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MRI Compatibility of Microfabricated Magnetic Actuators for Implantable Catheters: Mechanical Evaluations

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Abstract

Here we demonstrate the mechanical robustness of microfabricated torsional magnetic actuators in withstanding the magnetic fields produced by a 7 T MRI magnet. The static and dynamic mechanical characteristics of 30 devices were quantitatively measured before and after exposure to both uniform and non-uniform magnetic fields. The results show no statistically significant change in both the static and dynamic mechanical performance, which mitigates concerns about the mechanical compatibility of our devices with MRI scanners. Additional experiments are required to quantify the potential image-artifact size and radio-frequency-induced heating caused by the magnetic microactuators inside an MRI scanner.

I. Introduction and Motivation

The development of implantable medical devices has benefited tremendously from recent advances in micro and nano-fabrication technology. There now exists a variety of chronically implantable microscale sensors and actuators to monitor critical biological information and to proactively manage patient care. At the same time, the use of advanced medical imaging, such as the magnetic resonance imaging (MRI), in medical diagnostics has also increased. It is, thus, critical to ensure that next-generation microscale medical implants are compatible with often needed medical diagnostic procedures to ensure patient safety.

We have previously reported on the feasibility of using microfabricated magnetic actuators to improve the functionality of medical catheters used in hydrocephalus management [1]–[4]. Hydrocephalus is a serious neurological disorder that is often characterized by an

abnormal accumulation of cerebrospinal fluid (CSF) in the ventricles of the central nervous system. Patients with hydrocephalus are typically treated with chronically implanted shunt systems to divert excess CSF from the brain to the abdomen. Unfortunately, these shunt systems are plagued with a high rate of failure the first year of implantation (up to 40%), which is mostly due to cellular obstructions in its ventricular end [5].

Using our magnetic microactuators, we have demonstrated cell-clearing capabilities to combat cellular obstruction at the catheter pores [2], [3]. Our ultimate aim is to improve the existing hydrocephalus treatment option by increasing the lifetime of the ventricular catheter – the site of cellular obstruction – using the integrated magnetic microactuators.

The cell-clearing capability notwithstanding, the magnetic microactuators must be compatible with MRI scanners since hydrocephalus shunt systems are chronically implanted devices. In case these patients need MRI diagnosis in the future, it is imperative that the integrated magnetic microactuators not break due to strong magnetic torques and forces thereby causing damage to patients.

Assessing the MRI compatibility of a medical implant consists of several components. In fact, there already exist standards for determining MRI compatibility of conventional medical devices. The ASTM (American Society for Testing and Materials) International standards recommend testing of four different metrics to determine the magnetic resonance compatibility of medical devices. These include: the magnetic field induced torque [6], the magnetic field gradient induced displacement [7], radio-frequency (RF) induced heating, and image artifact. Although a comprehensive assessment of the MRI compatibility of a device requires testing of all four aforementioned metrics, in this report we assessed the potential impact of only the first two (i.e., MRI induced torque and displacement force) on mechanical integrity (e.g., material fracture or yield).

Since the ASTM standards are written for testing macro-scale medical devices, the prescribed testing methods from the standards are not directly applicable for our microactuators. To date, there has been no published report on evaluating the MRI compatibility of microscale devices. There have been, however, reports of using mechanical properties (e.g., resonance) of microdevices to determine changes in structural integrity [8]. Here we seek to demonstrate the mechanical robustness of our torsional magnetic microactuators by comparing mechanical properties of the device before and after repeated exposure to the strong magnetic forces inside an MRI scanner.

II. Device Design and Theory

A. Design and Operations of Torsional Magnetic Microactuators

This report examined the MRI compatibility of circular torsional magnetic microactuators. The microactuator consists of a silicon nitride structural plate anchored on its sides by two torsion beams (Fig. 1a). A ferromagnetic element is electroplated on top of the circular structural plate in order to apply magnetostatic torque to the device.

The torsional magnetic actuators used for this report were designed to have the same beam characteristics: 198 μm in length, 20 μm in width, and 1 μm in thickness. The length and thickness of the magnetic element were constant across all devices (400 μm and 7 μm , respectively), however, the magnet width varied (Table I).

In the presence of a magnetic field, the torque generated by the ferromagnetic element causes the structural plate to rotate about the long axis of the torsion beams (Fig. 1b). The degree to which the structural plate rotates is described by the following equation

$$\theta = \frac{\nu \cdot (\vec{M} \times \vec{H})}{k_\phi} = \frac{\nu \cdot M \cdot H}{k_\phi} \cdot \sin(\gamma - \phi), \quad (1)$$

with angular rotation θ , magnet volume ν , magnetization vector $\vec{M}$ and scalar magnitude M , magnetic field vector $\vec{H}$ and scalar magnitude H , and the angular torsion-beam stiffness k_ϕ [9], [10].

Dynamic response of the torsional actuators is also well described in literature [11], [12]. The primary resonant frequency mode can be described

$$\omega = \sqrt{\frac{k_\phi}{M_{\text{eff}}}}, \quad (2)$$

with the effective mass of the structural plate M_{eff} .

The MRI scanner generates both dc and ac magnetic fields. The dynamic behavior of our torsional magnetic actuators, however, depends on the driving frequency of the MRI scanner. Typically, MRI scanners operate at the megahertz range while the resonant frequency of our devices is approximately 1 kHz. At such a high driving frequency, the magnitude of actuator deflection is greatly attenuated to the point of being negligible.

In a dc magnetic field, the uniformity of the magnetic field plays an important role in inducing movement. Two different movement types can occur in a static magnetic field: rotation and displacement. The rotation occurs when magnetostatic torque is generated in a uniform magnetic field and the displacement occurs when translational force is induced by the non-uniform magnetic field.

Moreover, the orientation of the device in relation to the magnetic field is also critical in device movement. That is, the torque and the displacement could occur along each axis of the device depending on the orientation by which the device is entering the magnetic field (Fig. 2). Maximum movement and the corresponding stress, however, will typically occur along the beam dimension with least stiffness (i.e., along x and z axis for torsion and translation, respectively).

B. Torque-Induced Shear Stress

As can be seen in Eq. 1, the maximum torque that can be generated by the magnetic element occurs when the magnetization direction γ and the applied magnetic field direction ϕ are perpendicular ($\gamma - \phi = 90^\circ$). Since the torsional beam can rotate in either direction, the deflection does not exceed 90° (Fig. 3a).

The resulting shear stress on the torsion beams due to induced torque can be estimated using well-established formulas [13]. Rotating the torsion beam to the maximum deflection angle of 90° about their x axis will induce 801 MPa of shear stress (Figure 2, 3b). The literature indicates that the fracture-strength of a low-stress silicon-nitride layer deposited by low-pressure chemical-vapor deposition (LPCVD) is approximately 6 GPa [14], [15], which is far greater than the amount of shear stress that the beams would experience in an MRI scanner. Thus, we expect to see no effect on the torsion-beam mechanics of devices inserted into the strong magnetic field of a MRI scanner.

C. Displacement-Induced Tensile Stress

The magnetic field near an MRI scanner is typically nonuniform. A gradient in the magnetic field will induce a translational force F_{trans} on a ferromagnetic element, which can be expressed by

$$F_{\text{trans}} = v \cdot M \cdot \frac{dH}{dx}, \quad (3)$$

with magnetic field gradient dH/dx .

Since each MRI scanner has a unique magnetic field profile, it is difficult to estimate the non-uniformity of the stray magnetic field. For this report, we used a site-planning document provided for a high-field-strength (7 T) MRI magnet (70/30USR, Bruker Biospin, Billerica, MA) to estimate the maximum magnetic flux density gradient (17 T/m). As seen in Eq. 3, the translational force F_{trans} is linearly proportional with the magnet volume v . Thus, the largest force that the device will experience is approximately 14 μN .

The maximum tensile stress due to the translation along the z axis (out-of-plane) is approximately 208 MPa [13], which again is far less than the fracture strength of the LPCVD silicon nitride.

III. Experimental Methods

The microfabrication steps as well as the static and dynamic mechanical characterization procedures for our torsional magnetic microactuators are described in [2], [3]. Using a laser-based deflection setup, the static deflections and the dynamic responses of the torsional magnetic microactuators were measured before and after exposing the devices to a 7 T animal MRI scanner (70/30USR, Bruker Biospin, Billerica, MA).

The MRI exposure procedure was designed such that maximum torque and translation force are imparted on the magnetic microactuators. The location of the greatest uniform magnetic field is found within the MRI magnet bore, whereas the highest magnetic field gradient is found at the entrance of the MRI bore. Prior to exposure to the MRI scanner, the silicon chip containing the array of devices was glued inside a cubic plastic container (16.4 cm^3) to ensure that no magnet fragment could get lost inside the MRI scanner.

To test the effect of magnetic torque on the device, the plastic container was placed inside the MRI bore for 20 min along each of the three axes of the device. Following exposure, the devices were removed from the MRI scanner and were visually inspected using a small permanent magnet.

For the translational force, the plastic container with the device array was placed on the loading arm of the MRI scanner and guided into the MRI bore. The movement into and out of the bore was repeated five times for each of the three axes of the devices. Following exposure, the devices were visually inspected prior to post-exposure testing.

IV. Results

We tested a total of 30 devices with varying magnet volumes (Table I). Visual inspection following each exposure revealed no broken beams for any of the tested devices.

A. Static Response

We tested the static response of the devices with an increasing magnetic field from 0 to 7 kA/m, which resulted in a maximum deflection ranging from 25° to 30° . Within this magnetic-field-strength range, the angular deflection increased linearly with the applied field (Figure 3(a)). We compared the pre- and post-exposure value for the angular deflection of each device (Figure 4). A Kruskal-Wallis test showed no statistically significant change in static response due to exposure to the MRI magnetic field ($p = 0.395$).

B. Dynamic Response

Figure 5 shows the resonance characteristic of each device before and after MRI exposure. As expected, a before-and-after comparison of the first resonance mode also showed no statistically significant change due to MRI exposure ($p = 0.838$). These results indicate that, as expected, our torsional magnetic microactuators are capable of withstanding the magnetostatic torque and translational force induced by a very strong 7 T MRI scanner.

Some shunt systems prohibit patients from getting images using a stronger MRI scanner ($>3 \text{ T}$) due to concerns of potential complications with the shunt valves. Fortunately, typical MRI scanner used for human patients has a magnetic flux density of 1.5 T to 3 T. Compared to the 7 T scanner we used to test our devices, the relatively weaker magnetic field strength of the clinically relevant MRI scanners should induce much smaller magnetostatic torques and forces and thus have an even smaller impact on the mechanical integrity of our devices.

Although these results are encouraging for our efforts toward the ultimate goal of creating MRI-compatible MEMS-integrated ventricular catheters (Fig. 1), additional experiments are

required to fully assess the compatibility of our torsional magnetic microactuators. Our future plans include a quantitative assessment of the imaging artifacts and the RF-induced heating caused by our ferromagnetic devices.

V. Conclusions

Here we have demonstrated that our torsional magnetic microactuators are capable of withstanding the impact of the magnetic field produced by a 7 T MRI magnet. The magnetostatic torque and force are linearly proportional to the magnet volume (Eqs. 1 and 3) while the mechanical stiffness is dependent on the beam geometry [13]. Our results suggest that by using these parameters, we can effectively engineer a given level of mechanical MRI compatibility into our magnetic microactuators. Additional work remains to resolve the potential issues of image artifacts and RF-induced heating caused by our microdevices. Nevertheless, these results provide a promising outlook for the realization of mechanically robust MEMS-integrated ventricular catheters that may be used safely in a clinical setting without mechanical failure due to the use of an MRI scanners.

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Fig. 1.

A set of 3D schematic diagrams of the torsional magnetic microactuators: (a) unactuated released state, (b) actuated device with an applied magnetic field, (c) integrated into a ventricular catheter.

Fig. 2. COMSOL simulation of torsion and translation along each axis. Note that the maximum stress occurs for movement along the x axis for torsion and the z axis for translation.

(a) Torsional Beam Deflection vs. Magnetic Field

(b) Shear Stress vs. Angular Rotation

Fig. 3.

The static response of torsional magnetic microactuators and corresponding beam shear stress. (a) Maximum angular deflection in an increasing magnetic field. Note that no device rotates more than 90° and that the response is linear when magnetic field is <10 kA/m. (b) Amount of shear stress as a function of angular rotation. Note that at 90° , the shear stress is much less than the fracture strength.

Fig. 4.

Comparison of the angular deflection slope for each device pre-and post-MRI exposure. The slope of each device was measured using an increasing magnetic field ranging from 0 to 7 kA/m.

Fig. 5.
Comparison of the first-mode resonance for each device pre- and post-MRI exposure.

TABLE I

Number of devices tested and magnet width

Device ID	13-2	13-3	13-4	13-5	14-2	14-3	14-4	14-5
<i>n</i>	4	6	5	4	1	2	3	5
Width (μm)	364	304	220	112	376	336	280	208