Motor design for an MR-compatible rotating anode x-ray tube

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Introduction: The goal of this work is to position an x-ray fluoroscopy system adjacent to a high field (1.5 T) closed bore magnet, while maintaining a short distance (100 cm) between the x-ray and MR imaging fields of view. The short distance between the fields of view enables switching between imaging modalities with a minimum disturbance to critically placed catheters and other devices allowing for maximum flexibility in interventional procedures requiring image guidance. However, the close proximity of the x-ray source to the MR bore creates complications related to x-ray tube operation that arise from the MR fringe field. Modern fluoroscopy x-ray tubes use a rotating anode because the x-ray generation process is inefficient, and more than 99% of the energy is dissipated as heat. By using a rotating anode, the area bombarded by electrons is spread out around the circumference of the anode which increases the instantaneous heat loading capability of the tube, thus allowing for increased x-ray flux (figure 1). Rotating anode tubes typically use induction motors to turn the anode, but induction motors do not operate efficiently in the presence of an MR fringe field because the fringe field acts as a brake decreasing the rotation speed [1]. To overcome this problem we have developed an alternate motor design in which we propose to use the MR fringe field to generate torque. In our design we have phase windings embedded directly on the rotor. Applying voltage to a single phase winding creates a magnetic moment that prefers to be parallel with the direction of the fringe field. Constant torque can be generated by switching winding voltages on and off in such a manner that there is always at least one magnetic moment attempting to align with the fringe field. The rotation speed of our design is dependent on the strength of the MR fringe field, current values in the phase windings, factors related to the geometries of the rotor/phase windings, and frictional losses.

Methods and Materials: To demonstrate feasibility of our design a proof of concept has been assembled using a custom built Lucite cylinder (24.5 cm height, 14 cm diameter) as the rotor, magnet wire (AWG 22) for the phase windings, and AISI 316 non-magnetic stainless steel bearings (KMS bearings, Anaheim, CA) to support the rotor and serve as slip rings for the phase windings. Delrin disks are interleaved in between adjacent bearings in order to insulate the bearings from each other as well as to provide support for internal electrical connections from the phase windings to the inner and outer bearing races (figure 2). The voltage signals for the phase windings are generated using a three phase power module, which receives its control signals from a microcontroller (Microchip Technology, Chandler, AZ). An optical encoder (US Digital, Vancouver, WA) is mounted to the top of the rotor and monitors the angular position of the motor, which is used as a feedback control mechanism for the winding voltages. The MR fringe field is simulated by placing the motor in between a Helmholtz coil pair (Stangenes Industries, Palo Alto, CA) that provides a static magnetic field perpendicular to the axis of rotation of the motor (figure 3). The rotation speed of the motor is calculated by monitoring the voltage signals from the optical encoder using a data acquisition card (National Instruments, Austin, TX).

Results and Conclusions: This initial prototype of our motor has a maximum rotation speed slightly above 450 RPM when driven by a 45 mT fringe field with approximately 1A maximum current in each phase winding. Figure 4 displays the RPM values as the motor accelerates over a two minute interval. To effectively replace current induction motors being used in x-ray tubes our design will need to operate in vacuum at speeds above 3000 RPM and accelerate to max speed in a shorter time frame. To achieve these criteria several alterations to our initial design will be necessary. The rotor will need to be made from a ceramic material with solid copper embedded in the ceramic to serve as phase windings. Smaller diameter bearings will be used to reduce frictional losses and overall mass of the motor will be minimized in order to reduce rotational moment of inertia. The optical encoder will be eliminated by using back EMF signals from the phase windings for feedback control. We will also benefit from the fact that the actual MR fringe field will be closer to 100 mT and the maximum current in each phase winding will be larger than 1A. These alterations along with further optimization with the assistance of a lumped parameter mathematical model to predict motor behavior will allow us to obtain faster acceleration, greater maximum rotational speed, and the ability to operate in a strong vacuum.

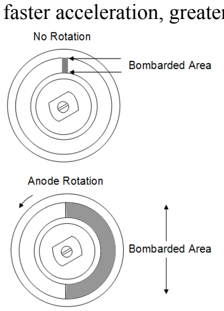
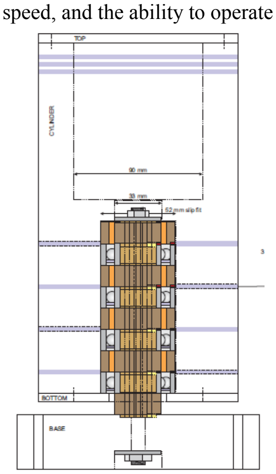
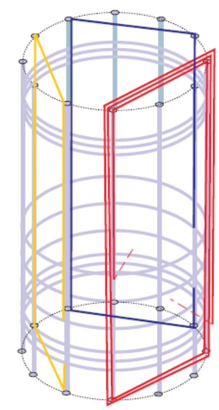


Figure 1: Diagram of a rotating anode demonstrating how rotation during x-ray exposure spreads the heat deposited on the anode.



(a)



(b)

Fig 2 a) Cross sectional view of the motor assembly illustrating the placement of the bearings and interleaved Delrin spacers in brown. b) Schematic of the phase winding layouts on the outside of the cylinder in red, blue and yellow. Each phase winding will have grooves in the cylinder which will hold the wiring in place.

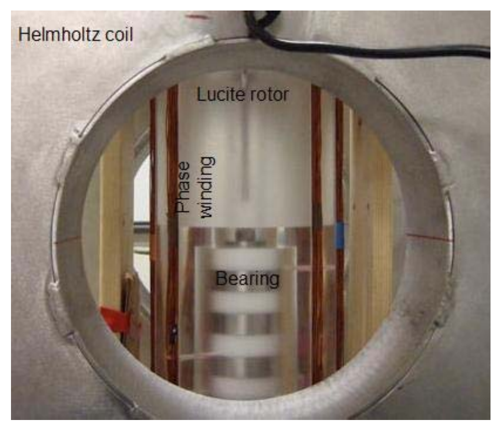


Figure 3: The motor prototype is shown positioned in between the Helmholtz coil pair. Not shown is the optical encoder which is positioned on top of the rotor.

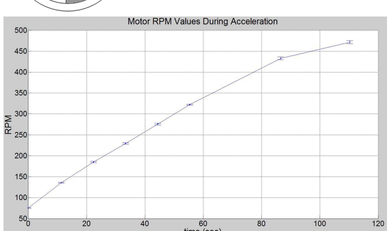


Figure 4: RPM values of the motor during acceleration to maximum rotational speed. RPM values were calculated using voltage signals from the optical encoder.

Acknowledgements Support was provided by NIH grant NIHR01 EB007626 and the Lucas Foundation

References [1] Brzozowski *et al*, "Compatibility of interventional x-ray and magnetic resonance imaging: feasibility of a closed bore XMR (CBXMR) system." Med Phys. 33(8), September 2006, 3033-3045.