Radiofrequency Coil Innovation in Cardiovascular MRI

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Introduction

Over the last decade advances in the design of cardiac MRI pulse sequences have allowed the assessment of ventricular function, first-pass perfusion and myocardial viability to be performed with increasing accuracy and speed [1, 2]. Much of this progress has been made possible by significant advances in gradient coil technology, without which the implementation of sequences such as balanced steady state free precession (SSFP, e.g. TrueFISP) would not have been possible. However, despite these important advances, cardiovascular MRI has yet to reach its full potential and additional transformative technological innovations will be needed to improve the spatial resolution, efficiency and speed of cardiac MRI. In this article we describe recent and ongoing innovations in radiofrequency coil design for cardiovascular MRI. After a brief review of existing 32-channel receive coils [3–5], we focus most of the article on a recently developed 128-channel cardiac receive coil and the potential impact of multi-channel transmission coils in cardiovascular MRI [6].

Multi-channel receive coils. Advantages compared to single-channel coils

The sensitivity of multi-element parallel array coils matches the sensitivity of coils with fewer but larger elements in the center of the body, but provides substantial gains in sensitivity closer to the surface of the coil [7–11]. More importantly, multi-element arrays play a crucial role in facilitating parallel imaging with high acceleration factors (R) [3–5]. The signal-to-noise ratio (SNR) during parallel acquisition, SNR_{PA}, is influenced by both the degree of undersampling, described by the acceleration factor (R), and also by the G-factor (G), which is the local geometry factor. SNR_{PA} is thus described by the following equation below.

$$SNR_{PA} = \frac{SNR_{full}}{\sqrt{R} \times G}$$

Simulations and experimental data have shown that G can be reduced, for a given acceleration factor and coil geometry, by increasing the field strength B_o and/or by increasing the number of coil elements in a phased array coil [12, 13]. The use of multi-channel phased array coils with a high number of individual elements thus not only produces higher SNR values near the coil plane, but also decreases the SNR penalty associated with parallel imaging. Cardiac arrays with 32 coil elements are now commercially available at both 1.5 and 3 Tesla, and have been shown to dramatically impact the ability to accelerate 2D images in one dimension. Highly accelerated 2D cine images have been obtained with these coils, principally using the tSENSE algorithm [14], allowing large volumes of the heart to be imaged in a single breath-hold. The experience with 3D imaging with these coils is less extensive, and is discussed further below. However, it is clear that these 32-element coils constitute a major advance and will have a significant impact on cine, first-pass perfusion and potentially other areas of cardiovascular MRI as well.

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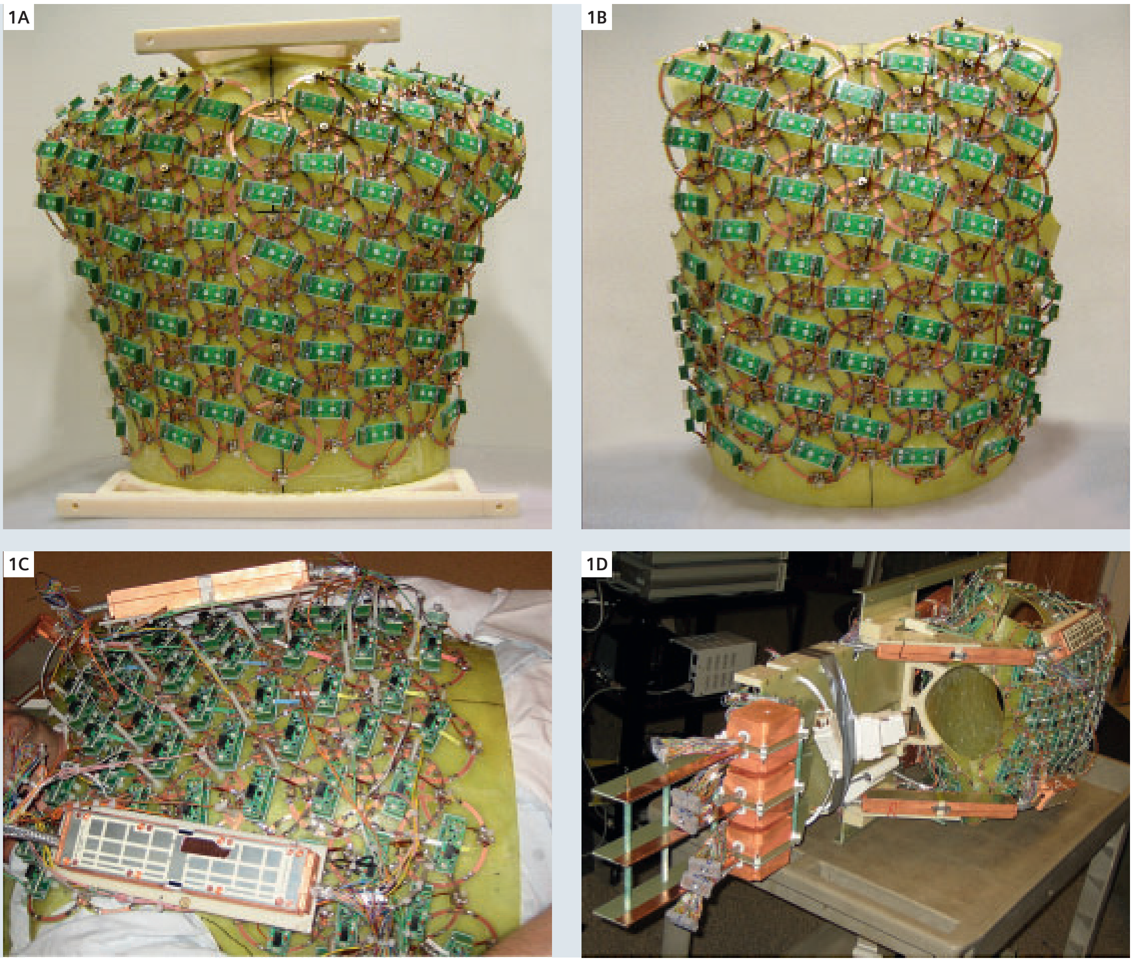
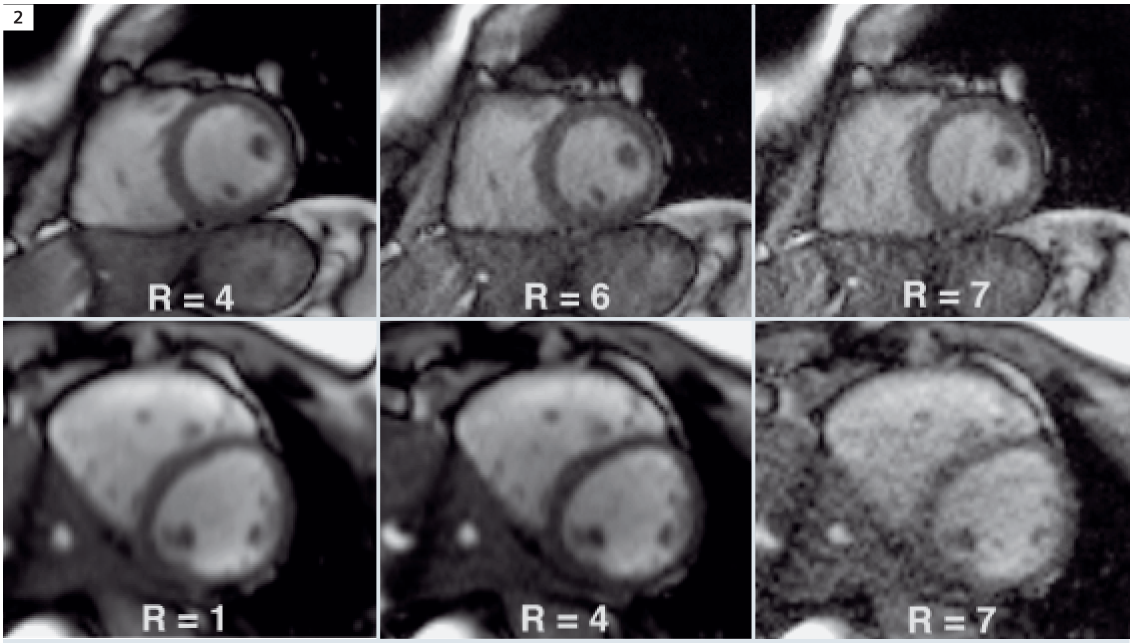
New development: 128-channel receive coil for cardiac MRI at 3 Tesla

The appeal of performing parallel acquisition at higher field strengths, such as 3 Tesla, lies not only in the higher unaccelerated SNR obtainable, but

also in the ability to use a larger number of smaller elements without a significant rise in the G-factor. We have thus developed a 128-channel coil for use on a prototype 3T scanner, expanded to 128 independent channels (Figure 1). Our aims in developing this system were to explore the feasibility, challenges and potential benefits of ultra-high element (> 32 elements) receive arrays and determine their potential impact on cardiovascular MRI. The MR system used with this coil is based on a clinical Siemens MAGNETOM Trio, A Tim System 3T scanner with 32 individual RF channels, which was expanded to a system with 128 individual receiving channels [15]. This allows the signal of each individual coil element to be detected with a single receive RF channel. The 128 individual coil elements (mean diameter d = 75 mm) are arranged on a fiberglass shell with 68 elements on the posterior portion and 60 coil elements on the anterior part of the coil (Figure 1).
SNR and G-factor maps were obtained with the 128-channel coil on torso-shaped oil and water phantoms, and compared to those acquired with a commercially available 32-channel coil (InVivo Corporation, Gainesville, FL, USA). At higher acceleration factors (R = 5, 6, 7 and 8) maximum G-factors with the 128-channel coil were approximately 50% those with the 32-channel coil. The gains in unaccelerated SNR with the 128-channel coil were more modest, but in human volunteers a gain in unaccelerated SNR at the apex of up to 1.3 could be obtained with the 128-channel versus 32-channel coil. The low G-factors obtained with the 128-channel coil (maximum G-factor of 1.7 at R = 7 in left-right direction) facilitated the acquisition of high quality accelerated images in several healthy human volunteers. 2D Cine images of the heart were obtained with a SSFP (TrueFISP) sequence using the following imaging parameters: field of view (FoV) 360 mm, TE / α / BW = 1.4 ms / 35° / 965 Hz / Pixel, matrix size: 101 x 192, producing an in-plane spatial resolution of 1.9 mm x 2.9 mm. Slice thickness



1 Prototype 128-channel cardiac coil.
(A, B) The coil consists of a fiberglass mold in a clam-shell geometry, with 60 elements on the anterior portion and 68 elements on the posterior portion of the mold. Each element is 75 mm in diameter and arranged in overlapping hexagonal symmetry to reduce next neighbor coupling. The preamplifier of each element lies 3 cm above the element to improve coil compactness and reduce cable related signal loss.
(C, D) Fully assembled coil with its cover and padding removed. The signal from 32 of the channels is routed to the regular receiving unit of the scanner, while the signal from the other 96 elements is routed to 3 additional receiving units.



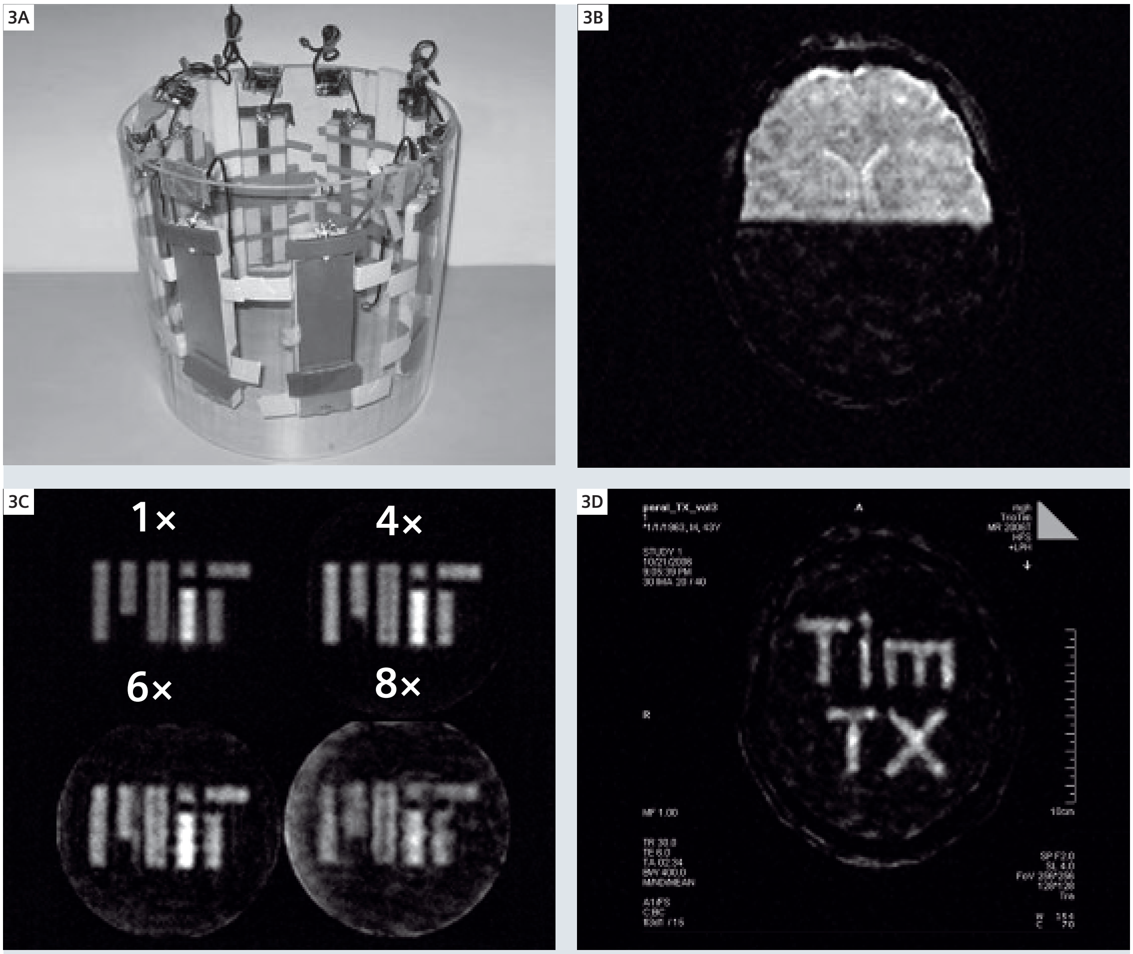
2 Highly accelerated 2D SSFP (TrueFISP) images in a healthy male volunteer (top panel) and a healthy female volunteer (lower panel). The GRAPPA algorithm was used in one direction with an acceleration rate of up to 7. In plane resolution was 1.9 x 2.9 mm and slice thickness ranged from 6–10 mm. Because of the need to acquire reference lines with GRAPPA, even at acceleration rates of 6 and 7, a maximum of only ten slices could be obtained in a single breath-hold. However, the acquisition of ten 10 mm thick slices, one of which is shown in the male volunteer in the top panel, allowed the whole heart to be easily covered in a single breath-hold. The female volunteer in the lower panel was imaged with a slice thickness of 6 mm and, in addition, the anterior portion of the coil did not lie close to her chest wall, since a rigid design (fitted a male chest) was chosen for this prototype coil to simplify experimental issues. Despite this, the rate 4 images are almost indiscernible from the unaccelerated images, and the rate 7 images remain diagnostic with well-preserved blood-tissue contrast. Future versions of the coil will likely include a flexible anterior portion to lie close the chest of all subjects scanned.

ranged form 6–10 mm and temporal resolution was set to16 frames per RR interval (Figure 2). Parallel acquisition was performed with the GRAPPA algorithm with acceleration factors up to 7 in one dimension. As shown in Figure 2, image quality was well preserved at acceleration rates as high as R = 6 and 7, in both male and female volunteers. The use of spatio-temporal compression algorithms such as tSENSE with these acceleration factors has the potential to allow high quality 2D cines to be acquired sequentially over the entire heart in a single relatively short breath-hold. In addition, 2D acceleration during 3D imaging will likely demonstrate the value of the low G-factors obtained with the 128-channel coil even more dramatically. The current results, obtained with accelerated imaging in one direction, however provide strong proof-of-principle for the use of this 128-channel coil, and suggest that highly accelerated volumetric 3D imaging with this coil will be highly promising.

Multi-channel transmit coils

The concept of multiple independent RF channels is now being applied to RF transmission as well as to RF reception. The potential of parallel RF transmission lies in its ability to allow RF (B_1) shimming as well as spatially tailored excitation to be performed. The challenge of uniform RF (B_1) excitation in the heart becomes significantly more problematic at field strengths of 3 Tesla and higher. Adiabatic excitation pulses can be used to address this challenge, as well shown by Nezafat, Stuber and Pettigrew, but are often long, limited by SAR thresholds and more difficult to implement in a slice selective manner. Vaughan and colleagues have recently shown that RF shimming with multiple RF transmission channels in human brains at 9.4T can be used as an alternative method to address this B1 inhomogeneity challenge.
Perhaps the greatest potential impact of multiple transmission coils, however, lies in their ability

Highly accelerated volumetric 3D imaging with the 128-channel coil will be highly promising.



3 Spatially tailored RF excitation with an 8-channel RF transmission coil. Several different coil configurations have been tested. An example of an 8-channel stripline transmit array is shown in **(A)**. **(B)** Spatially tailored excitation of the human brain at 3T with an 8-channel transmit array. **(C)** A spatially selective 2D RF excitation pulse has been used to excite the "MIT" logo in an oil phantom. Without parallel RF transmission, however, the duration of this excitation pulse approaches 10 ms. At rate 4 acceleration the pulse duration can be reduced to approximately 2.5 ms and at rate 6 to less than 1.7 ms. It should be noted that the duration of spatially tailored 3D excitation pulses can reach over 50 ms and the use of parallel RF transmission techniques in this setting may be even more important than in the 2D setting. The reduction of pulse duration with parallel RF transmission also has the benefit of reducing off-resonance effects. At present image quality at rate 7 and 8 acceleration becomes problematic but should improve with the design of better multi-channel RF transmission coils and more efficient algorithms to undersample excitation k-space. Even with present technology, however, complex patterns of spatially tailored excitation can be produced in vivo, as shown in the 3 Tesla image of a normal volunteer **(D)**.

to accelerate spatially tailored excitation pulses, so that the duration of these pulses becomes feasible for clinical implementation. The ability of parallel transmission to facilitate spatially tailored excitation in both two and three dimensions has recently been demonstrated both in phantoms and in the brain in vivo [6]. The experience with spatially tailored excitation in the body is less extensive but could be used, in theory, for the excitation of only a well-defined 2D or 3D region in the body, such as the heart. The selective excitation of only the heart could then facilitate the reduction of the imaging field of view (FoV), without the penalty of infolding artifacts, and thus facilitate a dramatic increase in either the spatial resolution or speed of a given acquisition.

Prototype 8-channel parallel transmission systems have now been installed on 3, 7 and 9.4 Tesla human scanners. The 3 Tesla prototype system is based on a clinical MAGNETOM Trio, A Tim System scanner using a 8-channel transmission head coil. The 8-channel transmit array is connected to the scanner's original RF channel and operated in a "master-slave" configuration, where the Trio is the master. Cloning the standard Siemens measurement control unit (MPCU) for all 7 slaves makes it possible to run 8 independent sequences with individual RF shapes and characteristics. Using the 8-channel transmission head coil in combination with a 2D spatially tailored excitation RF pulse, the use of acceleration factors as high as R = 8 allowed the RF pulse duration to be shortened from 9.47 ms (R = 1) to 2.42 ms (R = 4), 1.64 ms (R = 6) and 1.26 ms (R = 8). The image quality in the accelerated images was well preserved even with transmission acceleration factors as high as R = 6, as shown in Figure 3. In vivo imaging of the brain could also be performed with this coil and resulted in highly spatially tailored excitation of the human brain (Figure 3).

Conclusion

The breadth, flexibility and soft tissue contrast obtainable with cardiovascular MRI remain unsurpassed. However, further improvements in the speed, efficiency and spatial resolution of cardiac MRI are needed for it to reach its full clinical potential. We have described in this article recent and ongoing technical innovations in radiofrequency design, that have the potential to play a transformative role in cardiovascular MRI. Ultra-high element receive arrays, such as the 128-channel 3T coil described above, could allow high resolution volumetric 3D datasets to be acquired in a single breath-hold. Multi-channel transmission arrays could also dramatically increase both the speed and spatial resolution of cardiac MR acquisitions. While 128-channel receive systems and multi-channel transmission systems remain experimental, both configurations have already been used successfully to scan human volunteers, and 32-channel cardiac receive coils are already commercially available for both 1.5 and 3 Tesla scanners. The translation of ongoing technical innovation in radiofrequency design into routine clinical use thus seems not only highly feasible, but also cause for significant optimism and excitement regarding the future of cardiovascular MRI.

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