Magnetic resonance imaging (MRI) has emerged as the premier imaging modality for the evaluation of a multitude of diseases. Although MRI was initially restricted to nontrauma and nonemergency applications, it has become the method of choice for the diagnosis of several trauma conditions, such as acute ischemic stroke, leading to increases in MRI-based pharmacological stress tests, with human trials currently determining the feasibility of MRI-based exercise stress tests. MRI can also differentiate between chronic and acute infarcts, so its use in triaging chest pain in the emergency room may grow. The field of MRI-guided surgical intervention is also continuously growing. MRI-guided excision and ablation of head, spine, abdominal, and pelvic tumors are performed at hundreds of sites worldwide. Intraoperative MRI guidance during cardiac electrophysiological mapping and thermal ablative procedures has been under development in animal models for the past 15 years. These electrophysiological procedures in animal have shown that MRI methods can detect the causes for arrhythmia recurrence postprocedure and can also determine the exact amount of tissue that has been ablated. As a result, human trials are on-going to test the feasibility of intraprocedural MRI ablation monitoring during the electrophysiological ablative procedures.

However, there are large patient populations who are excluded from MRI and MRI-guided intervention. Patients, who are hemodynamically unstable, have subacute or acute...
ischemia, have implanted devices, or experience severe heart failure, have contraindications that, at many locations, lead to their exclusion from MRI studies and MRI-guided interventions.22,26

The major reason for the exclusions lies in the current inability to detect and intervene quickly should a cardiac event occur inside the MRI bore. For rapid detection of a cardiac event, MRI-compatible 12-lead ECG systems are currently being developed,27 although the existing MRI-compatible monitoring systems can detect hemodynamic and oxygenation emergencies.

The ability to rapidly intervene requires MRI-compatible external defibrillation systems, which are currently unavailable. In the authors’ sites, a cardiac event that occurs inside the bore entails (1) taking the patient out of the bore, (2) removing the MRI coils and transferring the patient from the MRI table to a transport gurney or undocking the MRI table and using it for transport, (3) moving the patient out of the MRI suite or finding an area in the suite, which is well removed from the magnetic field, and (4) attaching defibrillation pads and administering the defibrillation pulses. This sequence delays the administration of defibrillation by several minutes, which presents a significant risk of increased mortality because the survival rate is reduced by 5.5% every minute.28

In this, we constructed a prototype external defibrillation system for use in the MRI suite. The system is intended to be attached to patients while they are within the bore, enabling the immediate administration of defibrillation without moving the patient. We tested the system performance during MRI, to check for effects on image quality, and to verify effective defibrillation.

Methods

Specifications for the MRI-Conditional Defibrillation System

We specified that the defibrillation system performs all its diagnostic functions while inside the MRI suite and be ready for immediate administration of high-voltage (2–3 kilovolt) defibrillation pulses, with the subject in the bore of the MRI scanner. This required that the defibrillation generator be present inside the magnet room, and that all the associated cabling and surface pads remain attached to the subject during their entire sojourn inside the MRI scanner. Because the generator continuously tests its connection to the defibrillation pad by sending AC electric waves to the pads and measuring their impedance (and will not fire if it detects changes in impedance), the connection at this frequency between the generator and the pads needed to be maintained.

Second, we specified that the MRI performance of the scanner not be impaired by the permanent attachment of a working defibrillation system to the subject. This required that the MRI signal-to-noise ratio (SNR) not be reduced appreciably by the presence of the system. This required that we reduce interference at the scanner Larmor frequency to minimum. In addition, it required that the susceptibility artifacts created by the surface pads be spatially restricted, to avoid masking of important anatomy.

The third specification was related to patient safety. As per American Society for Testing and Materials and International Electrotechnical Commission29,30 requirements, we demanded that the common-mode interference at the MRI Larmor frequency (∼63.8 MHz at 1.5 T), we constructed a dedicated low-pass filter to remove this interference. The filter (Figure 1B), which was placed at the output of the generator, contained elements to remove both common- and differential-mode RF interference. It was designed to (1) withstand the 3 kilovolt pulses emitted during defibrillation and (2) enable unattenuated transmission of the 67 kHz sine wave continuously emitted by the generator to check the integrity of the connection to the surface pads. Specifically, it was designed to provide ≈120 dB attenuation at 63.8 MHz, while allowing frequencies below 22 MHz to pass freely. The defibrillation filter consisted of 3 sections. The section closest to the generator consisted of a common-mode choke, which was constructed from 5 turns of RG142 high-voltage coaxial cable wound around a central ferrite core (Amidon FT-140-33, Costa Mesa, CA). The second section was a 9-pole Chebyshev LC filter with a 25 MHz –3 dB cutoff and 3 dB band-pass ripple. The 25 MHz cutoff was chosen to provide adequate stop-band attenuation, while minimizing inductor values, thus avoiding parasitic self-resonance at the MRI proton frequency. Allowing a 3 dB passband ripple relaxed some component requirements in the design. The third section was a common-mode choke. The common-mode chokes prevented cable-scaffold currents and suppressed RF noise picked up in the MRI room. Three-kilovolt chip capacitors and high-current inductors were used as components within the filter to accommodate the large power pulses passing through during defibrillation.

Because the Zoll defibrillator chassis contained ferromagnetic components, it needed to be placed outside the MRI scanner 5 Gauss magnetic field lines, and thus was placed ≈1.5 m away from the bore on the side of the MRI gantry. To prevent MRI-induced heating of the cables connecting the low-pass filter and the surface pads, we constructed a 4-m long twisted pair high-voltage cable using 3 American Wire Gage gauge insulated copper cable. RF traps (Baluns) tuned to the MRI frequency were placed at 0.30 m increments along the entire cable. We used floating RF traps,31 positioned along the outside of the cable and inductively coupled to the cable, instead of filters placed within the cable (eg, in-line filters, such as a band-reject filter), so that (1) these traps would not receive the full power of the defibrillation current and (2) since their role was to attenuate waves at the MRI frequency, induced by the MRI scanner body coil onto the cable during imaging, which are primarily common mode in nature, for which such filters are sufficient. The Baluns were constructed from 2 concentric circular copper tubes (0.1 m in length and 0.01 and 0.025 m in diameter, respectively), connected by tuning capacitors. These waves, if allowed to freely propagate, might induce temperature rises in the cable or at the surface pads. At the cable distal end, we constructed connectors for the surface pads.

We used commercial Zoll Medical Stat Padz HVP multifunctional defibrillation pads (electrodes). We shortened the leads and replaced the connectors of these disposable electrodes, so that they would (1) enable rapid switching between fibrillation and defibrillation, (2) mate with the connectors we added to the distal portion of our cable, and (3) use the cable and Baluns we built, which are not disposable, but otherwise did not modify these pads.

Fibrillation System

To test the operation of the defibrillation system in a swine model, we used a custom-built system to induce fibrillation. Using mechanical switching, this fibrillation system delivered high-voltage, 60 Hz, electric pulses to the Zoll defibrillation pads, inducing cardiac arrest, at which point the connections were switched, disconnecting the fibrillator and connecting the defibrillator, to defibrillate the animals.

Theoretical Analysis of Defibrillation Inside the MRI Magnetic Field

There is a concern that defibrillation inside the MRI’s large static magnetic field may cause excessive displacement of the subject, as
Figure 1. The magnetic resonance imaging (MRI)–conditional defibrillation system. A, Block diagram illustrating the function of the system components, with arrows showing the defibrillation current flow path. Low-pass filter (B) circuit diagram and (C) frequency response. The filter passes frequencies <22 MHz and has ≈120 dB attenuation at 63.8 MHz. D, Photograph of the system components. Red arrows point to the key parts. RF indicates radiofrequency.
a result of the Lorentz force $F_\text{L}$ created by the interaction between the static field $\mathbf{B}_0$ and the large current induced by defibrillator $i$, which is 20 to 24 A during the M-series biphase defibrillation (Zoll 9650-4050-01-M Series Service Manual Rev. R; https://www.zoll.com/WorkArea/DownloadAsset.aspx?id=22691), and flows through a conducting cable (eg, the body) for a length $dL$.

$$F_\text{L} = i \times dL \times \mathbf{B}_0$$

We take the extreme case, where $\mathbf{B}_0$ is along $z$ and $i \times dL$ is entirely in the $xy$ plane, so the cross-product is maximal. We can then estimate the maximal value of $F_\text{L}$ as $24 \times 0.4 \times 1.5 = 3.6 N$ at 1.5 T (at 3 T: 12 N), where we have chosen $dL$ to be the chest cross section. If we further assume that the chest mass is at least 40 kg, then the maximal acceleration because of this force is as follows:

$$a = \frac{F_\text{L}}{m} = 0.17 m/s^2 \left( at 3T: 0.33 m/s^2 \right)$$

### Case I. Motion Against Gravity

If the electrodes lie on the left and right side of the chest, so the current flows along the $x$ plane (the left–right direction), then $F$ is along $y$ (the anterior–posterior direction). Here, it works against the gravitational acceleration $a_g$. Because $a_x < a_g = 9.8 m/s^2$, there will not be an elevation of the chest because of the Lorentz force at 1.5 or 3 T, except perhaps in light subjects, such as small children.

### Case II. Motion Against Friction

If the electrodes lie on the top and bottom of the abdomen, so the current flows along the $y$ plane (the anterior–superior direction), then the force is directed along the $x$ plane (left–right direction). Here, it competes with the frictional acceleration $a_{friction}$. Because $a_y = a_{friction}$, its magnitude depends on the material frictional coefficient $\mu$. For polytetrafluoroethylene (the plastic from which the MRI bed is made), $\mu=0.04$ to 0.2 (www.engineeringtoolbox.com/friction-coefficients-d7778.html), so $a_x < a_y = 0.4 \ m/s^2$. This is true at both 1.5 and 3 T. For skin against cloth, $\mu$ is somewhat larger, 0.3 to 0.8 (http://nopr.niscrirres.nist.gov/bitstream/123456789/19305/1/IJFTR%2019(3)%20151–155.pdf). This means that little or no sliding of the chest in the left–right direction will occur because of the Lorentz force.

### MRI Pulse Sequences

To test the system for SNR reduction because of the presence of the defibrillation system, we imaged volunteers with balanced steady state–free precession (SSFP) MRI sequences. Sequence parameters were as follows: Time to Repeat/Time to Echo=2.3/1.14 ms, flip angle 80°, acquisition matrix 160x256, field of view 40x40 cm, slice width 7 mm, bandwidth 1149 Hz/pixel.

To test the system for RF-induced heating, a 2-dimensional SSFP sequence, with an SAR rating of 4.4 W/kg as determined by the Siemens MRI scanner software, was run continuously for 11 minutes on a swine model, with temperatures measured below the center of the pads, within the gel that electrically links the pads to the subject skin, using 2 Neooptix (Quebec, QC, Canada) fiber optic probes. Additional temperature probes were placed on the high-voltage cabling, emphasizing regions of skin contact. Imaging was performed first with the heart at magnet isocenter. Thereafter, imaging was performed centered on the subject’s head, with the swine’s abdomen also moved laterally toward the scanner bore. Localizing the swine in this manner placed the defibrillation pads 20 cm off-center along the superior–inferior ($z$) direction and 15 cm in the left–right ($x$) direction, corresponding to 10 cm along $z$ and 25 cm along $x$ away from the body coil rungs, which maximized the electric field induced by the MRI body coil on the pads and associated cabling. SSFP parameters were as follows: Time to Repeat/Time to Echo=2.37/1.14 ms, flip angle 76°, acquisition matrix 192x144, field of view 50x40 cm, slice width 20 mm, and bandwidth 1532 Hz/pixel. Two-dimensional Turbo spin echo scans were also performed at the 2 scan locations, to test stronger RF pulses but with lower overall SAR ratings (3.3 W/kg). Turbo spin echo parameters were as follows: Time to Repeat/Time to Echo=774/2 ms, flip angle 180°, Echo Train length=129, acquisition matrix 256x256, field of view 30x30 cm, slice width 5 mm, and bandwidth 781 Hz/pixel.

### Study Population and Experiments Performed

Healthy volunteers ($n=2$) were imaged with Institutional Review Board approval, and after providing informed consent, in a 1.5-Tesla MRI scanner (Aera, Siemens, Erlangen, Germany). The volunteers were imaged using high SAR and high gradient cardiac SSFP sequences with the entire defibrillation system connected. Pads were placed on the middle of the sternum and the back (Figure 2). To assess RF interference noise levels, imaging was performed with the Zoll Medical generator in 3 states: (1) shutdown (OFF), (2) ON and powered by its internal battery, and (3) ON and powered from a wall power outlet (AC power) inside the magnet room. SNR measurements were conducted on the cardiac images, assessing SNR as the ratio between the mean signal in several selected regions of interest, and the SDs of the signal outside the anatomy, as measured at multiple regions of interest. Ten regions of interest in total were assessed.

Swine ($n=2$) were studied with local Institutional Animal Care and Use Committee approval, and the procedures were followed in accordance with the institutional guidelines. To assess the operation of the MRI-conditional defibrillation system, we used two 40-kg swine, with previously induced chronic left ventricular infarcts. They were anesthetized and intubated. During the experiments, the swine were defibrillated ($n=4$) using the custom fibrillation unit and, thereafter, defibrillated using the modified defibrillation system, using a 200 J, 8 ms biphasic pulse, which is a regulatory cleared protocol of the Zoll system.

Two cardiac events were induced within a conventional animal fluoroscopy suite, which tested normal (non-MRI) defibrillation functionality of the modified system. Four cardiac events were then induced inside a 1.5 T Siemens 0.7-m patient-bore Avanto scanner. Before the cardiac arrest, during the arrest, and during the recovery to sinus rhythm, which followed defibrillation, the swine were monitored using MRI-compatible hemodynamic monitors (Invivo Precess 3160, Gainesville, FL).

### Results

#### Human Imaging Experiments

MRI SNR (Figure 2) during SSFP imaging was measured to be $-1.6 \pm 0.4$ and $-0.6 dB \pm 0.4\%$, as measured in the 10 regions of interest, with the defibrillator ON and running on its internal battery or on line power, respectively, relative to the SNR obtained with the defibrillation system shutdown (OFF). The variation of the SNR over the imaged field of view was small, and no distinct regions or patterns of noise were observed.

During SSFP imaging (Figure 2B), artifacts from the Zoll Medical Stat pads extended only 6 mm below the skin surface. As a result, visualization of important anatomy was not affected.

#### Swine Heating Experiments

During the 11-minute continuous imaging experiment using the 4.4 W/kg SSFP sequence, and using the extremely off-center defibrillation pad localization, the peak temperature rise observed below the defibrillation pads was 1.4°C (Figure 3), which is within the US Food and Drug Administration guidelines. Measured high-voltage cable temperature rises were all <0.1°C. Turbo spin echo scans at this location, as well as SSFP scans at other locations, resulted in smaller (1.2 and 1.3°C, respectively) temperature increases. The volunteers’ skin regions below the pads were examined at the end of the heating tests, and no burns were observed, nor were any sensations reported.
Swine Fibrillation and Defibrillation Experiments

During experiments performed in the animal fluoroscopy laboratory (Figure 4), the modified defibrillator system chassis (generator and low-pass filter) was placed on a table ≈1.5 m from the swine table, whereas the fibrillation apparatus was placed on the fluoroscopy table. Stimulation from the apparatus produced ventricular fibrillation, which was reversed with either a single (n=1) or 2 repeated (n=1) defibrillation pulses. The Zoll Medical generator ECG display recorded the cardiac events and recovery properly (Figure 4B).

Figure 2. Human testing of the defibrillation system. A, Photograph of a volunteer before being inserted into the 1.5-T magnetic resonance imaging (MRI), with black arrows pointing to the location of key components of the defibrillation system. B, Results of steady state–free precession (SSFP) imaging performed to assess signal:noise ratio. Two slices are shown, acquired at 3 different conditions, with the defibrillation system shut down (left column), with the defibrillation system ON but using internal battery supply (middle column), and with the defibrillation system ON but using AC wall power supply (right column). Red arrows on images point to artifacts produced by the presence of the defibrillator pads on the subject skin. RF indicates radiofrequency.

Figure 3. Radiofrequency heating during imaging in a swine model. The swine’s head was placed at magnet isocenter and the torso moved laterally (left–right), to maximize the electric field (from radiofrequency pulses transmitted by the magnetic resonance imaging [MRI] body coil), which is induced onto the defibrillation pads and cables, which are found at +20 cm superior–inferior and +15 cm left–right. A steady state–free precession (SSFP) sequence with 4.4 W/kg specific absorption rate was used. The blue and red graphs denote temperatures recorded underneath the 2 defibrillation pads (between the pads and the skin). The red curve shows pad2, which leaned against the MRI bore wall (eg, the most extreme off-center situation) and consequently demonstrated the largest temperature variation.
During experiments performed within the MRI suite (Figure 5A), the modified defibrillator system chassis was placed on the right side of the MRI magnet, ≈1.5 m from the front of the MRI gantry. At this distance, it was easy to operate the M-series generator, and there was no magnetic attraction of the chassis. Fibrillation inside the MRI scanner resulted in ventricular fibrillation (Figure 5B), which was reversed with a single defibrillation pulse in each of the 4 animals studied. The Zoll Medical generator single-lead ECG display did not properly record the ventricular fibrillation event because it was obscured by large gradient–induced voltage overlays on the ECG trace during imaging,32,33 but this was noted properly using the MRI-compatible hemodynamic display (Figure 5C), allowing for the timely delivery of defibrillation. This 100% success rate, although obtained from a small sample set, is equivalent to that reported for defibrillation following the electrically induced fibrillation using the conventional defibrillator systems, when there is <1 minute delay post ventricular fibrillation.28

During defibrillation in the x-ray suite and inside the MRI, heavy swine torsos were elevated by ≈0.10 m. This did not result in any damage to the MRI scanner, the imaging coils, or the defibrillation apparatus.

**Discussion**

We demonstrated that an MRI-conditional defibrillation unit could be constructed by modifying a commercial defibrillation system. None of the features or capabilities of the commercial defibrillator was sacrificed by the modifications performed. The modified system did not significantly reduce the MRI scanner imaging SNR, nor did its cabling and pads produce heating. As a result, we feel that it can be both safe and effective for use during MRI and MRI-guided intervention.

Although not explored in this study, the system we developed can also be used to perform temporary external transcutaneous pacing of the heart. This will be investigated in future studies because it will further increase the use of the system.

The swine elevation (jump) during defibrillation did not damage the scanner, nor did it injure the subject. We used a wide-bore 0.7-m patient-bore MRI for these experiments, whereas most current scanners have a 0.6-m patient bore. However, swine dimensions and weight are smaller than most...
patients. Further investigation is, therefore, required to determine whether it is preferable to perform defibrillation within the bore, or more prudent to withdraw subjects from the bore and defibrillate on the MRI table, to facilitate performance of cardiopulmonary resuscitation. An additional solution to in-bore defibrillation is to use dedicated defibrillation protocols, with changes in pulse profiles or electrode configurations, which greatly reduce muscle contractions.34,35

The differences between the system SNR under battery and line-power operation point to the sources of the residual noise. It mainly originates from RF interference emission through the power cable, which we are currently engaged in removing, with a smaller contribution from radiation from the chassis, which we are also addressing.

Some sites performing interventional procedures may prefer to use other defibrillation electrodes, such as radiotranslucent electrodes, or those made by other vendors. Each of these electrodes will need to be examined to determine whether (1) they are ferrous or create large susceptibility artifacts, (2) they support RF eddy currents, and (3) heat beyond regulatory limits during high SAR imaging. RF heating of electrodes can be further reduced by cutting paths into the electrodes and, thus, reducing RF eddy currents.36 We have also not explored possible issues with pediatric patients, for which (1) the Lorentz forces may play a larger role and (2) special RF heating models are needed. Additional theoretical and in vitro thermal tests may be required before regulatory approval.

The generator chassis was slightly magnetic, and this might be addressed by (1) chaining the generator to the wall so that it could not be inadvertently brought closer to the MRI or (2) by removing the magnetic components in the generator chassis, which requires redesign of the high-voltage stages of the generator.

In addition, the generator ECG display did not properly display ECG traces within the MRI. This may preferably be addressed by linking the generator to an MRI-conditional ECG system.27 We consider the use of such a system a necessity because the gradient-induced voltages during MRI scans are of the order of volts,32 completely masking the real ECG, so the gradient-induced voltages must be removed to obtain reliable ECG traces.33 This system will also serve as a reliable alarm, notifying the user of a cardiac event at an earlier stage, should it occur inside the MRI. The Zoll system supports such a connection. It is important that this ECG system be able to sustain the defibrillation pulses, in terms of (1) preventing the large voltages from damaging the ECG system front end, (2) preventing large currents from propagating on the ECG leads, which can present a heating risk, and (3) reduce the effectiveness of the defibrillation.

Availability of this external defibrillation system to the larger clinical community will require obtaining regulatory approval, which will require performing more extensive testing. We are attempting to find a commercial entity that will undertake this process. In addition, we plan to make adjustments to the defibrillator (such as to the Baluns) to enable it to be used at other MRI field strengths.

Conclusions
An MRI-conditional external defibrillator was constructed and validated. MRI-conditional defibrillators may permit the diagnosis and treatment of higher-risk patient populations within MRI scanners, significantly improving the patient care options for these patients.

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Disclosures
None.

References


CLINICAL PERSPECTIVE
Large populations of patients are currently excluded from cardiac magnetic resonance imaging (MRI) or MRI-guided surgical interventions because of the lack of equipment that can rapidly treat a cardiac event if it occurs inside the MRI. In this study, an external defibrillator, which is MRI compatible, was developed and validated for the resuscitation of subjects who had incurred heart attacks inside the MRI scanner or in the MRI suite. Availability of this defibrillator will permit the use of MRI and MRI-guided interventional procedures to diagnose or treat severely ill patients for symptoms for which MRI holds distinct clinical benefits. In addition, such a defibrillator is essential to the emergence of MRI-guided treatment of cardiac arrhythmia (atrial fibrillation and ventricular tachycardia) disorders.
A Magnetic Resonance Imaging–Conditional External Cardiac Defibrillator for Resuscitation Within the Magnetic Resonance Imaging Scanner Bore


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