; x_i, y_j)$ is spins' magnetization distribution function at the pixel of (x_i, y_j).

To enhance image contrast, an external phase is generated along the slice selection direction (z-direction). A RF pulse is tailored so that it does not select only a slice but also puts a phase within a selected slice. By applying the tailored RF pulse which produce a phase, $\phi_c(z)$ in a voxel, the intravoxel signal becomes as

$$S(x_i, y_j,\ z_k)\,|_{\phi_c} = \int_{z_k - \Delta z/2}^{z_k + \Delta z/2} m(z; x_i, y_j)\exp[j\{\phi(z) + \phi_c(z)\}]dz, \qquad (5)$$

For normal voxel which has no inherent phase distribution from either susceptibility effect or flow effect, the phase distribution in a voxel is only the external phase generated by the RF pulse such as

$$S(x_i,\ y_j,\ z_k)\,|_{\phi_c}^{normal} = \int_{z_k - \Delta z/2}^{z_k + \Delta z/2} m(z; x_i, y_j)\exp[j\phi_c(z)]dz, \qquad (6)$$

Image contrast, therefore, can be represented by signal difference between the normal voxel and the voxel affected by inherent phase, i.e., the difference between equations (5) and (6).

By designing external phase $\phi_c(z)$, one can enhance a inherent phase contrast. For example, if external phase generated by RF pulse is $\phi_c(z) = c\,|\,x\,|$ the signal from normal voxel is zero because all spins cancel due to linear phase distribution along z-direction while the signal from the voxel with inherent phase is not zero and even enhanced because the total phase in the voxel is $\phi(z) + c\,|\,z\,|$. In the following, as inherent phase contrast enhancement by inserting external phase into a voxel, susceptibility effect enhanced contrast and flow effect enhanced contrasts are described.

1. Susceptibility Effect Contrast

A linear phase such as $\phi(z) = az$ is generated by strong field inhomogeneity such as susceptibility effect[13]. The

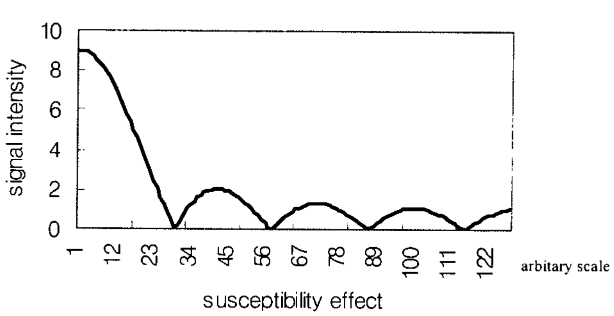


Fig. 1. The intravoxel signal intensity vs. susceptibility effect for normal voxel. As is seen, the signal intensity decreases as the susceptibility effect increases

steep of linear phase (a) indicates the strength of susceptibility effect. The signal intensity is affected by the susceptibility effect as shown in figure 1. Figure 1 is obtained by calculating the signal intensity of equation (4) as increasing a. As shown in the figure, signal intensity decreases as the susceptibility effect or linear phase (a) increases because the spins along the slice selection direction are dephased due to linear phase generated by the susceptibility effect. The signal difference between the voxel having the linear phase and the voxel having no phase is main source of image contrast due to the susceptibility effect. But this image contrast can not be differentiated from the T1 or T2 image contrast. Therefore the image contrast due to susceptibility effect normally is intermingled with conventional T1/T2 contrast so that it gives an artifact to the image.

To change image contrast for susceptibility induced-linear phase, a quadratic phase such as $\phi_c(z) = bz^2$ is added so that the total phase along the slice section direction is $az + bz^2$. As shown in figure 2, image intensity for linear phase a is changed, *i.e.*, the signal intensity does not decrease as linear phase increases in the region indicated in figure 2. This is because the quadratic phase generated by RF pulse can control the inherent susceptibility induced linear phase so that the signal degradation effect is compensated.

If a phase of $\phi_c(z) = c\,|\,z\,|$ is applied, on the other hand, the signal intensity distribution is changed as shown in figure 3. As seen in the figure, the signal from susceptibility affected region which has the linear phase in the voxel (*i.e.*, $a \neq 0$) enhanced while the signal from normal voxel is zero due to phase dispersion by external phase of $c\,|\,z\,|$. Since the signal from the normal voxels are all suppressed T1/T2 image contrast can be eliminated so that the image contrast with the phase of $\phi_c(z) = c\,|\,z\,|$ is only susceptibility in-

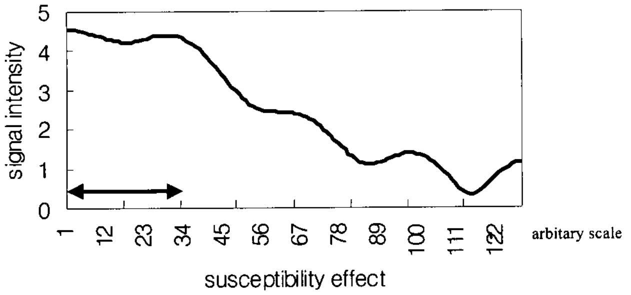


Fig. 2. The intravoxel signal intensity vs. susceptibility effect for external phase of $\phi_c(z)=bz^2$. As is seen, the signal intensity is independent of the susceptibility effect of the indicated range

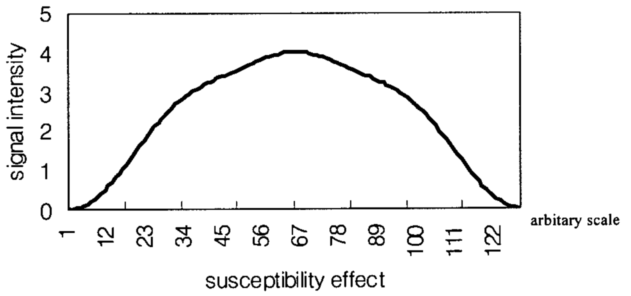


Fig. 3. The intravoxel signal intensity vs. susceptibility effect for external phase of $\phi_c(z)=b\,|\,z\,|$. As is seen, the signal intensity increases as the susceptibility effect increases

duced-phase contrast.

Figure 4 shows the signal difference between the voxel having linear phase due to the susceptibility effect and the voxel without inherent phase. First figure 4(a) is the case that external phase is not applied, i.e., $\phi_c(z)=0$. Figure 4 (b) and (c) are the case that the external phase are $\phi_c(z)=bz^2$ and $\phi_c(z)=c\,|\,z\,|$, respectively. Figure 4 (b) shows that one can compensate image intensity degradation due to the susceptibility effect by inserting a quadratic phase externally. Figure 4 (c) shows susceptibility only image contrast obtained by inserting a bilinear phase externally.

As is seen above, we can extract some useful imaging techniques. Namely using inserting quadratic phase along the slice selection direction, the image contrast is independent of the susceptibility effect. By inserting the bilinear phase in Figure 4(c) one can suppress the homogeneity region except the susceptibility region.

2. Flow Effect Contrast

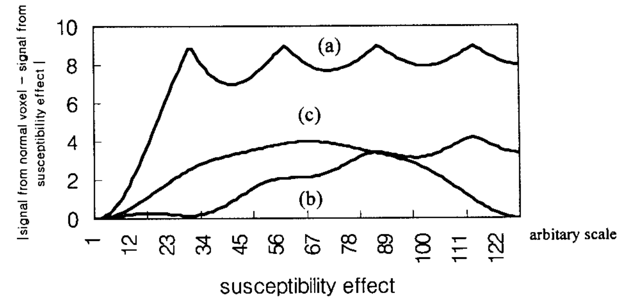


Fig. 4. The intravoxel signal intensity difference between normal voxel and susceptibility affected voxel for no external phase (a), external phase of $\phi_c(z)=bz^2$ (b) and external phase of $\phi_c(z)=c\,|\,z\,|$ (c)

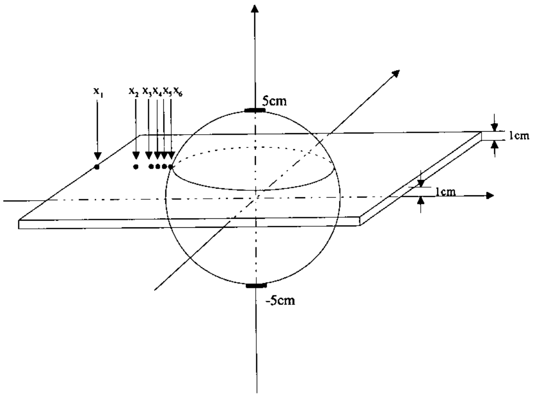


Fig. 5. The phantom for the susceptibility effect. The phantom consists of a sphere which is filled with air and surroundings of water. Image plane is selected 1 cm above at the center of sphere

The phase in a voxel of image is affected by a flow such as blood flow as well as the susceptibility effect. For the flow, signal intensity is also affected as equation (4). For the flow case, especially for a parabolic flow, the phase is represented as $\phi(z)=\frac{1}{2}v(z)t^2$ where $v(z)$ is flow velocity distribution along slice selection direction. For Laminar flow which is assumed as the blood flow in this paper, the flow velocity distribution is parabolic like $v(z)=v_m\left[1-\frac{z^2}{R^2}\right]$ where R is radius of blood vessel. Since the phase due to the flow velocity is varying along z-direction the intravoxel signal is affected as

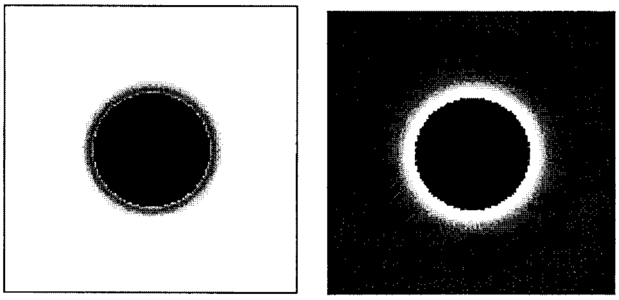
$$S(x_i,\ y_j,\ z_k)\,|_{\phi_c}=\int_{z_k-\Delta z/2}^{z_k+\Delta z/2} m(z;x_i,y_j)\exp\left[jv_m\left[1-\frac{z^2}{R^2}\right]\right]dz \tag{7}$$

The signal intensity due to the flow effect also can be changed by a tailored RF pulse which generates phase distribution of $\phi_c(z)$.

Normally the parabolic flow distribution generates parabolic phase distribution along z-direction. This cause intravoxel phase dispersion, i.e., signal degradation. By adding $\phi_c(z)$ using a tailored RF pulse flow affected signal can be enhanced. For example, if a phase of $\phi_c(z)=c\,|\,z\,|$ is applied the signal from a voxel with no flow effect is zero due to spins cancellation by external phase. However, by adding external phase into existing parabolic phase the signal from the voxel with flow effect is enhanced and thus enhancing flow effect contrast.

COMPUTER SIMULATIONS

To demonstrate the phase contrast enhancement by external phase generated by RF pulse, computer simulations were performed with a phantom. The phantom has a sphere as shown in figure 5. In the phantom, an air sphere with radius R (= 5 cm) is assumed to be located at the center of the homogeneous water bath. By solving the magnetic scalar potential, the external field B in z-direction is calculated.



Normal RF Tailored RF

(a) (b)

Fig. 6. Images obtained by (a) normal RF pulse and a tailored RF pulse with (b) external phase of $\phi_c(z)=c\,|\,z\,|$

An image plane is located at near the center of the sphere (1 cm above the z = 0 plane) in the z-direction is selected as shown in figure.5. The thickness of the slice and radius of the sphere were chosen to be 1 cm and 5 cm, respectively. The slice thickness is assumed to be a rectangular function. 32 spins are assumed in each voxel in the selected slice (in z-direction). Calculation shows that the spins in a voxel far from the center of the sphere or the interface field would be more or less a constant value, B0. The voxels near the interface regions between the air sphere and water, however, strong field gradients will be developed. In this interface region, even within a single voxel along the z-direction, spins will experience large field differences. The spin phase distributions of some voxels (e.g. x5, x6) near the interface region have stronger gradient than ones far from the interface regions (e.g. x1, x2) due to the stronger distribution of inhomogeneity or field gradients.

Figure 6 (a) shows an image obtained by normal RF pulse. As is seen, signal intensity near interfaces between air sphere and water is degraded due to the strong field inhomogeneity. The signal degradation due to the phase variation in a voxel is sometimes confused with T1/T2 weighted image contrast, i.e., image contrast due to the strong field inhomogeneity is not differentiated from that of other contrast sources such as T1 and T2 effect.

To enhance the phase contrast only, i.e., to enhance susceptibility only image contrast, a tailored RF pulse is applied, which generates external phase of $\phi_c(z)=c\,|\,z\,|$ in a voxel. Figure 6 (b) shows an image obtained by gradient echo sequence with the tailored RF pulse. As is seen, signal

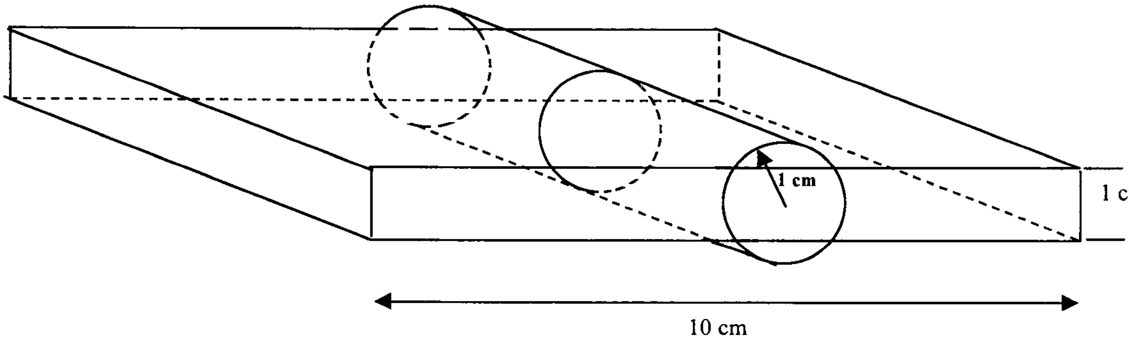


Fig. 7. The phantom for the flow effect. The phantom consists of a tube which

intensities near the boundaries are enhanced while other background water signal intensities are all suppressed. All the signal intensities with no inherent phase term in a voxel are suppressed due to external phase no mater what the signal is T1 or T2 weighted. Therefore one can get an inherent phase only contrast such as susceptibility effect only image contrast.

For flow contrast enhancement, a phantom is constructed as shown in figure 7. In the phantom, a tube with radius of 1 cm is located in static water bath. The tube has a flow with a parabolic velocity distribution. An image plane is located at the center of the tube (z=0) as shown in figure 7. The thickness of the slice is 1 cm. The slice thickness is assumed to be a rectangular function. 32 spins are assumed in each voxel in the selected slice (in z-direction). With parabolic velocity distribution in the tube, parabolic phase distribution is generated in the voxel of the tube.

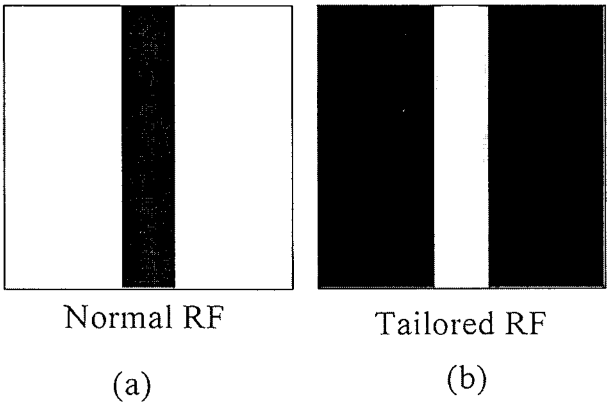

Normal RF (a) Tailored RF (b)

Fig. 8. Images obtained by (a) normal RF pulse and a tailored RF pulse with (b) external phase of $\phi_c(z)=c\,|\,z\,|$

Figure 8 (a) shows an image obtained by normal RF pulse. As is seen, signal intensity in the tube is degraded due to the parabolic phase distribution in a voxel. Flow signal is represented as dark line due to the phase dispersion in voxel. However when the tailored RF pulse which generates external phase of $\phi_c(z)=c\,|\,z\,|$ is applied the phase contrast, i.e., flow effect only image contrast, is enhanced. Figure 8 (b) shows an image obtained by the tailored RF pulse. As is seen, signal intensities in the flow tube are enhanced while other background water signal intensities are all suppressed. Finally the inherent phase only contrast such as flow effect only image contrast can be obtained.

As shown in figure 9 (a), signal intensity arising from the flow is degraded as increase of flow velocity when normal RF pulse is applied. Maximum phase due to the parabolic velocity distribution is increased from 0 to 32π Figure 9 (a) shows the signal intensity change at different positions in the tube. As is seen, signal intensity decreases as the maximum phase increase due to spin s dispersion. However, when a bilinear phase is applied in the voxel using tailored RF pulse, flow signal intensity increase as the maximum phase increases, thereby enhancing the phase contrast.

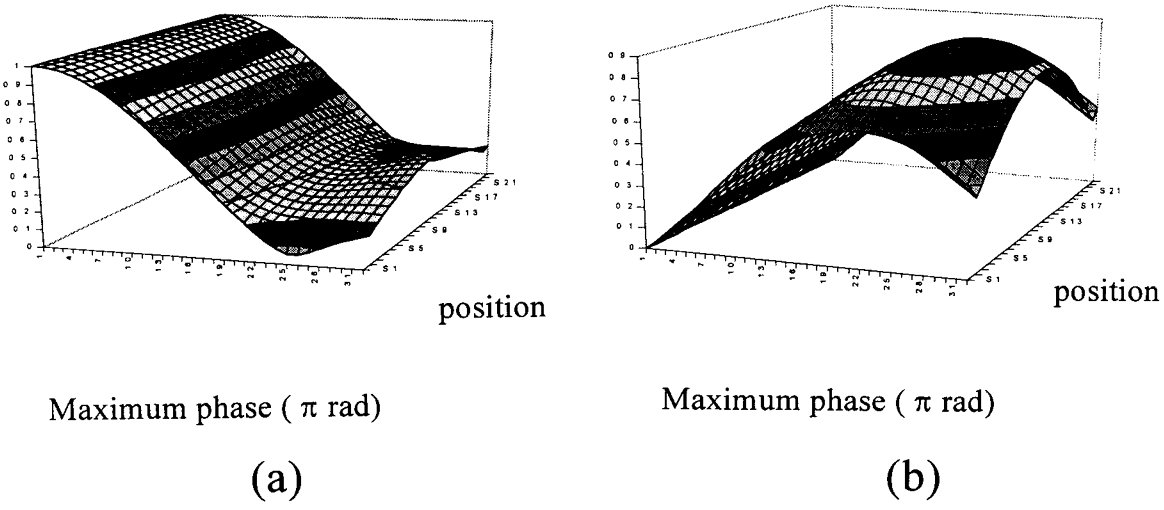


Fig. 9. Signal intensity distribution at different positions in the tube according to the increase of maximum phase in the case of no external phase (a) and the case of bilinear phase by the RF pulse (b)

CONCLUSIONS

In conclusion, image contrast enhancement in MRI is achieved using external phase generated by tailored RF pulse and this paper shows that the phase-dependent contrast can be controlled by external phase. In conventional, MR image contrast is mostly caused by T1 and T2. Using external phase of RF proposed in this paper, one could achieve that enhancement of contrast, especially in case of voxel having the phase due to susceptibility difference or flow. As a demonstration, two example of phase dependent imaging in MRI are shown such as susceptibility enhanced imaging and flow enhanced imaging. Computer simulations for susceptibility effect and flow effect were performed and the results showed that the external phase generated by tailored RF pulse is useful to enhance image signals affected by intravoxel phase.

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