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Dependence of RF Heating on SAR and Implant Position in a 1.5T MR System

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Purpose: We evaluated radiofrequency (RF) heating of a humerus implant embedded in a gel phantom during magnetic resonance (MR) imaging for the specific absorption rate (SAR), angle between the implant and static magnetic field (B_0), and position of the implant in the irradiation coil.

Methods: We embedded a stainless steel humerus implant 2 cm deep in tissue-equivalent loop and mass phantoms, placed it parallel to the static magnetic field of a 1.5T MR scanner, and recorded the temperatures of the implant surface with RF-transparent fiberoptic sensors. We measured rises in temperature at the tips of the implant by varying the SAR from 0.2 to 4.0 W/kg and evaluated RF heating of the implant for its angle to B_0 and its displacement along B_0 from the center of the RF irradiation coil.

Results: RF heating was similar for the loop and mass phantoms because the eddy current flows through the periphery of both. As the SAR increased, the temperature at the implant tip increased, and there was a linear relationship between the SAR and temperature rise. The values were 6.4°C at 2.0 W/kg and 12.7°C at 4.0 W/kg. Rise in temperature decreased steeply as the angle between the implant and B_0 surpassed 45°. In addition, as the implant was displaced from the center of the RF coil to both ends, the rise in temperature decreased.

Conclusion: The rise in temperature in deep tissue was estimated to be higher than 1.0°C for SAR above 0.4 W/kg. RF heating was greatest when the implant was set parallel to B_0 . In MR imaging of patients with implants, there is a risk of RF heating when the loop of the eddy current is formed inside the body.

Keywords: *MRI safety, metallic implant, RF heating, SAR, electromagnetic simulation*

Introduction

The main factors affecting human body function during magnetic resonance (MR) imaging are thought to be static magnetic field, switching magnetic field, and radiofrequency (RF) heating.^{1,2} The presence of a magnetic substance inside the body is absolutely forbidden in the static magnetic field. Nerve stimulation may be induced by the rapidly switching magnetic field. However, in clinical practice, the influence on human body function by the

static and switching magnetic fields is thought to be small. On the other hand, RF irradiation may produce heating in the human body.

Patients with implanted non-magnetic substances may suffer burn accidents from RF irradiation during clinical MR procedure as RF energy is converted to Joule heat inside the body. One index for measuring RF exposure dose is the specific absorption rate (SAR), which is defined as the absorbed electric power of RF per unit mass of human body (W/kg) and is expressed as $SAR = \sigma E^2 / \rho$, where σ is electrical conductivity of the tissue, E electric field induced by the RF, and ρ density of the tissue.³

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Resonance frequency increases as the magnetic field becomes higher. Recently, 3.0T MR scanners have been introduced in many hospitals, and fast spin echo (FSE) sequence using more RF pulses has become popular to shorten scan time. This imaging sequence uses a large number of RF pulses, and the duty cycle increases in proportion to the pulse width and number of excitations (NEX) in repetition time (TR) duration. In the environment of such a high magnetic field, SAR increases more by the FSE sequence than in an environment with lower magnetic field.

The Japanese Ministry of Health and Welfare established the SAR standard for MR imaging in Japan in 1991. The safety standard was set at 0.4 W/kg as the upper limit of whole-body averaged SAR. However, a review of the standard by the International Electrotechnical Commission (IEC) in 1995 that took into consideration the advances in MR scanners permitted SAR of 4.0 W/kg as the first-level operation mode.⁴ According to the IEC recommendation, the upper limit in the Japanese Industrial Standards (JIS) was also reviewed in 1999⁵ and provided for an upper limit of 2.0 W/kg for 6 min average as the usual operation mode and 4.0 W/kg for 15 min average as the first-level operation mode, which is the same as the IEC standard. However, risk of burn accidents remains during MR imaging following these standards,⁶⁻⁹ most resulting from contact between the skin of the human body and the cable of the electrocardiography monitor, oxygen color saturation monitor, or RF receiving coil.¹⁰ Knopp's group has also reported cases where burning occurred during MR imaging procedures without such environmental conditions.¹¹

On the other hand, MR imaging may not be possible in patients with metallic implants; many researchers have reported study of RF heating of such implants.¹²⁻¹⁴ Even under identical RF irradiation power, in a patient with such an implant, the uniformity of the electric field is disturbed near the implant, and the electric current density and energy absorption may greatly increase locally. Such local heating is termed a "hotspot." We experienced a case in which MR imaging had to be discontinued when the patient complained of pain in the vicinity of an implant. Although Shellock and colleagues have published a number of safety reports, there are no published guidelines about RF heating relating to MR imaging for patients with implants.^{15,16}

We previously evaluated the RF heating of a humerus implant embedded in tissue-equivalent phantoms¹⁷ and observed substantial heating at both ends of the implant, with maximum temperature

rise where there was large curvature at the tip. Concerning the depth for embedding an implant in a phantom, temperature rise was greater nearer the phantom's surface, which was found to reflect the outer-skin effect of RF power. Furthermore, loop formation enhanced heating. These results corresponded well with those of electromagnetic simulation based on the finite element method.

Therefore, in this study, we examined the dependence of RF heating on the variation in SAR using the same humerus implant embedded in new phantoms. A loop phantom and new mass phantoms were prepared to evaluate whether RF heating occurs without loop formation. We compared the results with those from electromagnetic-field analysis and investigated the dependence of RF heating on the angle between the implant and the static magnetic field and on the displacement of the phantom in the irradiation coil.

Methods

Phantom and implant

Tissue-equivalent phantoms were prepared as previously described¹⁷ following a modified procedure of Kato's group.¹⁸ For the phantoms, we dissolved powdery gelatinizer with 9% carrageenan (Cool Agar, Nitta Gelatin Inc.) in distilled water and mixed in a small amount of salt (0.19 wt%) to adjust the electrical property to be equivalent to that of soft tissue. For the large mass phantom, we used a polypropylene container 67 cm long by 35 cm wide as the vessel and added the above semi-solid gel to a depth of 14 cm as shown in Fig. 1a. The weight of the phantom was 35 kg. The loop phantom was the same used in our previous study, having an empty space 42 cm long by 10 cm wide in the central part (Fig. 1b). The weight of the phantom was 30 kg. A stainless steel (austenitic grade, ASTM F138) MD/N humerus nail 24 cm long (Zimmer Inc.) used to treat fractures in clinical practice was embedded parallel with the direction of a static magnetic field in both phantoms. The implant was positioned 2 cm deep from the phantom's surface and 2 cm distant from the phantom's outer wall on one side in consideration of the outer-skin effect of RF power (Fig. 2a, b). Moreover, we prepared a compact mass phantom 40 cm long by 20 cm wide to investigate the dependence of RF heating on the angle between the implant and the static magnetic field and on the displacement of the compact phantom in the irradiation coil (Fig. 2c). The weight of the phantom was 13 kg.

MR imaging exposure

All experiments were performed using a 1.5T

Fig. 1. Dimensions of tissue-equivalent gel-phantoms

a: Large mass phantom. The vessel is a polypropylene container 67 cm long by 35 cm wide. Semisolid gel was prepared with powdery gelatinizer (9% carrageenan) melted in distilled water, and the electrical property was approximated to muscular tissue. Its total weight is about 35 kg. **b:** Loop phantom. The outer container is the same dimensions as those of the mass phantom. The loop phantom has an empty space measuring 42 cm long by 10 cm wide in the central part.

MR scanner (MAGNETOM Symphony, SIEMENS) under the following RF irradiation conditions: coronal scan plane; pulse sequence, HASTE (half Fourier acquisition single-shot turbo spin echo); echo time (TE), 248 ms; TR, 1030 ms; field of view, 50 cm; slice thickness, 5 mm; matrix, 256×256 ; NEX, 32; number of slices, 25; number of echoes, 256; presaturation pulse, parallel anterior-posterior (A-P); and scan time, 15 min. We used a birdcage-shaped transmit/receive body coil, which contained 4 irradiation coils measuring 35×70 cm and 60 cm in inner diameter. These coils constituted 2 pairs of opposite coils for 90° phase-shifted exposure. To derive only the influence of RF irradiation, we did not apply the gradient magnetic field. The value of whole-body averaged SAR calculated by the IEC's pulse-energy method was displayed on the operator console for the weight of each phantom.⁴ The whole-body averaged SAR varied from 0.2 to 4.0 W/kg (upper limit of IEC first-level operation mode). Variation of the SAR was attained by adjusting the output power of the RF amplifier.

Temperature measurement

We measured temperature using 2 fiberoptic thermometers (FL-2000, Anritsu Keiki Inc.) as

Fig. 2. The humerus implant embedded in the gel-phantoms

a: Semi-solid gel was filled to a depth of 14 cm, and the implant was embedded 2 cm from the surface. **b:** The distance between the implant and the outer container edge is also 2 cm. **c:** A compact phantom 40 cm long by 20 cm wide was newly prepared to investigate the dependence of radiofrequency (RF) heating on various implant arrangements.

shown in Fig. 3a. These thermometers comprised optical fibers 10 m long and were not influenced by magnetic field and RF irradiation. The absolute values of the fiberoptic temperature sensors were calibrated with a thermocouple thermometer. The temperature was measured by forcing the sensors into close contact with the implant surface, and the difference between the values before and after RF irradiation was adopted as an experimental temperature rise (Fig. 3b). Measured positions for the mass and loop phantoms were 24 points at 1-cm intervals from the tip to the base of the implant (Fig. 3c). In contrast, in the experiment using the compact phantom, measured positions were only 2 points on the tip and base of the implant surface. The temperatures were recorded at one-second intervals from 15 min during to 10 min after RF irradiation. Room temperature was maintained in advance at 24°C, and the phantom was left there more than 4 hours prior to measurement.

Analysis of electromagnetic field

RF heating of the implant and phantoms was analyzed with pre- and post-processor Femap (UGS Co., Ltd.), and static- and transient-field analysis

software, PHOTO-EDDYj ω and PHOTO-THERMO (PHOTON Co., Ltd.), which were all based on the finite element method. The electric current of the RF irradiation coils was estimated from the RF power displayed on the MR console and the impedance of the coils. The resonance frequency of this MR scanner was 63.8 MHz, and the electric current flowing through each irradiation coil was 4.6 A at the maximum. PHOTO-EDDYj ω and PHOTO-THERMO software can calculate the direction of induced eddy current, electric current density, amount of heating due to Joule loss, and temperature distribution. In the electromagnetic-field analysis, the shape of the object, specific permeability, electrical conductivity, electric current, RF frequency, thermal conductivity, specific heat, mass density, and boundary conditions are given as input parameters. The parameters of the phantom and the implant required for the present analysis are previously reported.¹⁷

Experimental procedure

We performed this experimental study in 5 steps. (1) We measured the temperature of the implants in the phantoms of Figs. 1a and 1b at maximum RF

Fig. 3. Temperature measurement

a: Temperature was measured with 2 fiberoptic thermometers (FL-2000, Anritsu Keiki Inc.). **b:** These thermometers are composed of optical fibers 10 m long and are not influenced by the magnetic field and radiofrequency (RF) irradiation. **c:** Temperatures of the implant surface were measured at 24 points at 1-cm intervals from the tip to the base for the mass and loop phantoms.

power ($\text{SAR} = 4.0 \text{ W/kg}$) and compared the rises in temperature, and we performed electromagnetic-field analysis for the above conditions and qualitatively compared those results with experimental results. (2) We measured the rise in temperature for a compact phantom of Fig. 2c after 15-min RF irradiation by varying whole-body averaged SAR from 0.2 to 4.0 W/kg. (3) We investigated the dependence of the temperature rise on the implant position in the phantom and the angle between the implant and the static magnetic field. In the following measurements, the center of the implant coincided with the center of the static magnetic field. We defined the coordinate axes in Fig. 4, in which the origin is the center of the static magnetic field and the Z-axis is the direction of the static magnetic field. In the figure, the X-Z plane is parallel with the patient table and 15 cm above the table. Consequently, the direction of the Y-axis coincides with that of A-P arrangement. First, we embedded the implant at the center of the position-fixed mass phantom for the direction from 0° to 180° by 45° steps in the X-Z plane. Second, we embedded the implant parallel (0°) and anti-parallel (180°) in one

Fig. 4. Definition of coordinate axes. The origin of the coordinates is set at the center of the static magnetic field, and the direction of the static magnetic field is defined as the Z-axis. The X-Z plane is parallel with the patient table and is 15 cm above the table. Consequently, the direction of the Y-axis coincides with the anterior-posterior (A-P) direction. The center of the implant was set at the origin, and the compact phantom was turned 360° by 45° steps on the X-Z plane. In addition, the implant was inclined $\pm 45^\circ$ by 15° steps in the Y-Z plane.

side of the compact phantom of which position was fixed. Third, the entire compact phantom, including the implant in one side, was turned 360° by 45° steps in the X-Z plane. (4) We inclined the entire compact phantom $\pm 45^\circ$ by 15° steps in the Y-Z plane and measured the temperature rises for 7 inclinations. (5) Finally, we moved the phantom ± 50 cm by 10-cm steps from the center of the RF irradiation coil, which coincides with the center of the static magnetic field ($Z = 0$ cm), and measured temperature rises at 11 positions.

We performed all measurements except those of Step 2 at maximum SAR (4.0 W/kg), and the temperature rises of the tip and base of the implant were evaluated as the averaged value of 2-times measurements.

Results

Difference of RF heating with or without loop

Temperature rises of the tips of the implants embedded in the loop and mass phantoms are shown in Fig. 5a, where the abscissa is the elapsed time from the start of RF irradiation. Temperature rises at 15-min RF irradiation are also shown in Fig. 5b for the distance from the tips of the implants. As shown in Fig. 5a, in both cases, temperatures increased gradually after RF was turned on and rapidly decreased after RF was turned off. In both phantoms, the parts that heated most tended to be the tip and base of the implant, as shown in Fig. 5b. The maximum temperature rise was observed to be 12.3°C at the tip for the loop phantom and 11.4°C at that for the mass phantom at the time of 15-min RF irradiation.

We performed the simulation by electromagnetic-field analysis with PHOTO-EDDYj ω for both phantoms (Figs. 6a–d). Figure 6a shows the eddy current map for the loop phantom and Fig. 6b that for the mass phantom. The result for the loop phantom is the same as that previously reported.¹⁷ The eddy current generated in the mass phantom flowed densely in the region surrounding the implant and sparsely in the periphery and central region of the phantom (Fig. 6b). Figure 6c shows the temperature map for the loop phantom and Fig. 6d for the implant in the mass phantom. In Fig. 6c, large temperature rises were observed in the areas surrounding both ends of the implant. The largest temperature rises were observed at both ends of the implant, a similar finding to that for the implant in the loop phantom.

Dependence of RF heating on SAR

We investigated the dependence of RF heating of the implant on SAR by varying SAR from 0.2 to

Fig. 5. Temperature rises at the implant tip in the loop and mass phantoms
a: Temporal temperature changes from the start of radiofrequency (RF) irradiation. **b:** Spatial temperature changes just after 15-min RF irradiation.

Fig. 6. Simulated electromagnetic field obtained by PHOTO-EDDYj ω and PHOTO-THERMO
a: The eddy current map in the loop phantom. The eddy current is relatively large near the surrounding of the implant and is flowing toward the one-way direction. The unit of the current is A/m². **b:** The eddy current map in the mass phantom. The eddy current is flowing densely in the area surrounding the implant and sparsely in the periphery and central region of the phantom. The unit of the current is A/m². **c:** The temperature map for the loop phantom. The temperature change is largest near both ends of the implant, and considerable temperature rise is observed near the outer edge of the opposite side. **d:** The temperature map for the implant in the mass phantom. The temperature rises are larger at both ends of the implant than those in the central region. The maximum value is 12.3°C at the tip of the implant.

4.0 W/kg. The temperature rise of the implant tip is shown in Fig. 7, where the abscissa is whole-body averaged SAR, and the ordinate is the difference of temperature rises before and after 15-min RF irradiation. As a result, the temperature rise was in proportion to the increase of SAR and was well correlated between them (correlation coefficient: $r=0.998$, standard deviation: $\sigma=0.25^{\circ}\text{C}$). The temperature rise was 1.4°C for the SAR of 0.4 W/kg, which is the safety level prescribed by the Japanese Ministry of Health, Labour, and Welfare, and it was 6.4°C for 2.0 W/kg, which is the JIS-determined safety level.⁵ In addition, it was 12.7°C at 4.0 W/kg, which is the upper limit of the IEC's first-level operation mode.⁴

Dependence of RF heating on implant angle

Rotation in the X-Z plane

The dependence of RF heating on position and the angle between the implant and the static magnetic field was investigated under the maximum SAR of 4.0 W/kg. When the implant was situated in the center of the mass phantom, angular depend-

Fig. 7. Dependence of radiofrequency (RF) heating on specific absorption rate (SAR) in the mass phantom. The abscissa is whole-body averaged SAR, and the ordinate is the difference of temperature rises before and after 15-min RF irradiation. The SAR is varied from 0.2 to 4.0 W/kg. In the figure, the limits of the safety level prescribed by the Japanese Industrial Standards (JIS) for 6 min average and the first-level operation mode of the International Electrotechnical Commission (IEC) for 15 min average are also shown as dotted lines. There is good linear relationship between SAR and temperature rise.

ence was barely observed for the temperature rise of the implant tip. Next, when the direction of the implant in one side of the compact phantom was inverted, no difference in temperature rise was observed. The temperature rise of the implant tip was measured by varying the angle of the entire compact phantom from 0° to 180° in the X-Z plane. As shown in the results in Fig. 8a, for all angles, the temperatures increased gradually after RF was turned on and decreased rapidly after RF was turned off. Particularly, maximum temperature rise was 14.7°C when the direction of the implant tip was parallel with the static magnetic field (0°). However, the temperature rise decreased below 7.7°C as the angle was rotated above 90° . It is likely from the figure that there is an abrupt change in temperature rise between the 45° and 90° angles. In addition, maximum temperature rises at the tip and base of the implant are shown in Fig. 8b, where the angle was rotated clockwise in the X-Z plane to 360° by 45° steps. In Fig. 8b, the maximum temperature rises of the tip ($\square$) and base ($\circ$) for the same angle were measured simultaneously as a pair. The tip and base of the implant show nearly symmetric temperature changes with respect to 0° – 180° direction.

Rotation in the Y-Z plane

The implant was inclined from 0° to 45° by 15° steps in the Y-Z plane in the static magnetic field, and temperature changes of the implant tip are shown in Fig. 9a. Changes in temperature resembled those in the X-Z plane. The maximum temperature rise reached 15.9°C when the implant was parallel to the static magnetic field (0°) and decreased as the inclination increased. In addition, the implant was inclined $\pm 45^{\circ}$ by 15° steps in the Y-Z plane, and the maximum temperature rises of its tip and base are shown in Fig. 9b. The tip and base of the implant have similar temperature changes.

Dependence of RF heating on displacement

We moved the compact phantom ± 50 cm by 10-cm steps from the center of the irradiation coil ($Z=0$ cm) and investigated the dependence of RF heating on displacement. The maximum temperature rises of the implant tip and base after 15-min RF irradiation with maximum SAR (4.0 W/kg) are shown in Fig. 10, where the abscissa is the displacement, that is, the distance of the implant tip from the center of the irradiation coil. The figure shows that the maximum temperature rise reaches 15.1°C at the center. The rise decreased as the displacement enlarged and reached below 1°C at the edge of the irradiation coil of 35 cm. In addition, no RF heating was found outside the irradiation range of

Fig. 8. Dependence of radiofrequency (RF) heating on the rotation angle in the X-Z plane in the compact phantom. **a:** Temporal temperature rises of the implant tip were measured by varying the angle from 0° to 180° under the maximum specific absorption rate (SAR) of 4.0 W/kg. There may be a gap between the angles of 45° and 90°. **b:** The maximum temperature rises at the tip and base of the implant are shown where the angle is rotated clockwise to 360° by 45° steps. The temperature changes are nearly symmetrical with respect to 0°–180° direction.

Fig. 9. Dependence of radiofrequency (RF) heating on rotation angle in the Y-Z plane in the compact phantom. **a:** Temporal temperature rises of the implant tip were measured by varying the angle from 0° to 45° under the maximum specific absorption rate (SAR) of 4.0 W/kg. There may be a gap between the angles of 15° and 30°. **b:** The maximum temperature rises at the tip and base of the implant are shown where the angle is inclined $\pm 45^\circ$ by 15° steps. The tip and base of the implant show similar temperature changes. The value of temperature rise at the angle of -45° in the Y-Z plane is thought to include larger error because of the interference between the edge of the phantom container and the gantry wall of the magnetic resonance (MR) scanner.

70 cm, which corresponds to the length of the irradiation coil.

Discussion

In the experiments using mass and loop phantoms, there was little difference concerning heating

Fig. 10. Dependence of radiofrequency (RF) heating on displacement in the Z-axis in the compact phantom. The compact phantom was moved ± 50 cm by 10-cm steps along B_0 from the center of the RF irradiation coil. The maximum temperature rises of the implant tip and base just after 15-min RF irradiation with specific absorption rate (SAR) of 4.0 W/kg are shown. No RF heating occurs outside the irradiation coil 70 cm long.

parts and their temperature rises, which agreed with the simulation by electromagnetic-field analysis. That is, the eddy current in the mass phantom was smaller in the central region and larger in the periphery, and its map was similar to that of the loop phantom. When the implant was present in the loop of the eddy current, the current concentrated on both ends of the implant. Consequently, RF heating was thought to be caused by the Joule loss resulting from the sudden change of electrical resistance at the border between the implant and the phantom. In our previous study,¹⁷ eddy current maps were compared between loop and loop-cut phantoms, and interception of the loop was found to interfere with the flow of the eddy current, and the temperature rise was suppressed. However, the eddy current maps in this study show similar tendency to that of the previous loop phantom.

Dempsey and Condon supposed RF heating to be taking place in the MR imaging procedure by (1) electromagnetic induction, (2) formation of resonance circuit, and (3) antenna effect.¹⁹ In the present situations in which the implant is present in the flow of eddy current, these 3 mechanisms are thought to be related to RF heating. In addition, there was good linear relationship between the in-

crease of SAR and the temperature rise of the implant tip. Therefore, the value of SAR was thought to be an important factor and a good index of RF heating, although there may be greater or lesser differences according to the SAR calculation method of the MR scanner.²⁰ RF heating of the implant can be suppressed by decreasing SAR within the permissible range. However, the use of large-diameter transmit whole-body coil built into the MR scanner makes SAR large, except a transmit/receive coil with small diameter such as a knee coil.

Adair and Berglund have reported thermoregulatory reaction of the human body during MR imaging to be mathematically modeled, noting an increase in body temperature a maximum 0.6°C when RF irradiation was performed with whole-body averaged SAR of 4.0 W/kg.²¹ In addition, Czerski and Athey have reported that temperatures for local parts, such as 40°C for the head, 39°C for the torso, and 38°C for the extremities, do not harm the human body.²² Because the SAR is calculated based on weight, the operator performing the MR imaging is required to input accurate body weight on the console. Furthermore, the SAR of local parts of the body, such as the head, torso, or extremities, is calculated based on the input height. The SAR limits for these local parts of the body are also provided in detail.

In this experiment, we did not evaluate the SAR for local parts, but we demonstrated conspicuous temperature rise at the implant tip even within the safety level of whole-body averaged SAR. That is, the temperature rise of the implant tip, 6.4°C, surpassed the upper temperature limit at the SAR of 2.0 W/kg, which corresponds to the JIS upper limit of normal operation mode. Moreover, it was 12.7°C at the SAR of 4.0 W/kg, which corresponds to the JIS upper limit of the first-level operation mode. The temperature rise is considered to be suppressed by thermoregulatory reaction in the actual human body. Nevertheless, the temperature of local tissue probably exceeds 50°C in such condition, and this situation is very dangerous.

With implant placed parallel with the static magnetic field (0°) in the X-Z plane, temperature rises at both ends of the implant reached the maximum and particularly decreased steeply for angles over 45°. In this measurement, the center of the implant coincides with the center of the static magnetic field. In this situation, a large difference in temperature rise may be produced by the different arrangement among the implant, phantom, and RF irradiation coil. Similar results concerning the right/left difference in temperature rise was reported by

Baker's group for positioning deep brain stimulation lead.²³ Such a difference of RF heating is accredited to the different eddy currents generated by the RF irradiation for the phantom and the implant. However, further examination is needed to evaluate this phenomenon.

The distance from the center of the RF irradiation coil to the implant is considered to reflect the difference of RF power absorption for the phantom in the RF irradiation range. Within an RF irradiation coil 70 cm long, the temperature rise of the implant tip was maximum at the center and steeply decreased toward both ends of the coil. Furthermore, no RF heating occurred 40 cm from the center of the coil, which was outside the irradiation range. We evaluated the dependence of RF heating on SAR and displacement of the implant. In MR imaging of patients with implants, the increase of SAR and the position of the implant are important factors of RF heating, and burn accident is considered the result of various conspiring bad conditions.

Conclusion

In this study, RF heating occurred when the implant was present in the loop of the induced current and in the RF irradiation range in both the mass and loop phantoms. RF heating increases as SAR increases, and under such condition, there is danger even within the prescribed safety standard. RF heating decreased as the angle between the implant and the static magnetic field increased. When the implant was present outside the RF irradiation coil, there was little danger of RF heating. In addition, RF heating was greater when the direction of the implant was parallel to the static magnetic field and the implant was situated in one side of the phantom. In conclusion, we demonstrated that increased power of RF irradiation in MR imaging produced RF heating in a nonmagnetizing metal implant.

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