Improving the imaging performance of the 1.5 T MR-linac using a flexible, 32-channel, on-body receive array

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Abstract
High impedance coils (HICs) are suitable as a building block of receive arrays for MRI-guided radiotherapy (MRigRT) as HICs do not require radiation-attenuating capacitors and dense support materials. Recently, we proved the feasibility of using HICs to create a radiation transparent (i.e. radiolucent) window. In this work, we constructed a fully functional 32-channel array based on this design. The anterior element is flexible and follows the shape of the subject, while the posterior element is rigid to support the subject. Both elements feature a 2×8 channel layout. Here, we discuss the construction process and characterize the array’s radiolucency and imaging performance. The dosimetric impact of the array was quantified by assessing the surface dose increase and attenuation of a single beam. The imaging performance of the prototype was compared to the clinical array in terms of visual appearance, signal-to-noise ratio (SNR), and acceleration performance, both in phantom and in-vivo measurements. Dosimetry measurements showed that on-body placement changed the anterior and posterior surface dose by +3% and −16% of the dose maximum. Attenuation under the anterior support materials and conductors was 0.3% and ≤1.5%, respectively. Phantom and in-vivo imaging with this array demonstrated an improvement of the SNR at the surface and the image quality in general. Simultaneous irradiation did not affect the SNR. G-factors were reduced considerably and clinically used sequences could be accelerated by up to 45%, which would greatly reduce pre-beam imaging times. Finally, the maximally achievable temporal resolution of abdominal 3D cine imaging was improved to 1.1 s, which was >5× faster than could be achieved with the clinical array. This constitutes a big step towards the ability to resolve respiratory motion in 3D. In conclusion, the proposed 32-channel array is compatible with MRigRT and can significantly reduce scan times and/or improve the image quality of all on-line scans.

1. Introduction

Hybrid MRI-radiotherapy systems (Fallone 2014, Keall et al 2014, Lagendijk et al 2014, Mutic and Dempsey 2014) allow for monitoring of mobile tumors and organs at risk (OARs) with magnetic resonance imaging (MRI) before, during, and after radiation therapy treatments.

Based on observed changes, the tumor and OARs can be recontoured and the treatment plan can be adapted to improve tumor coverage and minimize toxicity. High image quality is therefore required to accurately delineate these structures. Furthermore, cine MRI can be used to track the anatomical changes...
that occur during treatments (intrafraction motion). However, as MRI is inherently slow, high-quality preparatory (pre-beam) imaging takes relatively long and real-time anatomy monitoring is limited to 2D planes or small, low-resolution 3D volumes. This limitation may be overcome when receive arrays with a high channel count are used, as these enable the use of high acceleration factors through their increased parallel imaging (PI) capabilities (Breuer et al 2005, Larkman et al 2001, Pruessmann et al 1999).

The 1.5 T Elekta MR-linac (Unity, Elekta AB, Stockholm, Sweden) currently comes with a clinical receive array that consists of two 4-channel elements. These elements are placed centimeters away from the anatomy: the anterior element is elevated above the patient while the posterior element is placed underneath the table (Hoogcarspel et al 2018). The limited channel count, combined with its 1 × 4-channel arrangements and distant positioning, limits the signal-to-noise ratio (SNR) and PI performance in all planes.

Previous work (Zijlema et al 2019) proposed the design of an 32-channel on-body array that is suitable for MRI-guided radiotherapy (MRigRT). High impedance coils (HICs) were used, as these exhibit less coupling with neighboring elements, are flexible, and do not require beam-attenuating lumped elements to tune to the resonance frequency. Instead, HICs use the distributed capacitance of the coaxial conductor (Zhang et al 2018). This way, a radiation transparent, or radiolucent, window was achieved with a double-row layout. It was shown that the materials of the flexible, anterior element did not significantly change the delivered dose and that on-body placement did not significantly increase the surface dose. Finally, an increase of the SNR was shown when comparing the imaging performance of a 5-channel prototype to the clinical array on a phantom.

In this work, we constructed a fully functional 32-channel array for the 1.5 T MR-linac based on the aforementioned design. We discuss the construction process and characterize the array’s radioluency and imaging performance. The dosimetric impact of the array is quantified by assessing the surface dose increase and attenuation of a single beam. The imaging performance of the prototype was compared to the clinical array in terms of visual appearance, signal-to-noise ratio (SNR), and acceleration performance, both in phantom and in-vivo measurements. The latter included $T_2$ turbo spin echo (TSE) sequences that are clinically used for on-line contouring and 3D cine sequences that can be used for motion tracking.

2. Methods

2.1. Construction

Based on the design of Zijlema et al (2019), the anterior and posterior elements of the 32-channel array contain 16 channels each, which are placed in a 2 × 8 channel layout (figure 1). As a reference, the design of the clinical array is schematically shown in figure 1(e). The high-impedance coil loops of the new array (9 × 19 cm$^2$) were constructed from AlphaWire (Elizabeth, NJ, USA) 9432 wire to resonate at 63.87 MHz and were sown onto a 0.5 mm polystyrene sheet (figure 1(c)). All loops were connected to newly designed matching and detuning circuitry (figure 2), which in turn connected to an interfacing box with the preamplification and digitization hardware. Foam layers were added to cover the electronics and create a conductor-to-surface distance of 15 mm to reduce the bolus effect (Zijlema et al 2019, Ghila et al 2016). Coupling between channels, which occurs when multiple channels are placed in close proximity (Roemer et al 1990), was limited due to the use of HICs and was further mitigated by overlapping the loops (figure 1(c)). Coupling of the final layout was assessed on the bench and found to be low (appendix A). Coupling values are compared to those of the clinical array in section 2.3. The use of light-weight materials resulted in an anterior element that weighs 1.3 kg and a posterior element that weighs 1.5 kg. Cables to and from the interfacing boxes are routed such that none cross through the radiation path, i.e. cable output is required on both sides of the bore. A safety assessment of the full array was performed to ensure volunteer safety by identifying potential risks and implementing design mitigations (Rispoli 2019).

2.2. Dosimetry

Two aspects of the dosimetric impact of the 32-channel receive array were evaluated: (1) the surface dose increase (bolus effect) and (2) the single-beam attenuation.

The bolus effect was quantified by comparing the anterior and posterior surface dose with and without the prototype present. The anterior bolus effect was assessed first. A GAFChromatic (Ashland, USA) EBT-3 film (lot: 10 241 901) was placed on top of a solid water phantom in a 1.5 T MR-linac. A 7 MV, 2000 MU, 20 × 10 cm$^2$ beam was delivered from 0$^\circ$ with and without the anterior element of the prototype directly on top. In order to compare the posterior bolus effect, the phantom was placed on top of the posterior element of the prototype. An EBT-3 film was placed between the prototype and the phantom and a 1000 MU, 20 × 10 cm$^2$ beam was delivered from 180$^\circ$. Next, the prototype was removed, the thin Elekta-provided...
Figure 1. (a), (b) Schematic views of the positions of the prototype’s anterior and posterior elements. (c), (d) Photos showing the anterior element without its cover (c) and the full posterior element (d). The electronic layout is the same for both elements. Loops are sown onto a thin plastic sheet and connected to a matching and detuning board. Signal cables are routed through custom cable traps and connected to interface boxes with preamplification and digitization hardware. The use of light-weight materials resulted in an anterior element that weighs 1.3 kg and a posterior element that weighs 1.5 kg, including cabling and cable traps. (e) Schematic view of the clinical array with the anterior element elevated above the patient using a double-arc bridge and the posterior element below the table.
Figure 2. Matching and detuning circuitry. At the port side of the HIC, the shield of both ends is connected. The circuitry is connected to a preamplifier PCB via a coaxial cable. On this PCB, the DC detuning current is superimposed onto the signal lines. Furthermore, a T-network is implemented to achieve optimal preamp decoupling with the 1.5 m long signal cables (Roemer et al 1990).

directly against the panel. Images were acquired using XIS (PerkinElmer, Waltham, MA, USA) with an acquisition time of 433 ms per frame and with all automatic corrections disabled. The beams (6 MV, 100 MU, 10 × 10 cm²) were delivered with a dose rate of 250 MU min⁻¹. 100-frame averages were saved that covered a full beam delivery and 5-minute waiting periods were employed to avoid ghosting effects. Reference images, i.e. beam-off (dark) images and acquisitions without phantom, were acquired and used to process the acquisitions ($I_{\text{proc}}$), as described by McDermott et al (2004). The dose attenuation fraction due to an array ($A_{\text{EPID}}$) can then be calculated with

$$A_{\text{EPID}} = \frac{I_{\text{proc, array}} - I_{\text{proc, no array}}}{I_{\text{proc, no array}}}.$$  

(1)

2.3. Imaging performance

Imaging performance was assessed by measuring the coupling between channels, signal-to-noise ratio (SNR), and parallel imaging (PI) performance. Additionally, the impact of radiation on the SNR is assessed. All imaging was performed on a 1.5 T Elekta Unity MR-linac and compared the performance of the 8-channel clinical array to the 32-channel prototype.

First, raw data were acquired of a pelvis-sized phantom (PVP- and agar-based, 2.6% NaCl) with dielectric properties that are representative of a human body. Noise correlation matrices were generated by calculating the Pearson correlation coefficients from the noise-only pre-scan. SNR maps were generated from 3D acquisitions (table 1) with a noise-only dynamic, as described by Kellman and McVeigh (2005).

Subsequently, the impact of the radiation beam on the SNR was assessed. In previous work, radiation-induced signal spikes in acquired k-space lines have been described when a beam passes through a receive array (Burke et al 2010, Hoogcarspel et al 2018). This effect was assessed by calculating the time-course SNR from two dynamic series (Hoogcarspel et al 2018): one with and one without a large beam (15 × 15 cm², 430 MU min⁻¹) that crossed the array. The experiment was performed with the full 32-channel prototype and the 8-channel clinical array.

Finally, in-vivo imaging of three volunteers was performed comparing the prototype and clinical array. The quality of the clinically used 3D $T_2$-weighted TSE pre-beam acquisitions (table 1) was compared and the sequence was optimized in terms of acquisition speed by taking advantage of the additional channels of the prototype. Furthermore, g-factor maps were generated from 3D in-vivo acquisitions (table 1) to compare the acceleration performance of the two arrays. The g-factor is a spatial measure of the noise amplification that occurs when an unfolded image is reconstructed from undersampled data (Pruessmann et al 1999). Here, a higher g-factor will lead to more SNR loss and increases the chance of unresolved artifacts. Mean g-factors were calculated within the body contour for all volunteers. Lastly, a 3D bSSFP cine sequence was optimized in terms of speed for each array (table 1).

3. Results

3.1. Dosimetry

The placement of the prototype array increased the anterior surface dose by 28% (+3% of $D_{\text{max}}$) with respect to the situation without array present. In contrast, the posterior surface dose was reduced by 41% (~16% of $D_{\text{max}}$) when the posterior element was present.
|
|---|---|---|---|---|
| **Sequence** | **In-vivo pre-beam** | **In-vivo 3D cine** |
| Phantom (SNR) | Clinical sequence | Prototype-optimized | Clinical-array-optimized | Prototype-optimized |
| 3D Spoiled GE | T2W TSE | bSSFP |
| TR / TE | 10.0 / 1.8 ms | 1535 / 277 ms | 3.7 / 1.8 ms |
| FOV | 425 × 250 × 250 mm³ | 425 × 250 × 250 mm³ | 400 × 350 × 193 mm³ |
| Voxel size | 3 × 3 × 10 mm³ | 1.5 × 1.5 × 2 mm³ | 5 × 5 × 5 mm³ |
| Flip angle | 20° | 90° | 40° |
| SENSE [R_FH, R_LR] | No | [1.0, 3.6] | [1.5, 4.0] |
| Partial Fourier [FH, LR] | No | [0.63, 1.0] | [0.63, 0.8] |
| Oversampling | Yes, FH: 1.4× | Yes, FH: 1.5× |
| Acq. time | 0:41 min | 2:09 min | 5.8 s |
| | | 1:18 min / 1:09 min | 1.1 s |

Figure 3 shows the attenuation maps that were obtained from the EPID measurements. The anterior element of the prototype has low-attenuating support materials (0.3%). As expected, the HICs attenuate slightly more: 0.8% under a single conductor and 1.5% under a crossing point. The support materials of the posterior element attenuate 0.6% and a single conductor and conductor crossing increase this value to 1.1% and 1.6%, respectively. In comparison, the anterior element of the clinical array attenuates approximately 0.4% (0.5% under conductor) and the posterior element 1.2% (1.3%).
3.2. Imaging
Coupling between channels during the MRI acquisitions is shown in figure 4. Coupling between the channels, i.e. the off-diagonal values of the matrix, are lower in the prototype ($\leq 0.30$ vs. $\leq 0.47$), although the loop size is smaller.

SNR maps were generated from scans with the 32-channel prototype and 8-channel clinical array (figure 5). These show an increase of the SNR at the surface when the prototype is used, while the SNR at larger depths is similar.

Figure 6 showcases the image quality of a clinically used $T_2$ TSE sequence with and without acceleration using SENSE. The prototype outperforms the clinical array in terms of SNR, which is best visible in the pubic bone. The reduced SNR, due to the acceleration of the acquisition, is especially pronounced when the scan time is reduced by 40%.

Figure 7 attests that the prototype improves the PI performance by reducing g-factors, which in turn reduces SNR loss and minimizes unfolding artifacts. The mean g-factor for several combinations of acceleration factors are shown in table 2.

Figure 8 shows that the improved PI performance of the prototype can be used to acquire highly accelerated 3D cine imaging of the abdominal region with a temporal resolution of 1.1 s. The clinical array required a scan time of 5.8 s to achieve a comparable image quality. Videos of these dynamic acquisitions are available in the supplemenray material (stacks.iop.org/PMB/65/215008/mmedia).

4. Discussion
In this work, we built a light-weight 32-channel HIC receive array based on the design of Zijlema et al (2019). We showed that the full 32-channel on-body array is dosimetrically feasible for use in a 1.5 T MR-linac and significantly improves the image quality and speed compared to the current clinical array.

On-body placement of the array resulted in a surface dose increase of 3% of $D_{\text{max}}$. This is even lower than was found by Zijlema et al (2019) and deemed acceptable. The difference can be explained by the new
support materials and the higher delivered dose, which improves the method’s accuracy. The surface dose on the posterior side actually reduced by 16% with the posterior element in place due to its significantly lower (electron) density compared to the treatment couch. The single-beam attenuation values due to the
Table 2. Mean and standard deviation of g-factors for several combinations of acceleration factors in feet-head ($R_{FH}$) and left-right ($R_{LR}$) direction. All data of three volunteers is combined. Mean g-factors above 2.0 are considered high and are marked in bold.

|       | $R_{FH}$ = 1 | 1.5  | 2    |       | $R_{FH}$ = 1 | 1.5  | 2    |
|-------|--------------|------|------|-------|--------------|------|------|
| $R_{LR}$ = 1 | 1.0±0.0      | 1.0±0.1 | 1.2±0.2 | 1.0±0.0      | 2.1±1.7 | 4.7±2.6 |
| 2     | 1.0±0.0      | 1.1±0.1 | 1.2±0.2 | 1.0±0.0      | 2.3±2  | 5.5±3.6 |
| 3     | 1.1±0.1      | 1.1±0.1 | 1.3±0.2 | 1.1±0.2      | 2.9±2.9 | 7.8±4.9 |
| 4     | 1.3±0.2      | 1.4±0.2 | 1.6±0.3 | 1.4±0.2      | 5.6±9.2 | 19.3±20.2 |
| 4.5   | 1.6±0.3      | 1.6±0.3 | 1.9±0.4 | 1.9±0.9      | 9.4±17.5 | 32.2±34.4 |
| 5     | 1.8±0.4      | 1.9±0.4 | 2.1±0.5 | 2.6±1.9      | 11.7±21 | 34.9±35.8 |

Figure 8. Three views (coronal, sagittal, transverse) of the 3D cine imaging volumes using the prototype (top) and clinical array (bottom). The prototype produces 3D images without noticeable artifacts, while the 1.4 s acquisition with the clinical array is unusable. The third acquisition shows severe motion artifacts, e.g. in the liver dome, due to the long acquisition time with respect to the respiratory cycle. Animations of these three dynamic series can be found in the supplementary information. Note that the last acquisition was acquired in a different volunteer.

Prototype ($\leq 1.5\%$) were in good agreement with the acceptably low values that were reported by Zijlema et al (2019). The maximum attenuation values (1.5%) were found directly under two overlapping conductors. Clinical impact of these slight, local underdosages is deemed negligible, as changes from one beam will be smeared out due to physiological motion and the use of multi-angle beam or volumetric arc treatments. Therefore, dose changes are expected to be well below 1%. These results indicate that the anterior element of the 32-channel array can be disregarded in the treatment planning process. The posterior element has a fixed position on the bed and can easily be included in the treatment planning process.
The use of HICs was a key factor to allow for dense placement of 32 channels without dosimetric impact, due to its lack of electronic components in the radiation window. Moreover, dense placement was possible due to the inherently limited coupling between HICs (Zhang et al. 2018). Replacing the single-row with a double-row channel layout provides an additional dimension in coil sensitivity variation (the feet-head direction) and thereby enhances the PI capabilities. Further increasing the number of rows in this direction would again improve the PI performance, but the placement of additional cabling and circuitry will be hindered by the requirement of a radiolucent window during MRgRT. Two rows therefore seem to be the maximally achievable.

The imaging performance of the prototype was compared to the current clinical array in several ways. First, its coupling between channels was found to be lower, although the channel size was smaller and each channel is neighbor to more channels due to the double-row layout. As discussed before, the use of HICs enabled dense channel placement with minimal coupling. Furthermore, the SNR of the prototype was found to be higher at the surface and similar at depth. The radiation beam did not seem to impact these findings. Beam-on imaging showed that the radiation did not change the tSNR, as was found by Zijlema et al. (2019). Previous work by Hoogcarspel et al. (2018) did report a slight change, which could be explained by the larger field size (22 × 22 cm² vs. 15 × 15 cm²) and higher default dose rate (680 MU min⁻¹ vs. 430 MU min⁻¹) in comparison to this work.

In-vivo acquisitions of volunteers showed that the g-factors were substantially lower when the prototype was used. Consequently, an approximate two-fold scan time reduction of a clinically used T₂ TSE sequence could be achieved. Current MR-linac protocols include three T₂ TSE scans of 2 or 4 minutes, i.e. total scan times can likely be reduced by up to 6 minutes. The improved acceleration performance may also be used for further optimization of the image quality, e.g. in terms of resolution or SNR. Finally, the temporal resolution of 3D cine imaging has been shown to improve more than five times, from 5.8 s to 1.1 s. Although the latter is still not able to completely avoid intradynamic motion, this constitutes a big step towards the ability to resolve respiratory motion in 3D. Further acceleration should be sought in the acquisition and reconstruction methods, e.g. using deep learning (Mehta and Majumdar 2017, Terpstra et al. 2020). Although the acquisition voxel size of 5 × 5 × 5 mm³ is relatively coarse, Glitzner et al. (2015) have shown that subvoxel motion can be detected at up to one third of the voxel size.

The 32-channel array could also improve diffusion weighted imaging (DWI) for treatment response monitoring on the MR-linac. In DWI sequences, echo-planar imaging (EPI) readouts are mostly used, which are prone to geometric distortions due to their long readout trains over which phase errors accumulate (Mansfield 1977). These distortions are clearly undesired in a radiotherapy (treatment) setting. The ability to use higher acceleration factors with the 32-channel array will reduce the length of the echo trains, which results in fewer geometric distortions. Additionally, shorter echo trains will reduce intrashot T₂ decay and thus signal blurring.

Now that a radiolucent HIC array has been shown to be feasible, the design concept could be extended to other sites that could benefit from the flexible properties, such as the head and neck. Furthermore, the application of the current array in hybrid PET/MRI systems could be investigated. In PET/MRI, attenuation correction of on-body receive arrays is similarly difficult and significant changes of the measured abdominal activity values can occur (Fürst et al. 2014, Wollenweber et al. 2014). The radiolucent properties of our proposed design could reduce the number of photons that are attenuated or scattered and thus improve the accuracy and SNR.

5. Conclusion

The presented receive array improves the imaging performance with respect to the current clinical array. Its 32 channels enable the use of high undersampling factors, which reduced pre-beam and 3D cine scan times by about half and 80%, respectively. All this can be achieved without inducing clinically relevant dose changes.

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The coupling between channels of the prototype was measured on the bench using an in-house developed automated VNA RF switch (Welting et al. 2020) and resulted in a $32 \times 32$ S-matrix. Bending of the loops did not significantly affect the tuning, matching and coupling of the array. The bench-measured coupling is displayed in figure A1.

**Appendix A. Coupling bench test**

The coupling between channels of the prototype was measured on the bench using an in-house developed automated VNA RF switch (Welting et al. 2020) and resulted in a $32 \times 32$ S-matrix. Bending of the loops did not significantly affect the tuning, matching and coupling of the array. The bench-measured coupling is displayed in figure A1.

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