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DEMONSTRATION AND EXPERIMENTAL VALIDATION OF PLASTIC-ENCASED RESONANT ULTRASONIC PIEZOELECTRIC ACTUATOR FOR MRI-GUIDED SURGICAL ROBOTS

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Abstract

Intra-operative medical imaging based on magnetic resonance imaging (MRI) coupled with robotic manipulation of surgical instruments enables precise feedback-driven procedures. Electrically powered non-ferromagnetic motors based on piezoelectric elements have shown to be well suited for MRI robots. However, even avoiding ferrous materials, the high metal content on commercially available motors still cause distortions to the magnetic fields. We construct semi-custom piezoelectric actuators wherein the quantity of conductive material is minimized and demonstrate that the distortion issues can be partly addressed through substituting several of these components for plastic equivalents, while maintaining motor functionality. Distortion was measured by assessing the RMS change in position of 49 centroid points in a 12.5mm square grid of a gelatin-filled phantom. The metal motor caused a distortion of up to 4.91mm versus 0.55mm for the plastic motor. An additional SNR drop between motor off and motor spinning of approximately 20% was not statistically different for metal versus plastic ($p=0.36$).

INTRODUCTION

Brachytherapy [1], a localized radiation based cancer treatment, biopsies, and tumor ablation [2] are examples of routinely performed medical procedures that require accurate positioning of a needle inside the patient. In all cases, improved 3D location of the needle and structures of interest during the procedure can lead to better results. Today, several imaging modalities are used for guidance including ultrasound, computed tomography and MRI. Ultrasound is the most common due to its availability and temporal resolution; however, at the expense of image quality. Computed tomography has limited soft tissue contrast but has good spatial and temporal resolution. Recently, MRI is becoming a stronger

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candidate for imaging due to its superior soft tissue contrast which yields higher sensitivity for detecting malignancies [3, 4]. The MRI is also able to perform multiparametric scans including flow diffusion, real time temperature mapping and oxygenation measurements. However, the confined space of the MRI bore, strong magnetic fields and susceptibility to electrical noise pose challenges, ones that can be addressed by the use of MRI compatible robots [5].

Several MRI compatible robots are available in literature and demonstrate unique ways to deal with the challenges posed by the environment in the MRI. In all cases, no ferromagnetic components are used due to projectile risks and an attempt is made to minimize or properly locate conductive components. In efforts to best achieve this, various groups have employed a variety of actuators to move these robots.

Pneumatically actuated MRI compatible robots have been demonstrated and the technology has been developed greatly since its first appearance. Stoianovici et al created the PneuStep motor, a pneumatic stepper motor that, having no electrical or conductive components, does not affect image SNR [6, 7]. Fischer et al developed a robot for intraprostatic needle placement that achieved 0.94mm RMS accuracy inside an MRI using custom made glass/graphite cylinders [8]. Comber et al presented a 5 DOF active cannula robot for lower brain interventions that builds upon Fischer et al's fail-safe rod lock to improve robot safety during procedure. Yakar et al also demonstrated the feasibility of pneumatically actuated robots in the MRI through a 10 patient human trial [9]. These motors can offer high torque and be constructed from all plastic parts to achieve MRI safe classification. However, they are limited by the slow reaction times of the valves and long air lines. This is partly addressed by Yang et al's controller design that includes a model of the transmission lines [10].

Hydraulic actuation addresses some of the latency limitation of pneumatic actuation. Kokes et al demonstrated the feasibility of using hydraulic cylinders in the MRI for breast cancer related intervention [11]. Lee et al make use of hydraulic rolling diaphragm motors in their multi-axis cardiac cauterization robot [12]. Similar to pneumatic motors, the long hydraulic lines affect the response time of these actuators and leakage of the hydraulic fluid during a procedure can compromise the sterile field.

Electrical actuation in the form of piezoelectric motors has also been demonstrated in MRI compatible robots. Masamune et al built one of the first such robots in 1995 [13]. Patel et al demonstrated the use of nonresonant piezoelectric motors in a three DOF needle steering robot [14]. Nycz et al characterized a stereotactic frame replacement robot with seven degrees of freedom powered by resonant piezoelectric motors [15]. Seifabadi et al's pneumatic powered positioning stage for prostate biopsy [16] was later converted to work with resonant piezoelectric motors [17]. Piezoelectric motors pose some challenges for MRI compatibility due to their primarily metallic construction. Although nonferromagnetic, the generation of eddy currents due to the MRI's gradient fields can lead to magnetic field distortions causing image artifacts. Motor vibrations due to interactions between the MRI's magnetic fields and those generated by the eddy currents can further degrade image quality

by causing image artifacts. However, piezoelectric motors have fast reaction times and robots that use them may have reduced setup time as compared to other modes of actuation.

In this paper, we demonstrate that image distortion related issues can be partly addressed by replacing metallic nonactive motor components from a resonant ultrasonic motor for nonmetallic equivalents. Furthermore, the replacement can be done with no detriment to the signal-to-noise-ratio (SNR) as compared to its metallic counterpart.

MOTOR CONSTRUCTION

Piezoelectric ultrasonic motors are designed around two components: A stator with a ceramic ring coupled to it and a rotor that is frictionally coupled to the stator surface by applied pressure. The piezoelectric element, usually composed of lead zirconium titanate (PZT), is patterned with conductive electrodes to which drive signals are connected. The excitation of the ceramic at a particular frequency by two properly shifted sinusoidal waves induces a vibrational mode with a traveling wave that rotates around the shaft. This traveling wave pushes the rotor generating torque [18].

A non-ferromagnetic USR-60 (Shinsei, Japan), shown in figure 1(a) was modified to reduce its metal content. Only the aluminum rotor and the copper stator with attached ceramic were incorporated into the new motor. A new enclosure was machined out of ULTEM™1000, a drive shaft was machined from Delrin™ stock and nylon screws were used in place of brass. A shim or metal spring is typically used to maintain a preload between the rotor and the stator in metal motors, however our design uses a polyurethane disk to accomplish the same function. The final assembly can be observed in figure 1(b).

METHODOLOGY

The motors under study, a standard USR-60 and our semi-custom motor, were analyzed inside a Philips Achieva 3T scanner. The USR-60 and the semi-custom motor will hereto be referred to as the metal motor and plastic motor, respectively. A 3D printed phantom 90mm × 90mm × 60mm with a 45mm homogeneous region and a 15mm tall grid composed of 12.5mm × 12.5mm squares in the lower portion was used as the object being imaged similar to [15,19]. The phantom was filled with a low concentration gelatin and care was taken to purge all bubbles.

The phantom was elevated approximately 5cm above the MRI table using a foam block and two Philips Flex Sense M imaging coil were placed on either side of the phantom. A wooden support structure was positioned above the phantom so that the motor under test could be located on it. The setup can be observed in figure 2.

Motors were powered from the controller box used by the robot in [15]. The controller box is fully shielded and is placed inside the MRI room. It was shown in [15] that it does not affect SNR when powered on. A fully shielded cable connects the controller box to a breakout board inside the scanner. The same motor driving card was used for both motors to reduce experimental uncertainty and contains a DS6060 (Shinsei, Japan) turned on and off via a control signal modulated at a frequency of approximately 300Hz on the clockwise

direction line at 50% duty cycle. PWM is commonly used in electrically powered MRI robots to control motor speed and for this reason we decided to perform tests under PWM. However, PWM with a piezoelectric motor can generate mechanical vibrations which can cause a signal drop in the MRI.

A T2 spin double echo sequence was used. The repetition time was set to 2000ms, echo time 20/80ms with 30 256mm × 256mm × 5mm slices. Voxels were 1mm × 1mm × 5mm. Slice overlap was set to 2.5mm. The sequence had a bandwidth of 16.6 KHz and lasted approximately 17 minutes per run. This sequence is representative of those used during intra-operative imaging with MRI robots.

Two repetitions of the T2-weighted scan was acquired in each of 5 conditions:

1. Baseline with no motor, controller off
2. Plastic motor inside, controller on, motor powered off
3. Plastic motor inside, controller on, motor spinning
4. Metal motor inside, controller on, motor powered off
5. Metal motor inside, controller on, motor spinning

In addition to the aforementioned imaging conditions, 2 more conditions were collected with the plastic and metal motor each positioned alongside the phantom instead of on the support structure. These conditions were used exclusively for distortion calculation. Figure 3 shows the positions of the plastic and metal motor both on top and alongside the phantom.

SNR was calculated based on the image difference method presented in [20] as “method 1”. In this method, the second image is subtracted from the first for each image pair to create a difference image. A region of interest (ROI) containing most of the phantom is chosen and kept constant for all image sequences. The average signal intensity μ is calculated as the average of voxel intensities in the ROI. The noise is calculated as the standard deviation σ of the pixel intensities in the ROI of the difference image divided by $\sqrt{2}$. The SNR is defined as in equation 1. SNR results between different test conditions were compared using IBM SPSS software based paired T-test. Statistical analysis is based on 14 slices through the homogeneous region of the phantom for each test condition.

$$SNR = \frac{\mu}{\sigma} \quad (1)$$

The image distortion was calculated following the same procedure as in [15]. A slice in the center of the phantom's grid was chosen for each imaging sequence. The centroid of each square was computed using an OpenCV based script and then the difference between the location of each centroid between the image of interest and the baseline was computed through an iterative closest point algorithm in MATLAB. The results were converted to mm. The distortion represents the distance between the centroid locations in the baseline image and those in the comparison image.

RESULTS

The SNR for each motor was compared to baseline for each of two test conditions: Motor off and motor spinning. Figure 6 shows the results. There was no statistical difference between baseline and the plastic motor inside the MRI ($p = 0.714$) when powered off. The metal motor when inside the MRI and powered off causes a reduction in SNR of 23% ($p < 0.001$). The drop in SNR for the metal motor powered off may have occurred due to vibrations caused by interactions between the magnetic field generated by the eddy currents on the metal enclosure and the magnetic field of the MRI itself or due to distortion in the magnetic field near the imaging location causing a signal drop. Both motors when spinning caused an additional drop in SNR of approximately 20% as compared to their powered off state. The additional drop in signal was not statistically different for both motors ($p=0.36$). The SNR drop was not significant enough to be apparent upon visual inspection of the images as can be observed in figure 4. Slices closer to motor exhibit lower SNR causing the large variance.

The plastic motor outperformed the metal motor in the distortion experiments. Table 1 shows the results. No difference was observed between the motor powered off and motor spinning. The larger difference between the motors was observed in the side configuration which is along the MRI's primary magnetic field and in close proximity to the region of the phantom that contains the grid. In this configuration the metal motor had over 10 times the maximum distortion of the plastic motor.

The raw MRI scans are shown in figure 4 using the same windowing. The distortion caused by the metal motor in the side configuration can be seen in figure 4(g) and quantified in figure 5. The surface plot shows a peak distortion of 4.91mm in the top left corner which is stretched upward. The metal motor creates a darkened region with an 80% drop in signal in the first grid square which only returns to nominal approximately 45mm into the phantom. The plastic motor shows no visible difference, see figure 3(d), to baseline when in the same configuration.

CONCLUSION AND FUTURE WORK

Results have shown that the plastic motor causes less distortion while maintaining an equivalent or better SNR. The maximum observed distortion decreased 79% from 4.91mm to 0.55mm for the plastic motor versus the metal motor when positioned on the side of the phantom. The lower distortion allows for higher flexibility in motor placement since it can be located closer to the ROI. Results have shown that the SNR for the plastic motor was better than its metal counterpart, despite it being often implied that a metal enclosure is required for a piezoelectric motor to shield the MRI from electrical noise. These are promising results for the field of piezoelectric powered MRI compatible robots since these modifications can be done to commercially available motors or used as a guidance for new motor development.

The demonstration that removing the metal enclosure of a piezoelectric motor that is driven by clean signals that do not contain harmonics near the Larmor frequency for the scanner

decreases distortion without causing an SNR decrease paves the way for fully plastic motors. Replacing the stator and the rotor's friction disk for plastic equivalents will lead to even lower distortion since only the wires and copper plating on the ceramic crystal will remain as conductive parts.

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FIGURE 1.

(a) Construction of a standard non-magnetic Shinsei USR-60 motor and (b) of the plastic-encased piezoelectric motor. In the plastic motor, the aluminum rotor, copper stator, and piezoelectric ceramic are taken directly from a Shinsei USR-60. The aluminum housing, bronze drive-shaft, shims, and screws have been replaced with nonmetallic substitutes.

FIGURE 2.

Full experimental assembly in MRI scanner with bed moved to extremity.

FIGURE 3.

(a) Metal motor positioned on top of the phantom. (b) Metal motor positioned on the side of the phantom. (c) Plastic motor positioned on top of the phantom. (d) Plastic motor positioned on the side of the phantom. In all cases, the motor was approximately 10mm from the start of the gelatin.

FIGURE 4.

MRI slice through phantom grid for each test condition: (a) Baseline (b) Plastic off (c) Metal off (d) Plastic spinning (e) Metal spinning (f) Plastic off, on side (g) Metal off, on side.

FIGURE 5.

Distortion magnitude with the metal motor positioned on the top side of the phantom. This is the test condition shown in Figure 3(b) and calculated from the change between Figures 4(a) and 4(g). In all other test conditions the maximum distortion is less than 0.7mm.

FIGURE 6.

SNR results of the five different testing conditions. For each condition, SNR was calculated on 14 slices through the homogeneous region of the phantom. Results are normalized with respect to the baseline scan.

TABLE 1.

Maximum distortion for 49 centroids in phantom grid. Three test conditions for each of plastic and metal motor.

Test Condition	Plastic	Metal
On Top & Power Off	0.54 mm	0.62 mm
On Top & Power On	0.55 mm	0.68 mm
Side & Power Off	0.34 mm	4.91 mm