Control of intravascular catheters using an array of active steering coils

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Purpose: To extend the concept of deflecting the tip of a catheter with the magnetic force created in an MRI system through the use of an array of independently controllable steering coils located in the catheter tip, and to present methods for visualization of the catheter and/or surrounding areas while the catheter is deflected.

Methods: An array of steering coils made of 42-gauge wire was built over a 2.5 Fr (0.83 mm) fiber braided microcatheter. Two of the coils were 70 turn axial coils separated by 1 cm, and the third was a 15-turn square side coil that was $2 \times 4 \text{ mm}^2$. Each coil was driven independently by a pulse width modulation (PWM) current source controlled by a microprocessor that received commands from a MATLAB routine that dynamically set current amplitude and direction for each coil. The catheter was immersed in a water phantom containing 1% Gd-DTPA that was placed at the isocenter of a 1.5 T MRI scanner. Deflections of the catheter tip were measured from image-based data obtained with a real-time radio frequency (RF) spoiled gradient echo sequence (GRE). The small local magnetic fields generated by the steering coils were exploited to generate a hyperintense signal at the catheter tip by using a modified GRE sequence that did not include slice-select rewinding gradients. Imaging and excitation modes were implemented by synchronizing the excitation of the steering coil array with the scanner by ensuring that no current was driven through the coils during the data acquisition window; this allowed visualization of the surrounding tissue while not affecting the desired catheter position.

Results: Deflections as large as 2.5 cm were measured when exciting the steering coils sequentially with a 100 mA maximum current per coil. When exciting a single axial coil, the deflection was half this value with 30% higher current. A hyperintense catheter tip useful for catheter tracking was obtained by imaging with the modified GRE sequence. Clear visualization of the areas surrounding the catheter was obtained by using the excitation and imaging mode even with a repetition time (TR) as small as 10 ms.

Conclusions: A new system for catheter steering is presented that allows large deflections through the use of an integrated array of steering coils. Additionally, two imaging techniques for tracking the catheter tip and visualization of surrounding areas, without interference from the active catheter, were shown. Together the demonstrated steerable catheter, control system and the imaging techniques will ultimately contribute to the development of a steerable system for interventional MRI procedures. © 2011 American Association of Physicists in Medicine. [DOI: 10.1118/1.3600693]

Key words: steerable catheter, Interventional MRI, endovascular interventions, catheter tracking

I. INTRODUCTION

Endovascular interventions are primary treatments for cardiovascular diseases such as atherosclerotic stenoses and brain aneurysms. These are typically minimally invasive procedures guided by x-ray angiography where the interventionalist has to steer and position a catheter to a specific target

location in order to initiate treatment. Steering and positioning the catheter in the target vessel is a challenging task, and it is highly dependent on the practitioner's ability to maneuver the catheter through the vasculature. The difficulty of the procedure depend upon vessel tortuosity, location, and size

of the pathology to be treated, among other factors, which can make the procedure time consuming and consequently, may expose the patient to increased radiation doses.¹

Many methods have been developed to attempt to overcome the difficulty of guiding catheters through complicated vasculature. One method uses noncontact guidance of a magnetic seed implanted in a canine brain via manipulation of dc magnetic fields and x-ray fluoroscopy visualization.² This initial work resulted in the magnetic stereotaxic system (MSS) that consists of a magnetically compatible x-ray fluoroscopy system surrounded by three orthogonal superconducting coils that generate an externally controllable magnetic field.³ Even though the MSS has been approved for human clinical procedures, its use has not been broadly extended due to its size, the need for important modifications to the traditional x-ray room, and the limited number of procedures that can be performed with this technology.

Interventional Magnetic Resonance Imaging (IMRI) has been a field of intensive research since the 1990's. Open and short bore scanners, more flexible software, real-time sequences and other improvements in MR systems have made it possible to perform several interventional procedures in an MR scanner; e.g., guided biopsies, thermal ablation, stenting, and others.⁴ Among the main advantages of IMRI are the lack of ionizing radiation, potential for 3D visualization, intrinsic tissue contrast, sensitivity to flow, as well as depiction of diffusion and detection of tissue temperature information.

Besides these standard clinical benefits, earlier work has shown that the strong magnetic field in an MR scanner offers a unique environment for driving devices.^{5,6} Based on Lorentz's force, Roberts *et al.* have demonstrated technical feasibility of developing a catheter, whose tip can be remotely oriented within the magnetic field of an MR scanner by applying a dc current in a single axial coil.⁵ While this work provided an exciting proof of concept, it failed to address several practical problems, which limit the clinical applicability of the method. Specifically, the deflections of the catheter tip depend on the initial relative position to the external magnetic field. Additionally, the catheter "torqueability" (a term used by interventional radiologists to describe the ability to rotate the catheter along its long axis) is reduced since a large axial coil is needed to attain clinically acceptable deflections. Consequently, high dc currents had to be applied to the coil, which increased heat generation and led to potentially dangerous temperatures at the catheter tip.

Alternatively, Martel *et al.* demonstrated the feasibility of steering iron microparticles by the magnetic force generated by the field gradients present in a MR scanner.⁶ This method is potentially interesting for guidance of untethered magnetic devices; however, the required electric power is very high (and may be higher than the gradient system limit), so that an external gradient system used only for propulsion would be necessary.^{7,8} Additionally, the propulsion sequence added to the standard imaging sequence may limit the kinds of images that can be generated during propulsion.

One must also have the ability to not only steer the catheter but also visualize its current position and orientation. To achieve this, several MR-guidance techniques have been developed based on fast imaging sequences.^{9,10} These imaging sequences can be combined with two general methods for catheter tracking: active and passive tracking. In active tracking, the RF (radio frequency) signal detected by a receiver coil incorporated in the catheter tip is used for localization of the catheter via a series of 1D Fourier transformations typically acquired during orthogonal readouts along the X, Y, and Z axes.^{11,12} On the other hand, the passive method uses signal voids created by the absence of resonating spin in the catheter material or through intravoxel dephasing. In one implementation, dephasing is achieved via local field distortions from paramagnetic markers attached to the catheter tip.¹³ In another implementation, passive visualization signal voids can be adjusted by creating a local magnetic field inhomogeneity by passing current through conductors incorporated in the catheter wall.^{14,15}

Thus, it is clear that having the ability to freely deflect the tip while tracking the device in the 3D coordinate system of the MR system may provide some benefits for many catheter based interventions. The aim of the present work is to extend the earlier work of Roberts *et al.* and to use the magnetic field of the scanner to create the driving force for steering interventional devices. To overcome some of the previous limitations of the concept, we propose integrating an array of steering coils into the catheter in order to generate an arbitrary 3D magnetization vector that will allow higher and more complex deflection and guidance of the catheter tip in space. In addition, we also demonstrate practical methods for imaging the catheter and surrounding areas while the catheter is deflecting. In a real interventional procedure, this will allow tracking of the catheter tip while obtaining clear images of the anatomy surrounding the catheter.

II. MATERIAL AND METHODS

II.A. Torque and deflection of a steerable catheter

The design of the steerable catheter is based on two basic parameters: (i) the origin of the torque that causes the catheter tip to deflect and (ii) the behavior of the catheter under the application of this torque. It can be shown that applying a dc current to a coil of wire will generate a magnetic moment perpendicular to the surface of the coil loops. The interaction of this vector with the static magnetic field will generate a torque at the catheter tip

$$\begin{aligned}\vec{T} &= \vec{m} \times \vec{B}_0 \\ |\vec{T}| &= |\vec{m}| |\vec{B}_0| \sin \theta = NIAB \sin \theta,\end{aligned}\quad (1)$$

where $\vec{m}$ is the magnetization vector, $\vec{B}_0$ is the main magnetic field, N is the number of turns of the coil, I is the current through the coil, A is the area of the coil, and θ is the angle between $\vec{m}$ and $\vec{B}_0$ [Fig. 1(a)]. For a 3D coil array located at the catheter tip (with separate coils in each of the

FIG. 1. (a) Diagram of a small deflection (δ) that results from the force (F) applied in the catheter tip due to the interaction of the magnetic moment (M) generated by an axial current loop and the static magnetic field of the scanner (B_0) (b) In a 90° deflection condition there is not net magnetization perpendicular to the static field and an additional force source (F') is necessary to generate further deflection.

primary axes of the catheter), the resultant total magnetization vector in the catheter reference frame will be

$$\vec{m} = N_1 I_1 A_1 \vec{n}_u + N_2 I_2 A_2 \vec{n}_v + N_3 I_3 A_3 \vec{n}_w, \quad (2)$$

where n_u , n_v , and n_w are the normal vectors of the coil array elements.

Independent current control for each coil allows modulation of the different components of the resulting magnetization vector. Translating this result to the laboratory reference frame and assuming the static magnetic field is along the z -axis, only the magnetization components orthogonal to this direction will generate torque on the catheter tip

$$\tau = -m_x B_0 \vec{n}_y + m_y B_0 \vec{n}_x. \quad (3)$$

Assuming that at a given time the catheter axis system is rotated an angle α around the x -axis, β around the y -axis, and γ around the z -axis, then the magnetization component in the laboratory reference frame can be represented as

$$\begin{bmatrix} m_x \\ m_y \\ m_z \end{bmatrix} = (R_x(\alpha) R_y(\beta) R_z(\gamma))^{-1} \begin{bmatrix} m_u \\ m_v \\ m_w \end{bmatrix}, \quad (4)$$

$$\begin{aligned} R_x(\alpha) &= \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos(\alpha) & \sin(\alpha) \\ 0 & -\sin(\alpha) & \cos(\alpha) \end{bmatrix}, \\ R_y(\beta) &= \begin{bmatrix} \cos(\beta) & 0 & -\sin(\beta) \\ 0 & 1 & 0 \\ \sin(\beta) & 0 & \cos(\beta) \end{bmatrix}, \\ R_z(\gamma) &= \begin{bmatrix} \cos(\gamma) & \sin(\gamma) & 0 \\ -\sin(\gamma) & \cos(\gamma) & 0 \\ 0 & 0 & 1 \end{bmatrix}, \end{aligned} \quad (5)$$

where $R_x(\alpha)$, $R_y(\beta)$, and $R_z(\gamma)$ are the corresponding rotation matrices and m_u , m_v , and m_w are the magnetization components in the catheter tip reference frame.

In terms of catheter response to an applied force, previous work formulated the catheter deflection as a classical-cantilever problem governed by the Euler–Bernoulli law.^{5,16–18}

Applying the appropriate boundary conditions, the equation for tip deflection can be written as

$$\delta(x) = \frac{\tau L^2}{2EI} \Delta(x) = 0, \quad (6)$$

where τ is the applied moment or torque that results from applying a force at the catheter tip, L is the beam length, E is the Young's elastic modulus, I is the second moment of inertia, and Δ the axial displacement. This equation establishes a linear relation between torque and deflection; however, several assumptions limit its validity. These include the assumption of a small deflection angle, homogenous and isotropic materials, weightless beam, etc. Clearly the most questionable assumption for the present application is that of a small deflection angle, since the catheter should be made to traverse realistic blood vessels, and thus, large deflections are to be expected. For this reason, a large deflection theory for a very flexible beam, which establishes a nonlinear relation between deflection and torque and nonzero axial displacement would be more appropriate.

Several studies^{19–21} have performed rigorous mathematical analysis of this complex problem to find an accurate expression for deflection. The exact analysis is beyond the scope of this work; however, it should be clear that Eq. (6) cannot be used for accurate prediction of large deflection, and therefore, in the present work it is used as starting point for design consideration only.

As the deflection approaches 90° (and beyond), an alternative method of deflection needs to be considered. For example, Fig. 1(b) shows the situation when the catheter has reached a 90° deflection. In this case, a force in the vertical direction will clearly not add any deflection to the catheter tip. However, an additional source of force in a perpendicular direction can generate torque and further deflection. One possible way to achieve this is to add additional sets of coils along the length of the catheter, such that the small deflection assumption may be valid for each individual set, while still providing the necessary large deflection for the catheter as a whole. One could then drive these coils sequentially exploiting their relative position to the main magnetic field. Figure 2 shows a simplified case where coil two is initially not activated. As the device deflects through 45°, this coil is

FIG. 2. Simplified diagram of driving steering coils sequentially (a) Only coil one is active and generates a magnetization vector (M) that interact with the static magnetic field (B_0) and produce deflection. (b) At 45° deflection coil two reached a position to add to the deflection and it is turn on, now the total magnetic moment is the combinations of the magnetization produced by each coil. (c) At 90° deflection coil one axis is now parallel with the static magnetic field and should be turn off.

moved to a position that allows it to add to the deflection. The effect of coil one is also diminished at this point, such that it should be deactivated by the end of the deflection. This multicoil approach forms the basis for the device presented here.

II.B. Steerable system

In order to control the magnetic torque, and therefore the catheter deflection, we designed a system to control the current through the combined sets of coils. To achieve 3D catheter steering, we have implemented an independent current

control for each of the coils in the array. Figure 3(a) shows the basic components of the system that makes this possible: PC, serial interface, microprocessor, current source, catheter, and the magnet where all experiments were carried out. To design the current sources, a pulse width modulation (PWM) technique was implemented due to its reliability for digital control and its design simplicity for the required current level. The outputs of the PWM current source were controlled by a microprocessor, which communicated directly with a host computer running MATLAB. The MATLAB routine generated serial data [Fig. 3(b)] that controlled the PWM to dc current source. The PWM current source consists of four

FIG. 3. (a) System schematic diagram. (b) Current value, current polarity and coil/s are set from the serial data sequence generated by MATLAB routine and transmitted from the main PC to a microprocessor. (c) Current bridge controlled by the PWM generated by microprocessor, and (d) Coil array made of 42-gauge wire and set on a 2.5 Fr braided microcatheter.

power NMOS transistors (STD10NF10) connected in an H-bridge configuration as shown in Fig. 3(c). A control signal enables M1 and M3 for a current polarity while the amplitude of the current is modulated by the duty cycle of a 20 kHz control signal applied to the MOSFET gates. When the inverse current polarity is desired then M1 and M3 are kept off, while M2 and M4 are enabled. Since the catheter coils have a small inductance, a large inductor was added in series in order to filter the switching transients that would otherwise appear at the load. This inductance is the dominant load of the bridge, and its value was chosen to minimize the current ripple under steady state conditions. At each side of the load, an additional set of capacitors, resistors and diodes ($C_1R_1D_3$ and $C_2R_2D_4$) was added to provide an alternative pathway for the load current when the MOSFETs are turned off, and at the same time to limit the transitory voltage peaks that are always present when switching voltage over a large inductor. Each current bridge (one per coil) was controlled with only two signals: (i) one signal that enables the odd labeled MOSFETs at one signal level, and enables the even labeled MOSFETs at the other level and (ii) a PWM signal that modulates the current amplitude. These signals were generated by the microprocessor (PIC18F4550, Microchip) and passed through a high frequency NMOS H-Bridge Driver (HIP4081, Intersil) to the gates of the MOSFET bridges.

For generation of an arbitrary 3D magnetization vector, arbitrary current values and current polarities are required in each coil. Based on the present system, the latter translates to arbitrary duty cycle values and bridge leg enable signals. Since the system is designed to work in real-time, dynamic control of the previously mentioned variables was implemented. The data for the bridge configuration were loaded in MATLAB and sent through the COM serial port to the microprocessor through an RS-232 serial interface. The data were transmitted asynchronously at 9600 baud (9600 bits/s) as described in Fig. 3(b).

II.C. Catheter

Three coils made of heavy insulated 42-gauge wire (Polyimide-ML-NEMA MW16-C, MWS Wire Industries) were built over a 2.5 Fr (0.83 mm) fiber braided microcatheter (Renegade, Boston Scientific) to create the catheter prototype shown in Fig. 3 (d). Two of these were 70 turn axial coils separated by 1 cm, and the third was a 15-turn square side coil that was $2 \times 4 \text{ mm}^2$. The resulting length of the steerable catheter was 18 mm long. The axial coils increased the outer diameter by less than $200 \mu\text{m}$, while the side coil (with 2 mm width) had the largest impact on the dimensions of the distal end of the catheter tip.

As mentioned previously, the additional axial coil was added in order to improve the deflection by sequentially exciting both axial coils to achieve a particular tip orientation. Separating one coil from the tip can potentially provide a greater total deflection than a single larger coil at the tip due to the nonlinear behavior of such a system at large deflections. Here, we placed this coil at a position that main-

tains the length of the curvature to that expected from the natural bending of the catheter from a force at its tip. Alternatively, by exciting this separate coil with the reverse polarity when compared to that applied at the tip coils, one could control the straightness of the catheter tip as desired. In our configuration, the length of all coaxial copper wire leads from the current control systems to the coils were minimized to avoid RF heating due to standing wave effects;^{22–24} they were also tightly twisted to avoid open loops that could cause undesired generation of torque.

II.D. Temperature measurement of the catheter tip

To evaluate dc heating on the catheter tip, the temperature of one axial coil was measured on the benchtop while exciting it continuously with 100 mA dc current. We used a type K thermocouple (Extech TP870-50 to 250°C) connected to a multimeter (FLUKE 189 true RMS multimeter) and in close contact with the coil using thermal grease.

II.E. Tip deflection measurement

To measure the maximum tip deflection, we applied the coil excitation sequence shown in Fig. 4. The diagram shows the ideal situation where at the time each coil is excited, its magnetization vector is perpendicular to the main magnetic field. In this situation, the maximum magnetic torque is possible as described in Eq. (1). In the present open loop system, there is no tip position feedback to allow automated readjustment to meet the ideal condition at the time each coil is excited. This is critical in the case of the side coil since its axis has to be initially aligned with the main magnetic field in order to add deflection and not tip rotation. Therefore, the method shown here will only give an estimation of the maximum deflection that could be attained. These measurements were made at a maximum current value of 100 mA per coil. Image based data were obtained with a real-time gradient

FIG. 4. Current excitation diagram for tip deflection experiment. The coil excitation time T_i is set in MATLAB and ideally at this time the resulting magnetization vector is perpendicular to the main magnetic field to generate maximum torque.

FIG. 5. (a) Modified GRE sequence with suppressed rewinding gradients and (b) Sequence diagram of imaging mode—excitation mode method. Current excitation sequence is synchronized with scanner imaging sequence.

echo sequence (GRE) sequence in a 1.5 T clinical MRI scanner (Espree, Siemens). The catheter was immersed in a water phantom containing 1% Gd-DTPA that was placed at the magnet isocenter with the catheter in the vertical direction, perpendicular to B_0 . This position was chosen in part because it is one of the most difficult orientations, since the catheter has to overcome gravity to deflect. A clinical knee coil was used for signal reception.

II.F. Catheter bright signal- modified GRE sequence

Catheter visualization was achieved by taking advantage of the field distortion artifact present when the dc current excites the catheter coils to produce tip deflection. Since the tracking signal originates from signal voids in this case, the method can be considered a passive tracking method.^{13–15,25} However, dark signals can be confused with many other effects during actual application *in vivo*. Therefore, a method that can produce specific hyperintense signals from the image would clearly be beneficial and potentially used together with an automated algorithm to track the tip of the catheter. To this end, similar to the work done for other catheter tracking experiments,²⁵ we modified a GRE sequence by changing the amplitude of the slice-select rewinding gradients. This has the effect of spoiling signal in homogeneous regions of the magnet while those regions near the magnetic field gradient at the catheter tip are refocused and appear “bright” in the image. As shown in the diagram of the modified GRE sequence [Fig. 5(a)], an easy choice for implementing this type of sequence is to completely remove the slice-select refocusing gradient by setting its amplitude to zero. Then switching between this and the standard sequences was possible by simply enabling/disabling the rewinding gradient in the user interface window.

II.G. Imaging mode-excitation mode

During a complete procedure the interventionist will switch back and forth tracking images of the catheter to images of the surrounding structures. This requires an imag-

ing method that allows clear imaging of the background even when the catheter is deflecting with driving currents on its steering coils. As described above, this current would normally result in a relatively large signal void around the catheter tip through field distortion. In order to avoid this situation, both an imaging and an excitation mode were implemented as shown in Fig. 5(b). In the imaging mode, the coils were turned off from the time that the RF pulse was applied until the end of the readout period. In the excitation mode, the desired coil was turned on through the rest of the repetition cycle.

Because this required more rapid switching of currents than our relatively slow serial MATLAB connection, a different device was required for this study. To this end, different excitation sequence data were generated in MATLAB matrix format and loaded into a data-timing generator (Tektronix DTG7058 750 Mb/s) synchronized with the RF triggering signal from the scanner. This device can more rapidly generate the PWM signals that control the current bridges, and can turn the signals off during the readout period. It is important to note that the applied method must not affect the final catheter position for a particular excitation condition and relative position to B_0 . For this reason, the imaging mode was made as short as possible and the excitation mode comparably much longer. Ideally, if one knew the time constant τ of the motion of the catheter-coil system one could set the time of the imaging mode much smaller than τ and the time of the excitation mode much larger than τ . For our purposes, a GRE sequence was used for imaging with 25° flip angle, 200 mm FOV, and 10 mm slice thickness. With these parameters fixed we adjusted the repetition time TR, readout time TS, and echo time TE in order to avoid catheter motion while still allowing a full deflection. As shown in Fig. 5(b), the current(s) through the coil(s) were switched off from the time the RF pulse is triggered until the end of the readout period. The total off time T_{OFF} is given by the following equation:

$$T_{OFF} = T_E + 0.5(T_{RF} + T_S), \quad (7)$$

FIG. 6. Increasing deflection of the catheter: (a) no coil excited, (b) first axial coil excited, (c) both axial coils excited, and (d) three coils are excited. The white arrow indicates the direction of the magnetic field (B_0). $B_0 = 1.5$ T. Current per coil $I = 110$ mA. The image was acquired using a real-time GRE sequence, 25° flip angle, 200 mm FOV, 10 mm slice thickness, and 7.8 ms TR. Artifact indicated by white arrow in Fig 6 (b) is due to interference of the current control system with the scanner.

where

$$T_S = (BW/\text{pixel})^{-1}, \quad (8)$$

T_{RF} (2 ms) is the length of the RF pulse, and BW/pixel is the bandwidth per pixel, which is the parameter adjusted in the scanner through the user interface.

III. RESULTS

III.A. Catheter tip deflection

Large deflections were possible by exciting the three coils sequentially (Fig. 6). Deflections as large as 2.5 cm from vertical at the tip were observed without heating the catheter considerably (Figs. 7 and 8). Also seen in Figs. 6(b)–6(d) are the significant signal voids in the image due to field dispersion effects from the field generated by the coils. The deflection of the catheter tip when only the distal axial coil was excited with different dc current values was also measured (Fig. 7). The current was sensed by a 0.1Ω 1% shunt resistor, and the deflection angle was measured indirectly by measuring its tangent over images using IMAGEJ free software

(NIH). The error bars in Fig. 7 were calculated through propagation of errors that results from the indirect measurement methods adopted to calculate current and deflection angle. The deflection was 2.5 cm in the case of exciting the set of coils sequentially with a 100 mA maximum current per coil, while in the case of exciting only the axial coil at the catheter tip, the deflection was reduced by half at 30% higher current. As can be seen from Fig. 7, the deflection vs current curve is nonlinear and can be approximated with a second order polynomial function. As the catheter deflects, the torque decreases with the sine of the angle between the magnetic moment generated by the current in a coil and the magnetic field. Therefore, at larger deflection angles, for the same initial position relative to the main field, the catheter will reach steady state at lower bending moment. This is in agreement with large deflection theory that establishes a nonlinear relation between deflections and bending moment. In Figs. 6(b)–6(d) there is a visible artifact (indicated by white short arrow) due to interference between the scanner and the current control system, which was located inside the room for this particular experiment. The temperature measurement of

FIG. 7. Total deflection of the catheter in the magnet while exciting one axial coil with increasing values of currents (solid diamond), and exciting sequentially the steering coil array with a 100 mA current per coil (open triangle).

FIG. 8. Temperature curve of steering catheter tip measured on the bench-top. Temperature was sensed on a single axial coil (70 turns of 42 AWG wire) driven by 100 mA.

FIG. 9. (a) Imaging catheter with GRE 5.5 ms TE, 400 ms TR, 300 Hz BW/pixel while switching off the current through the coils during 8.2 ms T_{OFF} . (b) Imaging catheter with GRE 1.8 ms TE, 400 ms TR, 980 Hz BW/pixel while switching off the current through the coils during 3.3 ms T_{OFF} .

the catheter tip, performed on the benchtop, demonstrated an initial temperature slope of $(0.48 \pm 0.2) ^\circ\text{C/s}$ and a temperature change from (24.3 ± 0.2) to (31.1 ± 0.2) in 110 s (Fig. 8).

III.B. Catheter imaging

Several iterations were carried out in order to find the correct timing parameter (e.g., TR, TE, T_{OFF} , etc.) values that led to satisfactory image quality. Figure 9(a) shows the result of imaging with a TR of 400 ms, a BW/pixel of 300 Hz, and a TE of 5.5 ms; applying these parameter values to Eq. (7) results in $T_{\text{OFF}} = 8.2$ ms. As indicated by the black arrow, the image presents a motion artifact due to oscillation of the catheter while switching on and off the current through the coils. Since the repetition time is already very long—approximately 70 times larger than the T_E —we reduced the off time by increasing BW/pixel and making TE three times smaller ($TE = 1.8$ ms) and increasing BW/pixel to 980 Hz. As can be seen in Fig. 9 (b), the oscillation disappeared for these parameter values. However, the selected TR is too long for real-time imaging, so it was further reduced and additional analyses explored how the reduction affected the

FIG. 10. Deflection angle that result from changing TR when exciting only one axial coil with 100 mA.

catheter motion. Figure 10 shows how TR affects the total deflection, referred to the plane perpendicular to B_0 . As shown in the figure, a TR as small as 10 ms can be used without significantly affecting the deflection.

Figure 11 shows the result of applying the different imaging methods. No coil was excited in Fig. 11(a), while in Figs. 11(b)–10(d) the distal axial coil was excited. The figure shows the images resulting from the conventional GRE sequence, the GRE sequence with the excitation-imaging mode and the modified GRE sequence with rewinding gradients suppressed. In Fig. 11(c), there is apparent deflection of the catheter free of field distortion and motion artifact allowing visualization of surrounding areas. We observed that the motion of the catheter during the imaging experiment was minimal as long as the readout window was shorter than 2 ms. Also, the deflection was suboptimal for TR less than 15 ms. On the other hand, as long as one is not interested in imaging the tissues surrounding the catheter, the field dispersion inhomogeneity can be advantageously used for catheter visualization, as shown in Fig. 11(d). In this case, a localized signal from areas surrounding the catheter tip is received, while the signal from other regions is suppressed.

FIG. 11. Catheter images using different imaging approaches: (a) no coil excited using a GRE sequence. After coil excitation (b) GRE sequence (c) GRE sequence with excitation and imaging mode (d) GRE sequence with unbalanced slice-select gradient. Artifact indicated by white arrow in Fig 10 (d) is due to interference of the current control system with the scanner.

IV. DISCUSSION

This work has shown that an array of steering coils is a potential solution for remote control of a catheter in an IMRI setting. The proposed method for controlling the array is flexible enough to perform complex and large deflections. In order to have independent force vectors generated by current circulating through the coils, we designed a set of micro-processor controlled current sources, which receive data from a PC running MATLAB. Even though the coil excitation parameters are currently set manually in MATLAB, in the future, a control algorithm running in a PC using the same platform should be able to output a coil excitation sequence based on the error signal between a target and current catheter tip position using the combination of this hardware and the catheter visualization approaches described here.

The direction of the catheter deflection is determined by the direction of the net magnetization vector relative to the catheter axis and its relative orientation with the main magnetic field. For the deflection measurements presented here, we placed the catheter near the magnet isocenter in a vertical orientation so that it had to overcome gravity to deflect. We then measured deflections in the z - y plane (scanner reference axis). To have larger deflections in this selected initial position one side coil was aligned in the direction of the static magnetic field of the scanner. Under these conditions, a second side coil would not add deflection if there was no rotation of the catheter tip and therefore it was omitted here; however, this second side coil would be necessary to have full control of the catheter tip in space and add deflection for other catheter orientations.

As mentioned above, an ideal system would incorporate a full control system to drive the catheter to its final position. However, full control requires implementation of feedback as well as a model of the system for predicting catheter location and orientation for a given coil excitation sequence. Based on the principle of stationary potential energy, Tunay formulated a model of a catheter with a permanent magnet in its tip immersed in a variable magnetic system.¹⁷ The complexity of such a system is reflected in the elaborate mathematical equations describing the catheter motion. The complexity of the system presented here is expected to be the same (or even higher as the number of coils increases). Here, the external main magnetic field has a unique direction and amplitude, while the magnetization generated by controlling the current in the coil in the catheter tip is variable in both magnitude and direction. Alternatively, we could use a very low level model of the physical system and adopt a non-classical modeling method (e.g., fuzzy logic) to model the system (black-box) from input–output data. However, a critical step in implementing these methods is obtaining a sufficiently accurate set of input–output data, or training data, in a region of interest and over a wide range of operating conditions. Thus, accurate current and deflection measurement methods are essential requisites for a successful model of the steerable catheter system presented here. The input to such a closed loop system model would be the target catheter location as set by the practitioner through a manual

steering device. This input would be compared with the current catheter position that would be available in the scanner interface through application of a tracking method. Then, this error signal would be processed by the control system, which would generate the necessary PWM data (coil selection, corresponding duty cycle, and current polarity) to match the desired tip location and orientation. The communication between the main PC and the external system was already implemented in this work, and future work will involve the design of the scanner-PC interface and implementation of the manual steering device.

As mentioned previously, the practitioner operating the system needs to have real-time information of the exact location of the interventional device relative to the tissue. Here a passive catheter tracking method using a GRE sequence without rewinding gradients that does not require modification of the system used for deflection has been implemented [Fig. 5(a)]. By applying this method, a bright spot can be generated at the catheter tip in addition to some signal along the catheter, which originates from the local spin dephasing introduced by the current circulating in the copper wires [Fig. 11(d)]. It is noteworthy that the size of the bright spot can be controlled by adjusting the current value during the acquisition window. An alternating method between excitation and imaging modes to obtain a clear image of the background with minimal interference from the catheter during deflection has also been successfully created [Fig. 11(c)].

In terms of the catheter prototype itself, limitations have been encountered in building the catheter due to the small size of the coils, the small diameter of the wire and the complexity of the array. It should be clear that the prototype used for the experiments shown here is not considered optimal. It is noteworthy that the coil array limits the torqueability of the catheter tip. This could introduce the potentially significant problem that incorporation of the coil array necessitates increased higher torque for a given deflection. As shown in Eq. (1) higher magnetic torque, for a fixed dc current and static magnetic field values, is possible by increasing number of turns and area of the coils; however, the catheter moment of inertia increased as well. Therefore, the determination of optimum parameter values requires analysis of the catheter dynamics, which will be the subject of future work. However, it is clear that microfabrication techniques could play a role in these developments going forward.

Finally, as with every potential medical device, there are safety concerns that arise from the system. We have focused only on an initial implementation of this concept here. However, it is clear that both dc and RF safety would have to be assured before any future comparable device was practically useful in the clinic. In our opinion, the primary risk in both of these areas is heating. In terms of heating due to dc currents, it was shown that if we excite one of the coils at the maximum current for about a minute on the benchtop, there is a change in temperature of approximately 7 °C. We also measured a maximum temperature slope of about 0.5 °C/s. It should be noted that all these temperature measurements were performed with the catheter in still air, which presents a worst-case scenario for heat dissipation when compared to

a real procedure where there is potential heat convection due to blood flow. In addition, to facilitate the task of the interventionist in steering the catheter through tortuous vessels, we would only need to excite the coils with a pulse-like current excitation to produce deflection at the point of curvature. Therefore, we expect that steering coils be driven by high current only for a short time at duty cycles less than 100%. In this case, heat dissipation will be lower than that measured in our benchtop measurements. Note that in the presented prototype, we have not added any additional dc safety mechanisms other than the insulation on the magnet wire. In the case of a rupture of the wires, this could expose the patient directly to the dc current. It is clear that this could be addressed at the time of clinical implementation by adding additional insulation to the wires by placing them inside the catheter wall as well as feedback mechanisms in the driving electronics to detect any circuit disruptions comparable to a ground fault current system. There is also a concern about RF heating due to induced currents in the catheter while the scanner is transmitting RF power. This is a problem common to many earlier MR-based catheter designs^{22–24,26} and in our study, we only evaluated the heating effects that arise exclusively from the dc currents on the steering device itself. However, we anticipate that some of the other approaches to this problem, such as baluns and shield traps, should be able to address these issues in any clinical implementation.

V. CONCLUSION

In this work, new strategies for steerable catheter design and imaging that may ultimately contribute to development of new interventional devices for MR guided intravascular procedures have been presented. In particular, it was demonstrated that one can obtain high deflections using an array of steering elements at the catheter tip. In addition, the work demonstrated two types of image acquisitions: one for tracking the steerable catheter and the other one for imaging the surrounding tissue near the catheter without interference

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¹J. J. Connors and J. C. Wojak, *Interventional Neuroradiology*, 1998.

²M. S. Grady *et al.*, "Preliminary experimental investigation of in vivo magnetic manipulation: results and potential application in hyperthermia," *Med. Phys.* **16**(2), 263–272 (1989).

³J. C. Chu *et al.*, "Performance of magnetic field-guided navigation system for interventional neurosurgical and cardiac procedures," *J. Appl. Clin. Med. Phys.* **6**(3), 143–149 (2005).

⁴D. R. Elgort and J. L. Duerk, "A review of technical advances in interventional magnetic resonance imaging," *Acad. Radiol.* **12**(9), 1089–1099 (2005).

⁵T. P. Roberts *et al.*, "Remote control of catheter tip deflection: an opportunity for interventional MRI," *Magn. Reson. Med.* **48**(6), 1091–1095 (2002).

⁶J. B. Mathieu and S. Martel, "Magnetic steering of iron oxide microparticles using propulsion gradient coils in MRI," *Conf. Proc. IEEE Eng. Med. Biol. Soc.* **1**, 472–475 (2006).

⁷S. Martel *et al.*, "Medical and technical protocol for automatic navigation of a wireless device in the carotid artery of a living swine using a standard clinical MRI system," *Med. Image Comput. Comput. Assist. Interv. Int. Conf. Med. Image Comput. Comput. Assist. Interv.* **10**(Pt 1), 144–152 (2007).

⁸V. Lalande, F. P. Gosselin, and S. Martel, "Catheter steering Using a Magnetic Resonance Imaging System," *Proceedings of the EBMC 2010* (Buenos Aires, Argentina, 2010) pp. 1874–1877.

⁹M. Bock *et al.*, "MR-guided intravascular procedures: Real-time parameter control and automated slice positioning with active tracking coils," *J. Magn. Reson. Imaging* **19**(5), 580–589 (2004).

¹⁰J. L. Duerk *et al.*, "Remember true FISP? A high SNR, near 1-second imaging method for T2-like contrast in interventional MRI at .2 T," *J. Magn. Reson. Imaging* **8**(1), 203–208 (1998).

¹¹M. E. Ladd *et al.*, "Active MR visualization of a vascular guidewire in vivo," *J. Magn. Reson. Imaging* **8**(1), 220–225 (1998).

¹²H. H. Quick *et al.*, "Interventional magnetic resonance angiography with no strings attached: wireless active catheter visualization," *Magn. Reson. Med.* **53**(2), 446–455 (2005).

¹³C. J. Bakker *et al.*, "Visualization of dedicated catheters using fast scanning techniques with potential for MR-guided vascular interventions," *Magn. Reson. Med.* **36**(6), 816–820 (1996).

¹⁴A. Glowinski *et al.*, "Device visualization for interventional MRI using local magnetic fields: basic theory and its application to catheter visualization," *IEEE Trans. Med. Imaging* **17**(5), 786–793 (1998).

¹⁵M. K. Konings *et al.*, "Development of an MR-safe tracking catheter with a laser-driven tip coil," *J. Magn. Reson. Imaging* **13**(1), 131–135 (2001).

¹⁶F. Settecase, "Magnetically-assisted remote control (MARC) steering of endovascular catheters for interventional MRI: A model for deflection and design implications," *Med. Phys.* **34**(8), 3135–3142 (2007).

¹⁷I. Tunay, "Modeling Magnetic Catheters in External Fields," *Proceedings of the 26th Annual International Conference of the IEEE EMBS* (San Francisco, California, 2004).

¹⁸I. Tunay, "Position control of catheters using magnetic fields," in *Proceedings of the IEEE International Conference on Mechatronics*, 2004.

¹⁹T. Belendez *et al.*, "Numerical and experimental analysis of large deflection cantilever beam under a combined load," *Physica Scr.* **T118**, 61–65 (2005).

²⁰K. E. Bisshopp and D. C. Drucker, *Quart. Appl. Math.* **3**, 272 (1945).

²¹T. K. Yu and L. C. Zhang, *Plastic Bending: Theory and Applications* (1996).

²²M. E. Ladd and H. H. Quick, "Reduction of resonant RF heating in intravascular catheters using coaxial chokes," *Magn. Reson. Med.* **43**(4), 615–619 (2000).

²³M. K. Konings *et al.*, "Heating around intravascular guidewires by resonating RF waves," *J. Magn. Reson. Imaging* **12**(1), 79–85 (2000).

²⁴C. J. Yeung, R. C. Susil, and E. Atalar, "RF heating due to conductive wires during MRI depends on the phase distribution of the transmit field," *Magn. Reson. Med.* **48**(6), 1096–1098 (2002).

²⁵J. N. Draper, M. L. Lauzon, and R. Frayne, "Passive catheter visualization in magnetic resonance-guided endovascular therapy using multicycle projection dephasers," *J. Magn. Reson. Imaging* **24**(1), 160–167 (2006).

²⁶A. J. Martin *et al.*, "MR imaging during endovascular procedures: an evaluation of the potential for catheter heating," *Magn. Reson. Med.* **61** (1), 45–53 (2009).