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Development and Testing of Implanted Carbon Electrodes for Electromagnetic Field Mapping during Neuromodulation

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Abstract

Purpose: Deep brain stimulation electrodes composed of carbon fibers were tested as a means of administering and imaging Magnetic Resonance Electrical Impedance Tomography (MREIT) currents. Artifacts and heating properties of custom carbon-fiber DBS electrodes were compared with those produced with standard deep brain stimulation (DBS) electrodes.

Methods: Electrodes were constructed from multiple strands of 7- μ m carbon-fiber stock. The insulated carbon electrodes were matched to DBS electrode diameter and contact areas. Images of DBS and carbon electrodes were collected with and without current flow and were compared in terms of artifact and thermal effects in phantoms or tissue samples in 7T imaging conditions. Effects on magnetic flux density and current density distributions were also assessed.

Results: Carbon electrodes produced magnitude artifacts with smaller FWHM values compared to the magnitude artifacts around DBS electrodes in SE and GE imaging protocols. DBS electrodes appeared 269% larger than their actual size in GE images, in sharp contrast to the negligible artifact observed in diameter-matched carbon electrodes. As expected, larger temperature changes were observed near DBS electrodes during extended RF excitations compared with carbon electrodes in the same phantom. Magnitudes and distribution of magnetic flux density and current density reconstructions were comparable for carbon and DBS electrodes.

Conclusion: Carbon electrodes may offer a safer, MR-compatible method for administering neuromodulation currents. Use of carbon-fiber electrodes should allow imaging of structures close to electrodes, potentially allowing better targeting, electrode position revision, and facilitate functional imaging near electrodes during neuromodulation.

Keywords

MR Image Artifact; Deep Brain Stimulation; Neuromodulation; Magnetic Resonance Electrical Impedance Tomography; MR Safety; Carbon Electrodes

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1. INTRODUCTION

Neuromodulation has gained popularity in the investigation of mechanisms underlying brain dynamics and in the treatment of neurological conditions (1,2). Deep Brain Stimulation (DBS) has been indicated for electrical regulation of brain networks responsible for Parkinson's disease, essential tremor, chronic pain, dystonia and epilepsy (3,4). MRI and Computed Tomography (CT) human head datasets are commonly fused to determine coordinates of DBS targets. DBS lead positions are commonly planned or verified using intra-operative CT (5) or pre-, post- or intra-operative MRI studies (6).

Typical DBS electrodes are composed of platinum-iridium (Pt-Ir) alloy, which has high magnetic susceptibility compared to surrounding tissues. Susceptibility differences between implanted metal objects and tissues can cause local magnetic field inhomogeneities and artifacts such as geometric distortion, fat suppression failure, signal loss and pile-up artifacts in MR images (7–10). Susceptibility artifacts scale with main magnetic field strengths and may obscure exact electrode boundaries on MR magnitude images (11). Safety issues caused by heating of metal objects, due to RF or gradient field switching, is a major limitation in imaging (12). Heating effects tend to be greater in long thin objects, such as leadwires or electrodes (13–15). Ongoing efforts to mitigate RF-induced heating include introducing RF chokes (16), modifying lead design (17), coiling wires (18), or designing new leads based on resistive tapered stripline technology (19). Despite these efforts, *in vitro* studies demonstrate risks of irreversible tissue damage due to DBS electrode heating (20,21) at 1.5T, even as field strengths of 3T and above are becoming standard (22) for brain and spine studies. Higher magnetic fields, coupled with higher gradients, accelerated imaging sequences and gradient slew rates could potentially worsen DBS-electrode susceptibility artifacts and tissue heating, necessitating a search for an MR-compatible alternative to metal implanted neuromodulation electrodes.

Carbon electrodes and leadwires have previously been used in ECG (23), fMRI (24), simultaneous EEG-MRI (25) and EEG-fMRI (26) studies to overcome the challenges of using metals in a magnetic field. Carbon-fiber electrodes have been successfully used for stimulation of epilepsy and subsequent fMRI studies (24). Carbon fiber nanotube yarns have also been tested as potential DBS implants, with the smaller susceptibility differences between carbon and surrounding tissues being found to reduce MR artifacts on T1- and T2-images compared to Pt-Ir electrodes (27,28).

DBS surgery involves uni- or bilateral implantation of DBS leads (e.g. Medtronic DBS 3387/3389, Medtronic, Minnesota, USA) to deliver current pulses to target structures (29). A DBS lead is composed of insulated Pt-Ir conductor wires and four Pt-Ir electrodes (29). DBS electrodes are commonly positioned at locations in the basal ganglia including the subthalamic nucleus or internal globus pallidus (3,4). Current injected through a Pt-Ir electrode ('source') may return to a pulse generator ('sink') in the subcutaneous pectoral area (29). DBS stimuli are tuned to individual patients, with pulses of 60–200- μ sec width, 1–5 V amplitude, and 120 – 180 Hz frequency (3,29). However, the mechanism underlying DBS is poorly understood (30). While extensive modeling of DBS fields and effect extent

have been performed (31,32) along with preliminary experimental determinations of DBS fields using MR techniques (33), DBS mechanisms remain largely unknown.

MR magnetic flux and electrical current density imaging may be useful in the measurement of effects caused by DBS current flow. Recently, Magnetic Resonance Electrical Impedance Tomography (MREIT) has been used to reconstruct *in vivo* conductivity, current density and electric field distributions using measurements of a single component of magnetic flux density (B_z), caused by the current administration, that is parallel to the main MRI field (34–36). Practical implementation of MREIT with clinically approved DBS systems would require no additional hardware. Measurements of electric field distributions caused by DBS stimulation nearby and throughout target structures may provide novel insights into the neuromodulatory mechanisms underlying DBS.

The novelty of this work relates to the characterization of carbon electrodes for MREIT imaging and stimulation as well as mapping current density distributions in DBS settings. Custom carbon-fiber DBS electrodes were developed and tested for MR magnetic flux and electrical current density mapping using MREIT. Carbon electrode performance is compared to diameter-matched Pt-Ir and clinical DBS electrodes at 7 T in terms of MR magnitude, phase and thermal properties, using uniform gel phantoms and anisotropic brain tissue samples. Impedance properties of carbon electrodes are also compared with DBS electrodes over a frequency range of 100 Hz–1 MHz to investigate whether carbon electrodes are suitable for current delivery in higher-frequency applications such as electroporation. It is anticipated that current density, electric field and conductivity imaging near carbon DBS electrodes may aid in the detailed investigation of DBS mechanisms, as well as facilitating conductivity-based direct functional imaging (37,38).

2. METHODS

2.1. Electrodes

In this study, a Pt-Ir DBS lead was compared with diameter-matched carbon-fiber DBS electrodes. Because the carbon-fiber thread used had a full diameter of 500 μm , electrodes with this width were combined to form sturdy 1-mm-diameter carbon-fiber electrodes. Electrode diameters were reduced to 250 μm for use in rodents. A Pt-Ir wire matching the 250- μm carbon electrode as closely as possible was chosen for artifact comparisons within brain tissue and gel phantoms. Electrode types and compositions are described in the sections below, and further details on electrode construction are described in the Supporting Information (Section S1, Figure S1).

2.1.1. One-contact (1C) carbon electrodes

2.1.1.1. Fabrication: Bundles of conductive carbon fibers (TC-33 3K, Formosa Plastics Group, Taiwan) were insulated using a three-stage dip-coating process: Nafion insulation, followed by immersion in two increasing concentrations of polyvinylidene fluoride (PVDF) in acetone. Fabrication stages of 250- and 500- μm one-contact (1C) carbon electrodes are shown in Figure 1(a) and Table S1.

The concentration and viscosity of PVDF solutions was increased from stage II to stage III of electrode fabrication, to protect Nafion-coated electrodes from buckling during immersion. Higher PVDF concentrations can be used with larger-diameter electrodes.

2.1.1.2. Quality Assurance: Before use, carbon electrodes were inspected under an optical microscope (Discovery V8 Stereo, Carl Zeiss Microimaging GmbH, Thuringia, Germany) to confirm electrode diameter, and check for cracks in insulation, stray carbon fibers and other obvious defects. The electrode insulation quality was electrically validated using two-terminal measurements of their electrical impedance in saline solution. Electrical validation and characterization tests used are detailed in the Supporting Information (Section S1.3, Figure S2). Impedance values were recorded as electrode depth in saline solution was increased in 3–5 mm increments. If the impedance at any depth deviated from 10% of the impedance measured with only the electrode tip immersed, electrode insulation was considered to be inadequate. The typical *closed-tip* impedance of 1C carbon electrodes (500- μm diameter) measured by the two-terminal test was around 100 k Ω at 1 kHz. Only electrodes with closed-tip impedance values >100 k Ω at 1 kHz were used in experiments.

2.1.1.3. Preparation for use in imaging experiments: Ends of 1C electrodes were exposed by cutting the end with scissors to construct either straight or beveled tip types. The insulative quality of open-tipped electrodes was validated using the methods described above. A 45°-beveled *open-tip* 1C carbon electrode (diameter 500- μm , 45-mm length, Figure 1(b)), had an impedance of around 100 Ω at 1 kHz.

2.1.2. Two-contact (2C) carbon electrodes

2.1.2.1. Fabrication: Two-contact (2C) carbon electrodes were constructed by combining an offset pair of 1C (500- μm diameter) carbon electrodes. The 2C electrode fabrication process is described in the Supporting Information (Section S1.2, Figure S1). The insulative quality of 2C carbon electrodes was measured using methods described in 2.1.1.2 above.

2.1.2.2 Preparation for use in imaging experiments: Electrode contacts were exposed by cutting a bevel through both electrodes. In the tests described in 2.2 below, both conductive sites were shorted for electrical current delivery to approximate DBS electrode areas. A beveled open-tip 2C carbon electrode with an approximate cross-sectional area of $1000 \times 500 \mu\text{m}^2$ (45-mm length) is shown in Figure 1(c).

2.1.3. Metallic wire electrodes (Pt-Ir)—Insulated Pt-Ir wire (203- μm diameter; 45-mm length; A-M Systems, WA, USA) was compared with diameter-matched 1C carbon electrodes (250- μm diameter; 45-mm length).

2.1.4. Deep Brain Stimulation lead (Pt-Ir)—A cylindrical quadripolar DBS lead (1.27-mm diameter Brain Neurostimulation Lead 3389, Medtronic Inc, Minnesota, USA) with four Pt-Ir ring electrodes was compared to 2C ($1000 \times 500 \mu\text{m}^2$ cross-sectional area) carbon electrodes. While it was not possible to measure the connector-to-electrode tip resistance directly for carbon electrodes, it was possible for the DBS lead. The lead resistance (connector-to-electrode resistance) was measured to be 41.5 Ω at 1 kHz. However,

the resistance of the DBS lead using the two-terminal method described in the Supporting Information (Section S1.3, Figure S2) was 42.4 Ω at 1 kHz. This two-terminal measurement was used as the reference resistance for DBS leads.

2.2. Samples

MRI and MREIT experiments were conducted using carbon or Pt-Ir electrodes placed in gel phantoms or fixed tissue samples.

2.2.1. Gel phantoms—Cylindrical gel phantoms shown in Samples 1–3 (Figure 2(a–c)) were prepared by combining 20 g Agar, 3 g NaCl, 0.5 g CuSO₄ and 1000 ml distilled water. The mixture was microwaved until air bubbles rose towards the top layer of the solution. The solution was allowed to cool at room temperature into a solid gel within a cubic chamber of side 5 cm. Two opposing inner surfaces of the cube were covered with copper plates. The two-terminal resistance (R) between the plates at 100 Hz was measured using a low-frequency impedance analyzer (HP4192A, Hewlett Packard, Palo Alto, CA). The gel conductivity (σ) was calculated using

$$\sigma = \frac{L}{RA}, \quad [1]$$

where L is the electrode separation (m) and A is the electrode cross-sectional surface area (m²). The area of the electrodes was assumed large enough that electrode impedances were negligible compared to gel resistances. Gel conductivity was calculated to be approximately 1 S/m at 100 Hz.

2.2.2. Phantom Configuration—In experiments using Samples 1–2 diameter-matched DBS 3389/2C or Pt-Ir/1C (‘source’) electrodes were placed within the phantom (diameter 55 mm, height 42 mm) and surface carbon (‘ground’) electrodes were placed on the phantom periphery. Current pulses (amplitude: 1 mA, total current duration: 20 ms, frequency: 4 Hz) were passed between source and ground electrodes. In Samples 1–3, care was taken to align Pt-Ir and carbon electrodes along the $\mathbf{B}_0$ (i.e. z -axis) during current injection to minimize stray field contributions from active electrodes.

In experiments using Sample 3, sinusoidal RF signals (amplitude = 0.13 mW, frequency = 298 MHz) amplified using a RF power amplifier (gain = 30 dBm) were transmitted into the sample via a custom-made RF surface coil that covered the upper half of the phantom. A maximum power (amplitude = 10 mW, SAR = 0.15 W/kg) was deposited in the center of the phantom. The field geometry was symmetric, and each electrode type was subject to similar B_1 field patterns. Fiber optic sensors (TS3, Optocon, Germany) were secured to DBS 3389/2C electrodes to measure temperatures at electrode tips. A third “central” sensor was placed equidistant from both electrodes, in the center of the phantom. Temperature sensors were connected to a temperature signal conditioner (FOTEMP-OEM-PLUS, Optocon, Germany) and temperatures at each sensor were recorded every 2 seconds over 30 minutes’ continuous RF application. More details of benchtop temperature experiments are provided in the Supporting Information (Section S2, Figure S3).

2.2.3. Paraformaldehyde-fixed rodent brain—A 250 g Wistar rat was euthanized by transcardial perfusion with saline (39) containing Heparin (1 unit/ml) and 4% Paraformaldehyde in 0.1 M Phosphate Buffer prior to surgery following an Arizona State University Institutional Animal Care and Use Committee-approved protocol. A 1C carbon electrode (250- μ m diameter) and an insulated Pt-Ir wire (203- μ m diameter) was implanted at a posterior inclination of 50° into the left and right hippocampus respectively (Sample 4, Figure 2(d)). The stereotaxic target coordinates were 4.20 mm lateral to midline, 3.40 mm anterior and 4 mm ventral to lambda. Current was injected from the 1C/Pt-Ir ('source') to surface carbon ('ground') electrodes. Care was taken to ensure stimulation electrodes (1C, 2C, DBS, Pt-Ir) in samples 1–4 were parallel to $\mathbf{B}_0$ (the z-axis) as shown in Figure 2(e).

2.3. Imaging Experiments

2.3.1. MRI Experiments—Imaging was conducted using a 12-cm bore Bruker Biospin 7 T MRI system (Barrow Neurological Institute, Arizona, USA). A 70-mm birdcage RF volume coil was used for imaging. MR data were acquired using both conventional spin echo (SE) and gradient echo (GE) pulse sequences.

2.3.2. MREIT Experiments—A constant current source (40) was used to synchronize monopolar current to a spin echo MREIT pulse sequence (SE-MREIT) via spectrometer transistor-transistor logic control signals (34). A schematic of a typical MREIT experimental setup is shown in Figure 2(e). In MREIT experiments, current was injected via carbon or Pt-Ir electrodes and surface carbon electrodes (cross-sectional area $10 \times 10 \text{ mm}^2$, Hurev, Republic of Korea). Complex k-space MREIT data resulting from two monopolar, biphasic current injections (I^+ , I^-) were collected, as shown in Figure 3. 'Monopolar' refers to a configuration having one electrode placed within a sample and a ground electrode on the sample surface. MREIT currents synchronized with spin echo sequences were reversed after the 180° RF pulse to align with re-phased spins to increase MR phase after the 180° refocusing RF pulse, and were therefore biphasic. Positive (or negative) current was injected for 9 ms between the 90° and 180° RF pulses followed by negative (or positive) current for 11 ms between the 180° RF pulse and start of the read-out gradient. The excess 2 ms current injection was intended to increase MREIT-phase changes using the ICNE pulse sequence (41) but may have produced some phantom composition changes. The imaging parameters used in MR and MREIT experiments are summarized in Table 1.

2.4. Data Processing

2.4.1. Field Inhomogeneity and MR Magnitude Image Analysis—Local field homogeneity in regions surrounding diameter-matched carbon and Pt-Ir electrodes were assessed using $\mathbf{B}_0$ and $\mathbf{B}_1$ field maps. $\mathbf{B}_0$ field maps were computed using MR phase images acquired using two different echo times (42). MR data acquired at two different tip angles was used to compute $\mathbf{B}_1$ field maps (43). Apparent sizes of diameter-matched carbon and Pt-Ir electrodes were measured on MR magnitude images. Electrode boundaries were estimated by computing FWHM values of magnitude profile plots through electrode centers. Ratios of FWHMs to actual electrode sizes were calculated to assess artifact extents associated with each electrode type.

2.4.2. MREIT Analysis

2.4.2.1. Experimental data processing (B_z): Raw data collected from positive and negative MREIT current injections were processed to complex MR images (34). Complex images from positive and negative MREIT current injections were divided to remove the systematic phase artifact and extract B_z (34,44,45), via

$$B_z(x, y, z_0) = \frac{1}{2\gamma T_c} \arg \frac{M_I^+(x, y, z_0)}{M_I^-(x, y, z_0)}. \quad [2]$$

2.4.2.2. Synthetic distributions ($\mathbf{J}^0, B_z^0$): Uniform-conductivity ($\sigma = 1 \text{ S/m}$) finite-element (FE) computational models constructed from experimental MREIT magnitude data were solved for synthetic magnetic (B_z^0) and current density ($\mathbf{J}^0$) distributions as follows. Experimental MREIT magnitude data were segmented using ScanIP (Simpleware Ltd., Exeter, UK) into source electrode, ground electrode and gel regions. The resulting meshes were exported to COMSOL 5.0 (COMSOL Inc., Massachusetts, USA). Tetrahedral element FE meshes were used to solve the Laplace equation for the voltage distribution (u) in a domain (Ω) using Neumann boundary conditions. The Laplace equation was solved in the form

$$\nabla \cdot (\sigma(\mathbf{r}) \nabla u(\mathbf{r})) = 0, \quad [3]$$

where $\mathbf{r} = (x, y, z)$ is a position vector in $\mathbb{R}^3$. Current density $\mathbf{J}$ was calculated using Eq. [4] based on the conductivity (σ) and electric field ($\mathbf{E} = -\nabla u$) distribution within Ω

$$\mathbf{J}(\mathbf{r}) = -\sigma(\mathbf{r}) \nabla u(\mathbf{r}). \quad [4]$$

The z -component of the induced magnetic flux density B_z in Ω was computed from the solved current densities in MATLAB using a Fourier transform version of the Biot-Savart law

$$B_z^0(\mathbf{r}) = \frac{\mu_0}{4\pi} \int \mathbf{J}(\mathbf{r}') \times \frac{\mathbf{r} - \mathbf{r}'}{|\mathbf{r} - \mathbf{r}'|^3} d\mathbf{r}', \quad [5]$$

where $\mu_0 = 4\pi \times 10^{-7} \text{ Tm/A}$ is the permeability of free space.

2.4.2.3. Projected current density ($\mathbf{J}^P$): Current flow distributions were reconstructed as the projected current density ($\mathbf{J}^P$) (46,47). Computation of $\mathbf{J}^P$ involved the combination of synthetic current and magnetic flux densities ($\mathbf{J}^0, B_z^0$) and experimental magnetic flux density (B_z) as described by

$$\mathbf{J}^P = \mathbf{J}^0 + \frac{1}{\mu_0} \left(\frac{\partial (B_z - B_z^0)}{\partial y}, -\frac{\partial (B_z - B_z^0)}{\partial x}, 0 \right). \quad [6]$$

3. RESULTS

3.1. Impedance characteristics of carbon and Pt-Ir electrodes

Two-terminal impedance magnitudes ($|Z|$) for carbon (1C and 2C) electrodes and Pt-Ir (DBS 3389) electrodes are shown in Figure 4. Data plotted for 1C and 2C electrodes were averaged over 6 samples of each type (one trial each), while measurements for the single DBS lead were averaged over six trials. As expected, 1C electrodes ($|Z| = 2570 \Omega$, $R = 1375 \Omega$), with the smallest contact surface area were found to have higher impedance and resistance values, with the resistance of this electrode type being around $1.4 \text{ k}\Omega$ at 100 Hz , approximately $6.5\times$ the resistance of the thicker 2C electrodes ($|Z| = 437 \Omega$, $R = 210 \Omega$). While both carbon electrode types had larger resistances and impedances than the DBS lead, the frequency dependence of all electrodes was broadly similar.

3.2. Field mapping and MR magnitude artifacts surrounding carbon and Pt-Ir electrodes

Local $\mathbf{B}_0$ and $\mathbf{B}_1$ field inhomogeneities surrounding diameter-matched carbon (2C) and Pt-Ir (DBS lead 3389) electrodes in sample 2 are shown in Figure 5. The $\mathbf{B}_0$ field map in Figure 5(b) remains similar nearby both electrode types. Figure 5(c) demonstrates larger spatial extent and scale of $\mathbf{B}_1$ inhomogeneity surrounding Pt-Ir electrodes compared to carbon electrodes. $\mathbf{B}_1$ inhomogeneities had an extent $3.32\times$ around Pt-Ir electrodes and $0.74\times$ around carbon electrodes compared with actual electrode sizes.

Artifacts surrounding diameter-matched carbon and Pt-Ir electrodes in Samples 1 and 2 are shown in Figure 6. FWHM values for electrodes in SE and GE acquisitions are summarized in Table 2. In general, apparent sizes of carbon electrodes were smaller than those of diameter-matched Pt-Ir electrodes. Substantial differences between apparent sizes were noted between the 1C and Pt-Ir wire electrodes, with apparent sizes of Pt-Ir electrodes being $21.19\times$ in SE and $15.62\times$ in GE images and apparent sizes of 1C electrodes being $2.59\times$ in SE, and $1.81\times$ in GE images.

3.3. Temperature changes near carbon and Pt-Ir electrodes

The temperature evolution recorded in sample 3 under the influence of continuous RF sinusoidal signals (amplitude = 10 mW , frequency = 298 MHz) transmitted through the surface coil is shown in Figure 7. Initial temperatures recorded from each sensor were subtracted and values presented here show two-minute moving-averages of temperature changes over the 30-minute recording period. Temperatures around Pt-Ir electrodes increased by 0.9°C during recording whereas those around carbon electrodes increased by 0.1°C under similar RF energy deposition conditions. The temperature at the central temperature probe steadily increased, with a final temperature ranging between those of carbon and Pt-Ir electrodes.

3.4. Magnetic flux (B_z) and current density (J^P) distributions

Reconstructed magnetic flux density (B_z) and current density magnitude (J^P) distributions established within homogeneous gel phantoms and fixed tissue samples during SE-MREIT are shown in Figure 8. Synthetic magnetic flux density (B_z^0) and current density magnitude

($\mathcal{J}$) were combined with experimental B_z to reconstruct $\mathcal{J}^P$ distributions. The slice shown was chosen close to exposed electrode areas. Active electrode areas (1C, 2C, DBS) were masked as precisely as possible to minimize noise contribution from electrode-artifact regions. Slight misalignments of stimulation electrodes (1C, 2C, DBS) contributed to the formation of dipoles in measured B_z distributions. B_z dipoles around the active DBS electrode (Figure 8(a)) were not observed in B_z^0 , possibly because the cylindrical geometry of the DBS electrode led to field cancellation. Possible mismatches in electrode orientation (Figures 8(b–d)) between experimental and simulation conditions led to differences in orientations of B_z and B_z^0 dipoles. The scales of B_z and $\mathcal{J}^P$ distributions were similar for diameter-matched 2C-G and DBS-G configurations (Figure 8(a–b)). Current densities surrounding the cylindrical DBS lead were symmetric. However, the $\mathcal{J}^P$ distribution around 2C electrodes was relatively asymmetric. Local current density magnitudes (arrows) around non-current-carrying DBS electrodes were larger than for 2C carbon electrodes. Error percentages in $\mathcal{J}^P$ reconstructions in samples 1, 2 and 4 were 9.9%, 5.16% and 6.44% respectively.

4. DISCUSSION

DBS currents have traditionally been administered via Pt-Ir electrodes implanted within disease-specific brain targets. Both safety factors caused by electrode heating and metal related artifacts may result when subjects with implanted electrodes undergo MR imaging. Even if imaging is performed safely at 1.5 T — the current safety standard for patients implanted with DBS leads (48), artifacts may prevent clear images being obtained near areas targeted by electrodes. The novelty of this work relates to the characterization of carbon electrodes for MREIT imaging and stimulation as well as mapping current density distributions in DBS settings using carbon-fiber DBS electrodes. Artifact effects related to each electrode type were compared under SE and GE scanning conditions, whereas thermal effects were studied under continuous RF energy deposition. It was found that the artifact and thermal characteristics contributed by carbon electrodes were overall smaller than those caused by Pt-Ir electrodes. The smaller temperature changes around non-current-carrying carbon electrodes, along with the finding that impedances, as well as B_z and $\mathcal{J}^P$ images created by these electrodes were similar or superior to those obtained using Pt-Ir electrodes, suggests that implanted carbon electrodes may both be suitable for safe delivery of neuromodulation currents in MRI environments and allow imaging of neuromodulation currents. Therefore, as anticipated, use of carbon electrodes may potentially make MR-guided surgical placement of DBS electrodes easier, enable routine MRI scans to be carried out, and facilitate studies of DBS dosimetry and mechanism via MREIT.

4.1 Impedance characteristics of carbon and Pt-Ir electrodes

Characteristic differences between electrode types could be explained using equivalent electrical circuits of the two-terminal test configuration, shown in Supporting Information (Section S1.3, Figure S2). Circuit properties depend on the inline resistance of the electrodes and properties at electrode-electrolyte interfaces. Contributions to the measured impedances due to properties of the copper reference were neglected because of the comparatively large area of copper exposed to the saline solution. As expected, impedances and resistances of

smaller carbon electrodes (1C: 250- μ m diameter) were largest. Resistances of 1C carbon electrodes were around $6.5\times$ those of 2C electrodes, which was consistent with the approximate ten-fold decrease in electrode surface area. Substantial differences between electrode resistivities ($\rho_{Carbon(1C,2C)} \approx 2\times 10^{-5} \Omega\cdot m$ (49), $\rho_{Pt-Ir(3389\ DBS)} \approx 2\times 10^{-8} \Omega\cdot m$ (50)), resulted in inline carbon electrode resistances being much larger than those of DBS electrodes, dominating overall impedance property comparisons. The overall trend of decreasing electrical impedance and resistance observed in all three electrodes with increasing frequency, as shown in Figure 4, was in accordance with behavior expected at electrode-electrolyte interfaces (51). The detailed properties of carbon electrode-electrolyte interfaces were not examined here because the exact areas of carbon electrode ends were not precisely known.

Because both the conductivity and geometry of carbon and Pt-Ir electrodes varied greatly, resulting electrode impedances were also very different. The 2C-ground circuit had a 210 Ω resistance compared to the DBS-ground value of 59 Ω . Direct replacement of existing Pt-Ir (DBS) electrodes with carbon (2C), would therefore result in a $3.5\times$ reduction of non-rechargeable battery life. If carbon electrodes were implemented, their impedance could be reduced by changing their conductive geometry. Alternatively, use of rechargeable batteries (52–55) may improve the longevity of DBS systems using carbon electrodes.

4.2 Field inhomogeneity and MR magnitude artifacts around carbon and Pt-Ir electrodes

B₁ field inhomogeneity around carbon (2C) electrodes had lower spatial extent compared to diameter-matched Pt-Ir (DBS 3389 lead) electrodes as shown in Figure 5. As expected, the local homogeneity of RF (i.e. **B₁**) fields around metal electrodes were affected to a greater extent than around carbon electrodes, potentially leading to larger MR magnitude artifacts around Pt-Ir electrodes as observed in Figure 6. Carbon consistently showed smaller MR artifacts compared to diameter-matched Pt-Ir electrodes as demonstrated in Table 2. Artifact extent improved by 12% around carbon compared to diameter-matched Pt-Ir (1C/Pt-Ir wire: $2.59\times/21.19\times$ in SE, $1.81\times/15.62\times$ in GE) electrodes, in agreement with similar studies (24,26–28). Therefore, reduced carbon artifacts are expected to allow easier localization of carbon compared to Pt-Ir electrodes on MR images.

Differential susceptibilities at electrode-medium interfaces produce field inhomogeneities, thereby contributing to artifacts surrounding electrodes (7–9,24,27). Eddy currents induced by gradient switching and/or the RF field are known to contribute to susceptibility artifacts (10). The paramagnetic susceptibility of Pt-Ir (at 90% platinum and 10% iridium) was reported as 230 ppm ($\chi_{Pt-Ir} = 230$ ppm) (28). The diamagnetic susceptibility of PAN (polyacrylonitrile)-based carbon, in sharp contrast with Pt-Ir, was reported as –26 ppm ($\chi_{Carbon} = -26$ ppm) (28). In sum, the lower susceptibility of carbon led to smaller artifacts surrounding carbon electrodes on MRI images.

4.3 MRI-induced heating mechanisms influencing carbon and Pt-Ir electrodes

Changes in temperature around carbon electrodes during the course of RF energy deposition were smaller than Pt-Ir electrodes, as shown in Figure 7. Overall, larger temperature increases were observed around Pt-Ir electrodes than for nearby carbon electrodes or the

central probe, demonstrating the expected RF heating caused by the presence of metal (12,13,26). Temperatures recorded from sensors next to the Pt-Ir DBS lead increased faster and attained larger values at the end of continuous RF energy deposition compared to carbon (2C) electrodes. Sensors measured both the temperature at electrode tips and that of gel in contact with these electrodes. While the thermal conductivity (k) of Pt-Ir (31 W/mK) is greater than carbon (1.5 W/mK) materials (56), RF heating of Pt-Ir electrodes likely exceeded any changes caused by heat conduction mechanisms. Furthermore, the substantially lower conductivity of carbon reduced the extent of antenna effects and gradient-related heating effects. Magnitudes of local temperature increases around Pt-Ir electrodes (0.9°C) exceeded carbon electrodes (0.1°C) by 8%. The central probe showed an intermediate increase in temperature, reflecting the larger B_1 values in this location.

RF heating near Pt-Ir electrodes in commercially-available DBS leads has been found to be influenced by the length, orientation, and presence of loops in the lead, with heating effects associated with looped DBS leads greater than for straight leads parallel to the longitudinal axial direction of the coil (14,15,57). In the benchtop experiments performed here, the DBS lead orientation was maintained parallel to the axial direction of the simulated-bore, and excess lead length was not looped. Magnitudes of temperature changes observed in gel-phantom experiments (0.9 °C) were of the same scale as those (1–5 °C) reported in excised porcine heads with similar DBS lead orientation (14,15). The total length of DBS lead used in these benchtop experiments was 40 cm, which was reported as the length associated with maximum heating (57). Smaller temperature increases around Pt-Ir electrode contacts were observed in this experiment, because of the much lower power used here.

Both this study and those reported in (14,15) used continuous wave RF excitation. *In vitro* safety studies using fast and high SAR sequences such as turbo spin echo and echo planar imaging have found temperature increases < 2°C at Pt-Ir electrode contacts (58–60). Overall, RF heating induced temperature increases around Pt-Ir electrodes under optimum conditions such as non-looped and parallel leads, observed in benchtop experiments were consistent with previous *in vitro*, *in vivo* and *ex vivo* MR safety studies.

4.4 Magnetic flux density component (B_z) and current density magnitude (J^P) distributions

Similar magnetic flux (B_z) and projected current density magnitude (J^P) distributions were produced using DBS-G and 2C-G electrode configurations (Figure 8(a–b)). Rectangular 2C electrodes were surrounded by asymmetrical J^P distributions whereas cylindrical DBS electrodes produced more symmetrical J^P distributions. Smaller 250-μm/270-μm diameter 1C carbon electrodes (Figure 8(c–d)), were used in homogeneous gel (Sample 1) and heterogeneous rodent brain (Sample 4) samples. B_z and J^P values were eight-fold larger in the fixed rodent head than the gel phantom because of the reduced effective electrode area and rodent head volume. As expected, magnitudes of J^P distributions surrounding electrode tips were large in all samples. Contact impedances of Pt-Ir wire and gel/tissue interfaces exceeded contact impedances of 1C carbon electrodes in both Samples 1 and 4. Currents could not be injected between Pt-Ir wire and ground electrodes, therefore J^P mapping was not possible.

Within gel phantoms (Figure 8(a–c)), the diameter-matched and non-current-carrying Pt-Ir electrodes were clearly visible in B_z and J distributions established using current-carrying carbon electrodes (1C-G, 2C-G). B_z and J values around these non-current-carrying yet highly conductive Pt-Ir electrodes were much larger than for carbon equivalents. Concentrations of current around Pt-Ir electrodes resulting from RF or magnetic-gradient-induced currents or MREIT currents may present a risk of tissue burns. Therefore, lower-conductivity carbon may be a safer alternative for electrode or implant materials in MRI environments.

4.5 Clinical and Experimental Implications

Carbon may provide a safer alternative to Pt-Ir due to improved magnetic field homogeneity, smaller MR magnitude artifact extent and RF-heating-induced changes observed near carbon electrodes. However, all safety hazards of Pt-Ir DBS electrodes are also potentially possible using carbon, albeit with reduced severity, and rigorous safety testing must be performed. Post-surgical image analysis near DBS electrode sites may aid in the early diagnosis and treatment of complications, as well as allowing verification of electrode placement. In other clinical imaging procedures, MR magnitude artifacts observed around Pt-Ir electrodes may obscure image features, particularly in fast MR pulse sequences such as echo planar imaging and would also indicate benefits from the replacement of Pt-Ir with carbon in DBS leads. Furthermore, the ability to perform MR imaging in the immediate surroundings of carbon electrodes will allow for DBS field mapping within neural structures directly affected by DBS currents, enabling experimental studies exploring neuromodulatory mechanisms underlying DBS.

5. CONCLUSIONS

Carbon electrodes had superior MR artifact and thermal performance, yet similar impedance properties to conventional DBS electrodes. The smaller thermal changes observed nearby carbon electrodes indicate that they may allow subjects to safely undergo routine MR imaging procedures. The carbon-fiber DBS electrodes developed in this work may be appropriate for use with fast imaging sequences and stronger main magnetic fields, potentially enabling DBS-fMRI studies and improved intra- and post-operative monitoring. Improved imaging near electrodes combined with spatial mapping of MR magnetic flux and electrical current density around carbon electrodes may allow experimental determination of current density within target brain structures and aid in probing mechanisms underlying DBS.

Supplementary Material

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Figure 1: Carbon electrodes.

(a) Fabrication stages of carbon electrodes (b) One-contact (1C, 500-μm diameter) and (c) Two-contact (2C, dimensions $1000 \times 500 \mu\text{m}^2$) carbon electrodes.

Figure 2: MREIT samples and experimental setup.

(a) Sample 1: Pt-Ir wire/1C (b) Sample 2: DBS 3389/2C along with ground (G) electrodes
 (c) Sample 3: DBS 3389/2C and optic-fiber temperature probes in cylindrical gel phantoms
 (d) Sample 4: Pt-Ir wire/1C and ground electrodes in fixed rodent brain (e) Schematic of MREIT experimental setup of sample 1. Magnetic flux density distributions caused by synchronized current pulses injected from source (S) to ground (G) electrodes were mapped in the imaging slice (dashed rectangle).

Figure 3: Spin echo-MREIT (SE-MREIT) pulse sequence.

Positive (I^+) or negative (I^-) MREIT currents were used interchangeably in the SE-MREIT pulse sequence.

Figure 4: Electrical characteristics of carbon (1C, 2C) and Pt-Ir (DBS 3389) electrodes.

Impedance ($|Z|$) or Resistance (R) of electrodes as a function of frequency are represented with solid and dashed lines respectively for 1C (250- μm diameter, black) and 2C carbon electrodes (dimensions $1000 \times 500 \mu\text{m}^2$, blue) and a single DBS 3389 electrode (1270- μm diameter, red). Mean ($\pm\text{SE}$, standard error) values measured for one trial of six samples of each carbon electrode type are shown. Measurements for the DBS lead were sampled six times and averaged to produce mean and SE values. Lengths of all carbon electrodes were approximately 45 mm.

Figure 5: B_0 and B_1 field maps.

(a) MR magnitude image showing Pt-Ir (DBS 3389) and carbon (2C) electrodes placed within cylindrical gel phantom. (b) B_0 (Hz) and (c) B_1 field maps showing flip angle variations (degrees) and vertical profiles through DBS 3389 (red) and carbon (black) electrodes.

Figure 6: Susceptibility artifacts.

Comparison of MR magnitude image artifacts around carbon (1C, 2C) and Pt-Ir (wire, DBS 3389) electrodes during spin echo (SE) and gradient echo (GE) acquisitions. (a) One-contact (1C, 250- μm diameter) carbon and Pt-Ir wire (203- μm diameter) (b) Two-contact (2C, dimensions $1000 \times 500 \mu\text{m}^2$) and Pt-Ir DBS 3389 (1270- μm diameter) electrodes placed in cylindrical gel phantoms (Samples 1 and 2).

Figure 7: MR-related thermal effects.

Local temperature changes around carbon (2C) and Pt-Ir (DBS 3389) electrodes under continuous RF excitation (amplitude = 10 mW, gain = 30 dB, frequency = 298 MHz). Inset shows a cylindrical gel phantom (diameter 55 mm, height 42 mm) with both electrode types (Sample 3), and temperature sensors with solid circles. The progression of local temperature changes at temperature-sensing tips under continuous RF excitation around carbon (yellow), Pt-Ir (blue) electrodes and a central (red) sensor over 30 minutes is shown.

Figure 8: Magnetic flux and current density distributions established by synchronized current injections during SE-MREIT.

MR magnitude (first column), reconstructed magnetic flux density (B_z , second column), synthetic magnetic flux density (B_z^0 , third column), synthetic current density magnitude (J^0 , fourth column) and projected current density magnitude (J^P , fifth column) distributions established via current injections through electrode pairs (a) DBS-G; Sample 2 (b) 2C-G; Sample 2 (c) 1C-G; Sample 1 (d) 1C-G; Sample 4 are shown. Electrode sizes: One-contact (1C) carbon electrodes (250- μm diameter in Sample 1, 270- μm diameter in Sample 4), Two-contact (2C) carbon electrodes (dimension $1000 \times 500\text{-}\mu\text{m}^2$ in Sample 2), platinum-iridium (DBS 3389) electrodes (1270- μm diameter in Sample 2). White arrows indicate non-current-carrying electrodes.

Table 1:

MRI and MREIT Imaging Parameters. Pulse sequences (SEQ) used were Spin echo (SE) and Gradient echo (GE). Abbreviations of imaging parameters: Repetition time (TR), Echo time (TE), Slice thickness (z), Number of averages (NAV), MREIT imaging current (Current) and total scan time (Scan time). For Sample 2, the imaging parameters used for B_0 mapping (first and second rows), B_1 mapping (third and fourth rows) and MR artifacts (fifth and sixth rows) are shown.

Sample No.	Imaging Parameters								
	SEQ	TR/TE (ms)	FOV (mm ²)	Matrix size	z (mm)	No. of slices	NAV	Current (mA, ms)	Scan time (mm:ss)
1	SE	1491/20	60 × 60	128 × 128	1/3	40/14	2/6	0/1, 0/20	06:21/12:48
	GE	450/6	60 × 60	128 × 128	1	40	8	0, 0	07:40
2	SE	1400/20	60 × 60	128 × 128	1	40	1	0, 0	06:21
	SE	1400/25	60 × 60	128 × 128	1	40	1	0, 0	06:21
	GE ^a	3500/6	60 × 60	128 × 128	1	40	4	0, 0	08:32
	GE ^b	3500/6	60 × 60	128 × 128	1	40	4	0, 0	08:32
	SE	1000/22	60 × 60	128 × 128	1/3	40/14	1/6	0/1, 0/20	06:21/12:48
	GE ^c	501/6	60 × 60	128 × 128	1	40	4	0, 0	08:32
4	SE	1000/22	40 × 40	128 × 128	2.5	10	6	0/1, 0/20	12:48

^{a,b} Flip angles used in GE sequences for B_1 field mapping were 40° and 80° respectively.

^c Flip angle used in GE acquisitions for MR magnitude image artifacts was 30°.

Table 2:

FWHM extents of artifacts surrounding diameter-matched carbon and Pt-Ir electrodes for spin echo (SE) and gradient echo (GE) sequences.

Electrodes	Actual diameter (mm)	Voxel size (mm)	Apparent diameter (FWHM, mm)	
			Spin Echo	Gradient Echo
1C carbon	0.27	0.47	0.70 (2.59×)	0.49 (1.81×)
Pt-Ir wire	0.21	0.47	4.45 (21.19×)	3.28 (15.62×)
2C carbon	1.40	0.47	1.88 (1.34×)	1.41 (1.00×)
DBS 3389	1.27	0.47	2.34 (1.84×)	4.69 (3.69×)