Effect of radiation induced current on the quality of MR images in an integrated linac-MR system

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Purpose: In integrated linac-MRI systems, the RF coils are exposed to the linac’s pulsed radiation, leading to a measurable radiation induced current (RIC). This work (1) visualizes the RIC in MRI raw data and determines its effect on the MR image signal-to-noise ratio (SNR) (b) examines the effect of linac dose rate on SNR degradations, (c) examines the RIC effect on different MRI sequences, (d) examines the effect of altering the MRI sequence timing on the RIC, and (e) uses a postprocessing method to reduce the RIC signal from the MR raw data.

Methods: MR images were acquired on the linac-MR prototype system using various imaging sequences (gradient echo, spin echo, and bSSFP), dose rates (0, 50, 100, 150, 200, and 250 MU/min) and repetition times (TR) with the gradient echo sequence. The images were acquired with the radiation beam either directly incident or blocked from the RF coils. The SNR was calculated for each of these scenarios, showing a loss in SNR due to RIC. Finally, a postprocessing method was applied to the image k-space data in order to remove partially the RIC signal and recover some of the lost SNR.

Results: The RIC produces visible spikes in the k-space data acquired with the linac’s radiation incident on the RF coils. This RIC leads to a loss in imaging SNR that increases with increasing linac dose rate (15%–18% loss at 250 MU/min). The SNR loss seen with increasing linac dose rate appears to be largely independent of the MR sequence used. Changing the imaging TR had interesting visual effects on the appearance of RIC in k-space due to the timing between the linac’s pulsing and the MR sequence, but did not change the SNR loss for a given linac dose rate. The use of a postprocessing algorithm was able to remove much of the RIC noise spikes from the MR image k-space data, resulting in the recovery of a significant portion, up to 81% (Table II), of the lost image SNR.

Conclusions: The presence of RIC in MR RF coils leads to a loss of SNR which is directly related to the linac dose rate. The RIC related loss in SNR is likely to increase for systems that are able to provide larger than 250 MU/min dose. Some of this SNR loss can be recovered through the use of a postprocessing algorithm, which removes the RIC artefact from the image k-space. © 2012 American Association of Physicists in Medicine. [http://dx.doi.org/10.1118/1.4752422]

Key words: linac-MR, radiation induced current

I. INTRODUCTION

Our research group has integrated a linear accelerator (linac) with a magnetic resonance imaging (MRI) system. This system will provide real-time, intrafractional images with tumor specific contrast to allow significant reductions in margins for the planning target volume. As a result, both improved normal tissue sparing and dose escalation to the tumor will be possible, which are expected to improve treatment outcomes.

The radio frequency (RF) coils used in MR imaging are exposed to the pulsed radiation of the linac in the integrated...
linac-MR system. The receive coil will either sit close to or right in contact with the patient. Therefore, there will be beam orientations in a treatment plan where the coil will be irradiated. This has been shown to result in instantaneous currents being induced in the MR coils—called radiation induced current (RIC).4 RIC has been widely reported on in various materials when exposed to various sources of radiation.5-9 These extraneous currents have the potential to adversely affect MR imaging by distorting the RF signal being measured by the RF coils. Our more recent results have shown that the RIC signal in RF coils can be reduced with the application of appropriate buildup material to the coils.10 This buildup method was effective with planar or cylindrical coil geometries and was unhindered by the presence of magnetic fields. This work explores another method for RIC removal that does not involve altering the RF coils, but instead uses image processing techniques. Recently published work by Yun et al. discusses the importance of imaging signal-to-noise ratio (SNR) for real-time tumor tracking3 and this provides the motivation for the use of a postprocessing algorithm to recover some of the SNR lost due to RIC. Their work showed that the accuracy of the autocontouring algorithm was reduced when the field strength was reduced from 0.5 T to 0.2 T, due to the decrease in contrast-to-noise ratio (CNR).3 The measured, average centroid root mean squared error in their tracking algorithm was increased by factors of 1.5 and 2.4, respectively, in their spherical and nonspherical phantoms (see Table III in Ref. 3). At a given field strength, an increase in the image noise due to the RIC noise spikes will reduce both the SNR and the CNR, thus further decreasing the accuracy of the autocontouring and tracking method. At high magnetic fields the RIC artefact may not be of great importance due to the inherently higher SNR. However, performing fast imaging, which is required for real-time tracking, at low fields dictates that we are in a SNR challenged environment and as such, any further degradation of SNR is highly undesirable.

In this work, we image phantoms in the linac-MR system in the presence of pulsed radiation from the linear accelerator. These experiments clearly demonstrate the presence of RIC in the MRI raw data, i.e., k-space. The purpose of this work is to (a) visualize the RIC in MRI raw data and determine its effect of the MR image quality, specifically the SNR (b) examine the effect of linac dose rate, in monitor units (MU) per minute, on the SNR degradation caused by the RIC, (c) examine the RIC effect on different MRI sequences, (d) examine the effect of altering the MRI sequence timing, specifically the repetition time (TR), on the visual appearance of the RIC in MRI raw data, and (e) use postprocessing methods to remove the unwanted RIC signal from the MR images.

II. MATERIALS AND METHODS

The linac-MRI system used in these experiments is that described by Fallone et al.,2 and shown schematically in Fig. 1. The system is comprised of a 0.22 T biplanar magnet from MRI Tech Co. (Winnipeg, MB, Canada) and a 6 MV linear accelerator with its beam directed to the imaging volume of the magnet. The x-ray beam direction is perpendicular to both the main magnetic field and the superior-inferior orientation of the patient. It should be noted that Fig. 1 does not clearly show the RF coil. The B1 field in MRI is perpendicular to the main magnetic field. Thus, the axis of the RF coil is either along the patients’ head-foot direction or along the radiation beam direction. Moreover, the RF coil sits closer to the patient for the best SNR in the receive signal. The standard RF coils are either cylindrical with axis along the patients’ head-foot direction or surface coils resting directly on the patients’ skin. In all of these cases, the coil conductor will be directly exposed to the radiation. The maximum gradient strength of the MR system is specified as 40 mT/m and the MR system is controlled using a TMX NRC console (National Research Council of Canada, Institute of Biodiagnostics, Winnipeg, MB, Canada). The console software is PYTHON-based [Python Software Foundation, Hampton, NH (Ref. 11)] to allow full user control of the development and modification of pulse sequences. Analogic (Analogic Corporation, Peabody, MA) AN8295 gradient coil amplifiers and AN 8110 3 kW RF power amplifiers are used.

The linac components are composed of salvaged parts from a decommissioned magnetron-based Varian 600C system, which include the straight-through waveguide (without bending magnet). The distance of the linac target to the magnet center is 80 cm. Presently, the MV x-ray beam has primary collimators and the final prototype design will include secondary collimators and the multileaf collimator (MLC).2 As such the radiation field size was larger than the coils, so the entirety of the RF coils was irradiated during the experiments. However, our previous work (Ref. 4, Fig. 8) has shown that the RIC amplitude increased as the irradiated area of the coil conductor is increased.

Two RF coils were used in the imaging experiments. The first was a small, ~3 cm diameter solenoid coil with 14 turns of wire. The tuning and impedance matching of this coil is accomplished by variable capacitances and it contains an integrated pin-diode transmit/receive switch. All active components are outside the volume of the solenoid such that these can be placed outside the radiation beam. The second coil was a 10 cm diameter solenoid coil containing 5 concentric
rings made of 0.64 cm diameter copper pipe. The tuning of the coil is accomplished by a variable capacitance while the impedance matching is accomplished with a variable inductor. As with the smaller coil, this coil contains an integrated pin-diode transmit/receive switch with the active components residing outside the solenoid volume. Both coils were constructed by NRC and resonate nominally at the appropriate frequency of 9.3 MHz for the 0.22 T MRI.

The phantom used in the smaller coil was an acrylic rectangular cube, 15.95 × 15.95 × 25.4 mm³, with 3 holes of diameters 2.52, 3.45, and 4.78 mm drilled into it. The cube was then placed in a 22.5 mm diameter tube and filled with a 10 mM solution of CuSO₄. This arrangement fills the holes in the cube with the CuSO₄ creating three circular signal regions in the MRI image. The phantom used in the 10 cm diameter coil consisted of four tubes of 27 mm diameter filled with a solution of 61.6 mM NaCl and 7.8 mM CuSO₄. The tubes were stacked into a 2 × 2 matrix arrangement and held together with an adhesive tape. This arrangement again created four circular signal regions in the MRI image.

II.A. Effect of RIC and linear accelerator dose rate on MR images

This experiment was designed to determine the effect of RIC on the SNR in MRI images including the impact of the linac dose rate. A standard gradient echo sequence was used in all experiments. For the phantom in the smaller coil the imaging parameters were as follows: slice thickness – 5 mm; acquisition size – 512 (read) × 128 (phase); field of view (FOV) – 50 × 50 mm²; TR – 300 ms; echo-time (TE) – 35 ms; flip angle – 60°; no signal averaging. For the phantom used in the larger coil the imaging parameters were as follows: Slice thickness – 3.5 mm; acquisition size – 256 (read) × 128 (phase); FOV – 100 × 100 mm²; TR – 300 ms; TE – 35 ms; flip angle – 90°; no signal averaging. For the smaller coil, 512 points in the read direction were chosen for easy visualization of the RIC artefact; more points in the read direction means a longer acquisition window, which in turn leads to more radiation pulses being present during signal acquisition.

Images of both phantoms were first obtained with the linac not producing radiation. The same imaging was then repeated with the linac producing radiation at 50, 100, 150, 200, 250 monitor units per minute (MU/min, i.e., the dose rate). The imaging experiments in the presence of the radiation beam were further divided into two parts. In the first experiment, the radiation was directly incident on the RF coils. In the second experiment, a lead block was placed in the beam path to attenuate completely the radiation from reaching the RF coil. This was done to ensure that any effect seen in the MR images was caused only by the direct irradiation of the coils, resulting in RIC in the coil and not due to any residual RF noise. The residual RF noise, if it exists, will still reach the coil even if the x-ray beam was completely attenuated by the lead block. The method and effect of RF shielding for this system has been previously described. This means that a total of 11 (beam off, beam on at five different dose rates, beam on but blocked at the same five dose rates) different imaging conditions were examined for each phantom and coil combination.

Five images were taken in each condition to assure reproducibility and to provide statistical information. The resulting images were then analyzed first by calculating the SNR of the image and second, by examining the k-space data associated with each image, using appropriate window and level, to visualize the RIC artefact (see Fig. 5). The SNR was calculated by taking the mean of the signal divided by the standard deviation of the background noise. For each of the 11 imaging conditions the mean and standard deviation of the five SNR values were calculated.

II.B. Dependence of RIC artefact on imaging sequence

The effect of the MR imaging sequence on the RIC artefact was examined by repeating the imaging experiments from Sec. II.A, using a spin echo sequence and a balanced steadystate free precession (bSSFP) sequence instead of the gradient echo sequence used in Sec. II.A. The small coil described above was used for both sequences. SNR was calculated as in Sec. II.A.

The imaging parameters for the spin echo sequence were: slice thickness – 5 mm; acquisition size – 256 (read) × 128 (phase); FOV – 50 × 50 mm²; TR – 300 ms; TE – 30 ms; no signal averaging; flip angle – 90°. The imaging parameters for the bSSFP sequence were: slice thickness – 5 mm; acquisition size – 128 (read) × 128 (phase); FOV – 50 × 50 mm²; TR – 18 ms; no signal averaging; flip angle – 60°.

II.C. Dependence of RIC artefact on imaging parameter TR

The next imaging experiment was done by keeping the linac dose rate constant at 250 MU/min and the imaging parameters identical to those in Sec. II.A, except for the repetition time, TR, which was changed. These experiments were only performed with the smaller coil and the gradient echo sequence was used. Images were acquired at TR values of: 299, 299.8, 299.9, 300, 300.1, 300.2, 301, 302, 303, 304, and 305 ms. This investigation examined the relationship between the RIC artefact and the MR sequence timing. The SNR and k-space data were again examined.

II.D. Removal of RIC artefact from MR data using postprocessing

Finally, the software program MATLAB (The MathWorks, Inc., Natick, MA) was used as a postprocessing tool in an attempt to remove the RIC artefact from the image k-space data and restore some of the SNR lost due to RIC. The algorithm is similar in application to an adaptive filter used to removed speckle noise from synthetic aperture radar images as discussed by Russ (see Ref. 13 Chap. 3, p. 165 top), which uses a neighborhood comparison of pixel brightness, with a threshold based on the average and standard deviation, and
replaces those above the threshold with a weighted average value of the neighborhood.

The algorithm searches pixel-by-pixel for anomalous signal spikes in k-space and then removes them. These spikes are found by searching the k-space data for pixels with intensities above a global threshold value; the global threshold value was the average background plus three standard deviations. The average background and standard deviation are determined from a group of pixels near the edge of the k-space image (thus ensuring that it is background).

Once an anomalous pixel is found, its magnitude is then compared to the mean pixel magnitude in the local neighborhood surrounding the pixel to determine whether the pixel resides in a background region (i.e., toward the edges of k-space) or in a signal region (near the center of k-space). If the pixel’s value is larger than the local average (plus 3 standard deviations) then the anomalous pixel lies in the background regions, otherwise it lies in the signal region. In other words, in order for the pixel to be replaced, its intensity has to be larger than both the global threshold and local average before it is replaced. The number of pixels in the local neighborhood used for comparison in this work was the 5 × 5 square centered on the point of interest. If the algorithm determines that the anomalous pixel is in a background region, the pixel value is changed to that of the average background. If the algorithm instead determines that the anomalous pixel is in a signal region, then no action is taken.

It is obvious that this algorithm will not eliminate all RIC spikes from the k-space data, as it will not be able to discern between RIC signal and the MR signal near the center of k-space. However, this may be acceptable as the RIC spikes near the center of k-space have a minimal effect on SNR because the spikes are sparsely distributed compared to the MR signal. Also, the magnitude of the RIC noise spikes is small compared to the MR signal near the center of k-space. The MR image was reconstructed from the processed k-space data. The SNR was then recalculated and compared to the original values.

II.D.1. Comparison to median filtering

The postprocessing algorithm described above is a custom scripted algorithm. Standard image processing techniques may also be used to remove the RIC spikes in the k-space. In order to compare the postprocessing method with the standard techniques, two common median filters were investigated. A median filter sets a pixel to the median value of the pixels in the user specified neighborhood around it. This filter is widely used to remove impulse noise.13,14 as it will replace pixels with excessively large or small values with a more “normal” value. First, the standard median filter (function “medfilt2” in MATLAB) was applied globally to the k-space. The “symmetric” option was used in MATLAB that causes the boundaries of the images to be extended symmetrically to allow the filter to work at the edges of the image. Second, the adaptive median filter as described by Gonzalez and Woods14 was also investigated using the MATLAB implementation “adpmedian” as given in Ref. 15. The “symmetric” option is also used in the adaptive median filter. The adaptive median filter, “adpmedian,” contains a condition which will cause the selective replacement of pixels with the local median values. The algorithm will determine the minimum and maximum values in the neighborhood of the pixel of interest, and then if the pixel is either larger than the maximum or smaller than the minimum, the median filter is applied. However, if the pixel value is between the minimum and the maximum values, then the pixel value remains unaltered. This selective application will effectively remove impulse noise while preserving more of the fine detail in the image.14

Typically, median filters are applied directly to the MR image, and not to the k-space data, as this is where the impulse noise is seen. However, since our postprocessing algorithm is applied to the k-space data, both median filters were also applied to the k-space data. To avoid SNR gains related solely to non-RIC related noise reduction, the filters were applied to all images, whether acquired with or without radiation. The data are then presented as a percentage calculated using Eq. (1):

\[
\text{Percentage of non-RIC SNR} = 100 \times \frac{\text{SNR}_{\text{filter, radiation incident on coil}}}{\text{SNR}_{\text{filter, beam blocked}}} \times \frac{\text{Images acquired at same dose rate}}{}
\]

where “percentage of non-RIC SNR” is the percentage of the original SNR, calculated from the MR data with the radiation beam blocked, “SNR_{filter, radiation incident on coil}” is the SNR calculated after the filter has been applied to the MR data acquired with the radiation striking the coil, and “SNR_{filter, beam blocked}” is the SNR calculated after the filter has been applied to the MR data acquired with the radiation beam blocked. Both MR data sets are acquired with the same linac dose rate.

III. RESULTS

III.A. Effect of RIC and linac dose rate on MR images

The first three columns of Tables I and II show the SNR values calculated for each imaging condition described in Sec. II.A for the phantoms imaged with the 10 cm and 3 cm coils, respectively. When the lead block stops the radiation beam, the radiation dose rate is 0.0 18.2 ± 0.4

<table>
<thead>
<tr>
<th>Linac dose rate (MU/min)</th>
<th>Radiation beam blocked by lead block</th>
<th>Radiation beam incident upon MRI RF coil after RIC noise is removed</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>18.2 ± 0.2</td>
<td>17.9 ± 0.1</td>
</tr>
<tr>
<td>50</td>
<td>18.0 ± 0.4</td>
<td>17.7 ± 0.1</td>
</tr>
<tr>
<td>100</td>
<td>17.8 ± 0.3</td>
<td>17.4 ± 0.3</td>
</tr>
<tr>
<td>150</td>
<td>18.2 ± 0.3</td>
<td>16.9 ± 0.2</td>
</tr>
<tr>
<td>200</td>
<td>17.8 ± 0.1</td>
<td>16.5 ± 0.2</td>
</tr>
<tr>
<td>250</td>
<td>17.8 ± 0.1</td>
<td>16.2 ± 0.3</td>
</tr>
</tbody>
</table>

Medical Physics, Vol. 39, No. 10, October 2012
TABLE II. SNR for images acquired with 3 cm coil using a gradient echo sequence. SNR was calculated by taking the mean of the signal in the magnitude image divided by the standard deviation of the noise in the real image.

<table>
<thead>
<tr>
<th>Linac dose rate (MU/min)</th>
<th>Radiation beam blocked by lead block</th>
<th>Radiation beam incident upon MRI RF coil</th>
<th>After RIC noise is removed</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>19.7 ± 0.3</td>
<td>~</td>
<td>19.1 ± 0.4</td>
</tr>
<tr>
<td>50</td>
<td>19.5 ± 0.4</td>
<td>18.7 ± 0.3</td>
<td>19.0 ± 0.2</td>
</tr>
<tr>
<td>100</td>
<td>19.5 ± 0.3</td>
<td>18.0 ± 0.3</td>
<td>19.0 ± 0.2</td>
</tr>
<tr>
<td>150</td>
<td>19.5 ± 0.4</td>
<td>17.7 ± 0.4</td>
<td>19.0 ± 0.3</td>
</tr>
<tr>
<td>200</td>
<td>19.3 ± 0.2</td>
<td>17.0 ± 0.1</td>
<td>18.7 ± 0.3</td>
</tr>
<tr>
<td>250</td>
<td>19.1 ± 0.4</td>
<td>16.9 ± 0.3</td>
<td>18.7 ± 0.4</td>
</tr>
</tbody>
</table>

from reaching the coil, the SNR stays relatively constant with linac dose rate for both coils; however, when the lead block is removed and the RF coils are irradiated there is a loss in SNR. Furthermore, the loss in SNR increases as the linac dose rate increases. At the maximum dose rate, 250 MU/min, Table I shows a decrease in SNR from 18.2 to 16.2 when compared to the no radiation scenario, representing an 11% loss, or a decrease from 17.8 to 16.2 (9% loss) when compared to the radiation blocked scenario at the same dose rate. At the same 250 MU/min dose rate, Table II shows a decrease in SNR from 19.7 to 16.9 (14% loss) when compared to the no radiation scenario, or a decrease from 19.1 to 16.9 (11.5% loss) when compared to the radiation blocked scenario at the same dose rate. A graphical representation of the data in Table II is shown in Fig. 2.

The other objective of the experiments described in Sec. II.A was to visualize the RIC artefact. As mentioned above, the $k$-space data were examined to accomplish this goal. To illustrate the need to examine the $k$-space data rather than the image itself we can look to Figs. 3 and 4. The two images shown in Figs. 3 and 4 were taken with the 10 cm and 3 cm solenoid coil, respectively, with linac dose rates of 0 and 250 MU/min. Visual inspection alone does not show any artefact due to RIC, although the previous analysis shows a loss in SNR. Figure 5 shows the $k$-space data corresponding to the images in Fig. 4. In the top panel, the $k$-space without radiation is shown. In the bottom panel, the $k$-space data are shown for the case when the linac producing radiation at 250 MU/min that reaches the coil unattenuated. The middle panel shows the $k$-space data from an image taken with a 250 MU/min dose rate where the beam was blocked from reaching the RF coil. Each $k$-space image has the same window and level applied for consistency. Here the RIC artefact is clearly visible in the $k$-space data of the image taken with a 250 MU/min linac dose rate, but is not visible in the other two $k$-space data sets. It is clearly based on Fig. 5 that the vertical lines in $k$-space are due to the RIC as they are only present when the linac is producing radiation and its beam is incident on the RF coil.

III.B. Dependence of RIC artefact on imaging sequence

Tables III and IV contain the calculate SNR values for the imaging experiments using a spin echo and bSSFP sequences, respectively. Again when the radiation beam is stopped by a lead block the SNR remains essentially constant at all linac dose rates. When the radiation beam is incident on the RF coil, there is a loss in SNR that increases with increasing dose rate. At the maximum dose rate, 250 MU/min, Table III shows a decrease in SNR from 19.8 to 16.3 (18% loss) when compared to both the no radiation scenario and the radiation blocked scenario at the same dose rate. At the same 250 MU/min dose rate, Table IV shows a decrease in SNR from 20.2 to 16.5 (18% loss) when compared to the no radiation scenario, or a...
FIG. 5. $k$-space data from images acquired with linac dose rates of 0 and 250 MU/min. The top image was acquired with the radiation not pulsing. The middle image was acquired with a linac dose rate of 250 MU/min but the radiation beam was blocked from reaching the coil; it shows no RIC effects. The bottom image was acquired with a linac dose rate of 250 MU/min and the radiation beam incident on the RF coil; it clearly shows the RIC artefact, which presents itself as near vertical lines in $k$-space.

decrease from 19.9 to 16.5 (17% loss) when compared to the radiation blocked scenario at the same dose rate.

### III.C. Dependence of RIC artefact on imaging parameter TR

The set of imaging experiments described in Sec. II.C was designed to see differences in the RIC artefact, visible in $k$-space, when the imaging repetition time, TR, was changed. It should be stressed that the loss of SNR as a function of dose rate remained unaltered for all values of TR investigated. Figure 6 shows some representative images of the $k$-space data for the TR values specified in Sec. II.C. It is immedi-

![Fig. 6](image)

#### TABLE III. SNR for images acquired with 3 cm coil using a spin echo sequence. SNR was calculated by taking the mean of the signal in the magnitude image divided by the standard deviation of the noise in the real image.

<table>
<thead>
<tr>
<th>Linac dose rate (MU/min)</th>
<th>Radiation beam blocked by lead block</th>
<th>Radiation beam incident upon MRI RF coil</th>
<th>After RIC noise is removed</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>19.8 ± 0.3</td>
<td>–</td>
<td>19.8 ± 0.4</td>
</tr>
<tr>
<td>50</td>
<td>19.9 ± 0.3</td>
<td>19.2 ± 0.6</td>
<td>19.4 ± 0.4</td>
</tr>
<tr>
<td>100</td>
<td>19.7 ± 0.2</td>
<td>17.9 ± 0.6</td>
<td>19.3 ± 0.4</td>
</tr>
<tr>
<td>150</td>
<td>19.6 ± 0.4</td>
<td>17.4 ± 0.5</td>
<td>19.0 ± 0.6</td>
</tr>
<tr>
<td>200</td>
<td>19.4 ± 0.4</td>
<td>16.7 ± 0.7</td>
<td>18.8 ± 0.3</td>
</tr>
<tr>
<td>250</td>
<td>19.8 ± 0.3</td>
<td>16.3 ± 0.5</td>
<td>18.7 ± 0.4</td>
</tr>
</tbody>
</table>

ately obvious that even a small change, 0.1 or 0.2 ms, in TR results in a large change in the $k$-space distribution of the RIC artefact. If the TR is changed from 300 to 299.8 or 300.1 ms, the slope of the lines seen in $k$-space changes dramatically and more lines are seen; 12 lines are seen in top image of Fig. 6 (TR = 300 ms), while 14 are seen in the middle image (TR = 300.1 ms). When the TR is changed by larger amounts (i.e., 1 ms and up) the RIC appears as random background.
spikes, seen in the bottom image of Fig. 6 (TR = 301 ms); a closer inspection shows that the random spikes are still regularly spaced on each read encode line (horizontal line). Aside from the SNR loss, the various $k$-space artefact patterns had no discernible effects in image space. This appearance of $k$-space spikes due to RIC is easily explained by examining the timing between the linear accelerator and the imaging sequence as discussed in Sec. IV.

### III.D. Removal of RIC artefact from MR data using postprocessing

The results of the postprocessing algorithm can be seen qualitatively in Fig. 7 and quantitatively in the final column of Tables I–IV. Figure 7 shows the same $k$-space image, acquired with the 10 cm coil, before and after the algorithm is applied. It is evident that the majority of the RIC artefact has been removed from the $k$-space data. Tables I–IV show that image SNR can be recovered by using the postprocessing algorithm. In all cases, but particularly at higher dose rates (200 and 250 MU/min), there is a useful gain in SNR when the data are processed to remove the RIC noise.

#### III.D.1. Comparison to median filtering

Table V shows the results of the experiments described in Sec. II.D.1. Upon inspection it appears that the median filter does not provide any advantage, and in fact shows a decrease in SNR except in the 100 MU/min case, while the other three filters appear to provide an advantage. The altered postprocessing algorithm replacing anomalous pixels with the median value of their neighborhood, rather than the average background value, provided results which were extremely similar to our original algorithm – with all corresponding values in Table V being within 2% of those for the original algorithm. However, the adaptive median filter appears to provide superior SNR to the original for the case of 150 and 200 MU/min. In this case, the images themselves were further examined to determine the cause of this anomaly.

Figure 8 shows images for the original MR data taken with 250 MU/min, Fig. 8(a), MR data filtered with the original postprocessing algorithm, Fig. 8(b), MR data filtered with median filter, Fig. 8(c), and MR data filtered with adaptive median filter, Fig. 8(d). The postprocessing algorithm altered to use median replacement was omitted because its results were essentially the same as the original postprocessing algorithm. Figure 8(b) shows that our custom postprocessing algorithm does not introduce any obvious distortions or artifacts to the MR image when removing the RIC noise. However, Figs. 8(c) and 8(d) show that applying median filters to $k$-space data can have a deleterious effect on image quality. Figure 8(c) shows that standard, global median filtering of all of the $k$-space data can severely affect the image quality, seen here as severe changes in image brightness and some blurring in the background. Figure 8(d) shows that while the adaptive

<table>
<thead>
<tr>
<th>Linac dose rate (MU/min)</th>
<th>Postprocessing algorithm with median replacement</th>
<th>Median filter</th>
<th>Adaptive median filter</th>
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<tr>
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<td>95.0</td>
<td>89.7</td>
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<td>100</td>
<td>92.4</td>
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<td>95.0</td>
</tr>
<tr>
<td>150</td>
<td>91.0</td>
<td>96.2</td>
<td>94.5</td>
</tr>
<tr>
<td>200</td>
<td>87.9</td>
<td>95.0</td>
<td>94.0</td>
</tr>
<tr>
<td>250</td>
<td>88.1</td>
<td>96.5</td>
<td>95.6</td>
</tr>
</tbody>
</table>
median filter provides an improvement over the standard median filter, there are still artifacts introduced into the image. It appears to have introduced in-homogeneity in image intensity, especially in the background region, but it is also present in the signal region to some extent. Therefore, of the methods tested, the custom postprocessing method we have developed, using either mean or median replacement techniques, is the best suited to removing RIC noise from MR k-space data.

IV. DISCUSSION

The images in Figs. 3 and 4 show that the effect of RIC in MR imaging is not always visually obvious, i.e., the loss in SNR reported in Tables I–IV is not easily noticed in the images. No obvious difference in terms of spatial distortion or intensity in-homogeneity is present between Figs. 3 and 4. When instead, the k-space data are examined visually (Fig. 5) the RIC related noise spikes are immediately obvious. Since these spikes are only present when the linac is producing radiation and the beam is incident on the RF coil, these must be caused by RIC. Tables I–IV also illustrate the dependence of the artefact on the linac dose rate. As the dose rate increases, the SNR of the corresponding images is increasingly degraded. Though not shown, the number of lines seen in k-space also increases with increasing linac dose rate. It is expected as the linac nominally pulses at 180 Hz, but will drop pulses in order to achieve the desired dose rates; thus, the 50 MU/min rate will have approximately 5 times fewer pulses per minute than the 250 MU/min rate. The increasing number of lines seen in k-space at higher dose rates will increase the overall image noise, thus decreasing the SNR as seen in Tables I–IV. This investigation is limited to a dose rate of 250 MU/min due to the limits of the Varian 600C linac. If the linac dose rate is further increased by increasing the pulse repetition frequency, as in the high energy linacs that pulse at 360 Hz, then the SNR degradation at higher dose rates is likely to increase. Also, our previous investigation demonstrated that the RIC amplitude increases proportionally with the irradiated area of RF coil. Thus, the RIC induced degradation of SNR may be different in human sized coils. The largest SNR loss from ~20 to ~16 is not visually discernible in these small images; however, the importance of SNR should be looked at in the context of real-time imaging with respect to autodetection of tumor needed for tracking.

The SNR loss seen with increasing linac dose rate appears to be largely independent of MR imaging sequence. The SNR loss at 250 MU/min, compared to no radiation, for the gradient echo, spin echo, and bSSFP sequences are ~14%, 18%, and 18%, respectively. It was initially expected that the bSSFP sequence might be less affected by the RIC since its acquisition time is very short and less RIC spikes would be present in the MR data. However, in this fast sequence the magnitude of the MR signal is smaller compared to the other two sequences. This smaller signal means that the magnitude of the RIC signal is comparatively larger in this case and although there are fewer RIC spikes in the data, they still raise the background noise level significantly—the lowering the SNR. It should be noted that the bSSFP is the sequence of choice for real-time imaging. It is important to note that 250 MU/min is much lower than the dose rates used in IMRT treatments, which will undoubtedly be implemented on clinical linac-MR systems, and looking at the trend shown in Fig. 2 and Tables I–IV, we can expect SNR losses of greater than 25%, possibly much greater, for our system at these higher dose rates.

Alteration of the MR imaging parameter TR has interesting visual effects that occur due to the interplay between linac pulse timing and TR. Figure 6 demonstrates the effect of changing TR—the lines in k-space change their slope with small changes (0.1 ms) and appear as random spikes with larger changes. Changing the parameter TR only changes the appearance of the RIC artefact; RIC’s effect on SNR remains constant for a given linac dose rate. This phenomenon is explained by examining the timing mechanisms of the linac and MR systems.

The linear accelerator pulses with a frequency of 180 Hz and, if no pulses are dropped, this leads to 1 radiation pulse every ~5.6 ms, with a width of approximately 5 μs/pulse. Examining the original imaging sequence with a TR of 300 ms, a simple calculation reveals 54 (300 ms * 180 pulses/s) radiation pulses per TR—not all of these are seen in k-space because the actual acquisition time, t_acq, for the MR sequence is much shorter than the total TR. This exact integer multiple means that the radiation pulses will occur at approximately the same sampling points along a given read encode k-space line (read encode lines are horizontal lines in k-space which are acquired during a single TR), resulting in a near vertical line in k-space (Fig. 6, top image). Changing TR even by a small amount, such as 0.1 ms to 300.1 ms, means that the number of radiation pulses per TR is no longer an exact integer multiple (300.1 ms * 180 Hz = 54.02) so the radiation pulses will shift along the read direction resulting, in this case, a line sloping downward from left to right (Fig. 6, middle image). When TR is changed by a larger amount, for example, to 301 ms (54.2 pulses per TR), the shift in location of RIC spikes between subsequent read encode lines becomes so large that the noise appears random (Fig. 6, bottom image), though on closer examination the noise spikes due to RIC are still spaced the same distance apart on each read encode line of k-space—as is expected because the linac is still pulsing at 180 Hz. Since the appearance of k-space spikes is the result of deterministic interplay between the linac timing pulses and the MRI read cycle during each TR, it is intriguing to use this information for identifying the anomalous pixels in the k-space. However, the pulse drop servo mechanism of the present linac is not repetitive enough for this scheme to work for all the spikes. Thus, this approach was not pursued.

In a separate investigation, we have shown that the RIC results from the lack of electronic equilibrium in the conductor of the RF coil. In theory, the RIC can be reduced by establishing electronic equilibrium with appropriate buildup material and thickness. However, the placement of buildup may have negative consequences on the patient’s skin dose, thus we have attempted a postprocessing method of RIC removal in this investigation. This simple postprocessing algorithm showed promising results, as seen in the final
column of Tables I–IV, and a large percentage of the lost image SNR was restored. Though not shown, the algorithm works equally well for images taken with various TR values, i.e., it does not depend on the structural distribution of actual spikes in \(k\)-space. The buildup method used in our previous work showed that 55% of the lost SNR was recovered through the application of a simple buildup to the RF coil with a linac dose rate of 250 MU/min.\(^{10}\) Our postprocessing algorithm compared very well to this as the data in Tables I–IV show recovery of 50%, 82%, 69%, and 38%, respectively, for the case of 250 MU/min dose rate.

Both a median filter and an adaptive median filter were shown to be unsuitable for use in removing RIC from MR \(k\)-space data as they introduced signal intensity variations in the images. However, the alteration of our algorithm to use the median value of the pixels in the neighborhood of the anomalous pixel, instead of the mean background value, proved to be a suitable alteration of our algorithm. The results shown in Table V for this scenario were very similar to those obtained with our original algorithm.

It is important to note that while a global convolution of the \(k\)-space with a moving average filter may have an undesirable effect in the image space, as partially shown for the global median filter, we are not applying local averaging to the entire \(k\)-space. Only the selected points contaminated by the radiation pulses are replaced by the local average, and even these have to be far away from the meaningful \(k\)-space values to be selected.

At this point the algorithm relies on the user to manually import the MR data and run the postprocessing algorithm in MATLAB; however, future plans involve full automation of the algorithm and the incorporation into the imaging chain so that the \(k\)-space data are processed as they are acquired, rather than postacquisition. To accomplish this goal, the current algorithm may need to be altered in order to examine points line-by-line as they are acquired, rather than using the algorithm from Sec. II.D, which is applied after all data are acquired.

### V. CONCLUSIONS AND FUTURE WORK

The data presented show that radiation induced current, present when RF coils are irradiated by linear accelerators and visible as signal spikes in the image \(k\)-space data, can negatively affect MR images by lowering the image SNR. The amount of SNR loss increases with increasing linac dose rates. In a typical treatment regime high dose rates, 250 MU/min and above, are routinely used to provide the patient with short treatment times. At these high rates there is significant SNR loss, some of which can be recovered through the use of a postprocessing algorithm which removes the RIC artefact from the image \(k\)-space data.

Our work has established beyond doubt that the RIC exists in RF coils of our integrated linac-MR system. The impact of the RIC on the images may or may not be significant. As is shown in this work, while the SNR loss is quantitatively significant, it is not discernible in the images with the naked eye. This does not mean that the RIC phenomenon should be ignored. The integrated systems will definitely use intensity modulated radiation therapy at very high dose rates – i.e., not only lot more pulses of irradiation per unit time, leading to more RIC noise spikes per image, but also, a much higher dose per pulse, leading to higher intensity RIC noise spikes. At high dose rates, this effect may become important and may require mitigation. As mentioned in Sec. I., the authors do not expect that the SNR loss seen in our system will be problematic for higher field systems, due to the inherently higher SNR. It is also possible that due to the higher frequency and different bandwidths used in high-field systems, the RIC will be inherently filtered from the acquired MR data.

Future work will involve automating the RIC-removal algorithm and possibly its incorporation into the image acquisition chain. While it would be possible to synchronize the linac with the MRI so that no acquisition occurs during beam-on time, this would severely limit the real-time benefits of an integrated linac-MR system and so is not desirable.

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