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FULL PAPER

Commissioning of a new wide-bore MRI scanner for radiotherapy planning of head and neck cancer

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Objective: A combination of CT and MRI is recommended for radiotherapy planning of head and neck cancers, and optimal spatial co-registration is achieved by imaging in the treatment position using the necessary immobilisation devices on both occasions, something which requires wide-bore scanners. Quality assurance experiments were carried out to commission a newly installed 1.5-T wide-bore MRI scanner and a dedicated, flexible six-channel phased array head and neck coil.

Methods: Signal-to-noise ratio (SNR) and spatial signal uniformity were quantified using a homogeneous aqueous phantom, and geometric distortion was quantified using a phantom with water-filled fiducials in a grid pattern. Volunteer scans were also used to determine the *in vivo* image quality. Clinically relevant T_1 weighted and T_2 weighted fat-suppressed sequences were assessed in multiple scan planes (both sequences fast spin echo

based). The performance of two online signal uniformity correction schemes, one utilising low-resolution reference scans and the other not utilising low-resolution reference scans, was compared.

Results: Geometric distortions, for a ± 35 -kHz bandwidth, were <1 mm for locations within 10 cm of the isocentre rising to 1.8 mm at 18 cm away. SNR was above 50, and uniformity in the axial plane was 71% and 95% before and after uniformity correction, respectively.

Conclusion: The combined performance of the wide-bore scanner and the dedicated coil was adjudged adequate, although superior-inferior spatial coverage was slightly limited in the lower neck.

Advances in knowledge: These results will be of interest to the increasing number of oncology centres that are seeking to incorporate MRI into planning practice using dedicated equipment.

Developments in the field of radiotherapy, such as intensity-modulated radiotherapy and arc therapy, have provided the means to deliver highly conformal treatments, offering the promise of both improved sparing of normal tissue and escalation of tumour dose. To achieve full potential of these techniques, there is an increasing need for anatomical and functional imaging to be used at the planning stage to localise disease with a high level of accuracy. MRI has a number of advantages over CT in the head and neck region, including robustness to dental artefacts and improved soft-tissue contrast, either with or without the use of contrast agents [1,2]. The improved visualisation and delineation of organs at risk and tumours has been shown to be of value in the planning of nasopharyngeal [3], sinonasal [4] and oropharyngeal [5] cancers. For tumour sites lower down, the case is less conclusive and is further complicated by the difficulty in fusing MRI with CT in this region [6]. Although the case for MR-only planning has been made for some anatomical sites [7], MRI planning scans still require

fusion to CT to facilitate electron density-based dose calculations. Ideally, these images should be acquired in the identical treatment position as in the case of the planning CT so as to facilitate accurate registration. This involves immobilisation of the patient with a specially made shell, which reduces interfraction and intrafraction displacements and also reduces deformations [8]. However, the desire to achieve imaging in the treatment position with MRI has often meant a compromise in the image quality, as immobilisation devices and flat tabletops are incompatible with standard diagnostic radiofrequency (RF) coils and scanners. Previous attempts have used diagnostic surface coils for brain planning [9], although image registration and image quality are usually less problematic at this site. Over the last couple of years, MRI vendors have begun to cater to the requirements of the radiotherapy community, and a new generation of wide-bore systems with dedicated RF coils can permit imaging in the treatment position for notoriously difficult sites, such as the head and neck. This report describes

a wide-bore scanner and a new radiotherapy planning-specific RF coil. This coil arrangement can be expected to produce a compromise in the image quality, and the reduced spatial uniformity of signal intensity requires some form of intensity correction to be employed.

The geometric integrity (spatial accuracy) of MRI is poor when compared with CT, and this has been the focus of interest in the radiotherapy community. Distortions in MRI arise from a departure in the linear spatial relationship of the resonant frequency (*i.e.* effective magnetic field gradient) used to encode the images. These effects can be subdivided into system-related ones and those that are induced by the patient. Inherent system effects are caused by the non-linearity of the imaging gradients themselves and the non-uniformity of the static magnetic field, both of which behave less ideally further from the isocentre of the scanner. It has been shown that such effects can be minimised by operating at a high RF receiver bandwidth (BW), as the magnitude of distortions is inversely proportional to gradient field strength [10]. Although this rarely poses a problem in a diagnostic setting, the positional uncertainty needs to be properly understood for planning purposes. Previous studies on low-field open magnets have demonstrated large magnitudes of distortions of up to 23 mm. Images acquired on such systems can be successfully corrected by utilising measurements made in a test object containing a grid of MR-visible points [11,12]. In scanners with horizontal tunnel-bore designs, these effects are much less pronounced, although they are still an important consideration when dealing with the demand for high accuracies in radiotherapy planning. The new generation of wide-bore systems, described as having a bore diameter of at least 70 cm and which are also shorter in length, will inevitably have a worse geometric integrity.

At the field strengths used in routine clinical practice, *i.e.* 1.5 T and above, the patient-induced effects become a considerable feature and simple phantoms are no longer adequate for measuring total distortion [13]. Chemical shift (*i.e.* resonant frequency) differences between fat and water protons can also create a significant pixel shift in the final image in human tissue. Susceptibility changes introduced by the presence of the patient in the field further distort the linear spatial-frequency relationship and have been shown to produce effects of varying magnitude [14].

In most cases, a judicious selection of a high receiver BW, thereby increasing the gradient strength, and reduced field of view (FOV), thereby limiting the volume of the scanner being utilised, can render the distortion effects to within acceptable limits [15]. However, these factors are at the expense of worsening signal-to-noise ratio (SNR) and a smaller region of anatomy for MR-CT fusion, respectively. It is, therefore, always important to characterise and understand the limitations of each specific imaging system [16].

This article describes part of a wider quality assurance (QA) programme that was implemented following the installation of a new wide-bore system to commission its use in the radiotherapy planning of head and neck cancers. The aims of this study were three-fold: first, to characterise the system-related distortion of the new wide-bore scanner using a large FOV

phantom; second, to assess the image quality in the preferred treatment set-up obtained using the dedicated coil in relation to other non-compatible but relevant imaging coils both in phantoms and *in vivo*; and, finally, to demonstrate an end-to-end test of the optimum imaging protocol in a realistic head and neck phantom. The work will be of interest to the increasing number of oncology centres that are looking to implement their own MR planning service.

MATERIALS AND METHODS/VOLUNTEERS

All imaging was acquired on a 1.5-T wide-bore Optima™ scanner (GE Healthcare, Milwaukee, WI), which was installed in 2010. This system offers a number of advantages including a wider diameter patient bore (70 cm) and an optical RF system, which improves the SNR. As part of the system, a bespoke six-channel RF oncology coil was purchased with the intention of using this for head and neck planning. The coil consists of a pair of semi-flexible surface coils for lateral positioning around an immobilisation device. However, this coil was not made compatible with the Optima RF platform until 2011, thus delaying the completion of the work described here.

To provide comparative data, three other RF coils were used in this study. The first, the integrated body coil, is enclosed within the plastic covers of the scanner. This coil also permits imaging in the treatment position but at the expense of reduced image quality owing to the coil's large sensitive volume (*i.e.* a markedly reduced filling factor). A flat tabletop was designed and manufactured in-house to accommodate the thermoplastic shell and frame system used in our centre for imaging with the body and oncology coils. The second coil, a head-neck-spine (HNS) coil array, was the currently preferred diagnostic coil for this anatomy and consists of a six-channel enclosed birdcage design head part attached to a lower neck element that rests on the patient's chest. The final coil examined in the study was the linear head coil that was designed for brain imaging and is a commonly used coil for performing QA tests and benchmarking scanner performance generally.

System-related distortion was evaluated using a large-diameter phantom (Philips Healthcare, Eindhoven, the Netherlands), which consisted of a test object with a diameter of 40 cm that included a section containing a disc with a series of small circular holes arranged in a grid with separation of 5.0 cm. The phantom was filled with a copper sulphate solution, so that the holes showed up on MR images as hyperintense points, which could be used to assess geometric accuracy over a large FOV. In total, 45 points are available for providing measurements of known distances of up to 180 mm radially from the centre of the image (*i.e.* 360 mm in diameter). The phantom was imaged first with its axis parallel to the bore of the scanner (and also, therefore, to the direction of the static magnetic field) and second with the object placed in a perpendicular orientation, supported on the flat bed on both occasions. The phantom was aligned accurately using the scanner's alignment lasers and a spirit level. In each case, the phantom was imaged with the body coil using a spin echo sequence [echo time (TE)/repetition time (TR)=30/1000 ms with 256×256 matrix and BWs of ± 32.5 kHz and ± 9.0 kHz, 3-mm slice thickness] after first allowing for a period of 10 min to enable the fluid to settle.

Figure 1. (a) An axial image of the phantom used to measure geometric distortion, with the isocentre of the scanner indicated by the arrow. (b) A map of the geometric error measured as a function of the grid position within the xy (axial) plane for the phase/frequency direction setting showing the greatest distortion. (c) The corresponding map acquired with the phantom repositioned for the xz (coronal) plane and a receiver bandwidth (BW) of ± 32.5 kHz. (d) As for (c), but with a ± 9 -kHz receiver BW. The right, left, anterior, posterior, superior and inferior anatomical directions are indicated by their respective initial letters.

Images were acquired a second time with the phase and frequency directions swapped. To obtain known positions within the phantom, a CT of the phantom was first acquired using a Philips Brilliance™ wide-bore CT scanner (Philips Healthcare, Guildford, UK). The position of each MR-visible point was then measured on MRI using a semi-automated software programme developed in-house using MATLAB® (Mathworks, Natick, MA). This involved the user selecting a point roughly coincident with each hole, and then a histogram-based algorithm was used to find the perimeter of the hole and the centre of mass was calculated from the region enclosed by the perimeter. The deviation from the true position, as provided by CT, was then calculated.

To compare image uniformity before and after the use of the two intensity correction strategies available on the scanner, a uniform, aqueous head sphere phantom of 20 cm diameter was imaged. In each case, single scans at the isocentre were acquired in each orthogonal plane using the same sequence and a 5-mm slice thickness. In the case of the two multielement coils, images were acquired both with and without the application of uniformity correction. The first technique used, called surface coil intensity correction (SCIC), utilises the pixel statistics within the image and requires no additional scan. The second method, called phased array uniformity enhancement (PURE), requires low spatial resolution and wide FOV proton density-weighted

calibration scans to be acquired with both the body coil and the surface coil prior to the rest of the imaging. These scans are then subsequently used to map out the RF magnetic field inhomogeneities (*i.e.* the B_1 sensitivity) by comparing the signal intensity of the surface coil with that of the relatively homogeneously profiled body coil. Both methods can be performed either retrospectively, to allow comparison of pre- and post-correction images, or prospectively to save disc space. Uniformity measurements in the head and body coils were also performed.

In all cases, the average absolute deviation (AAD) and fractional uniformity (U) were computed according to Equations (1) and (2), respectively, where S is the mean signal intensity from a region of interest (ROI) encompassing 85% of the total signal area and S_i is the intensity of the i th pixel for a total of n pixels [17]. In addition, uniformity maps were generated by comparing, on a pixel-by-pixel basis, S_i with S with respect to three tolerances. Maps were displayed using a three-colour scale to indicate whether pixels were within $\pm 10\%$, $\pm 20\%$ or $> 20\%$ of S .

$$\text{AAD} = \frac{1}{n} \sum_{i=1}^n (|S_i - S|) \quad (1)$$

Table 1. Signal-to-noise ratios (dimensionless) as measured in a homogeneous aqueous phantom and in the three orthogonal imaging/anatomical planes shown for the head, body and oncology (head and neck) coils

Radiofrequency coil	Axial	Coronal	Sagittal
Head	96	101	98
Body	24	24	24
Oncology	88	65	51

$$U = 1 - \frac{AAD}{S} \quad (2)$$

In the case of the body, head and oncology coils, the spherical phantom was additionally placed in a cylindrical loading annulus to measure SNR (the annulus ensuring coil loading similar to the *in vivo* conditions). For this analysis, two repeat images were acquired with the same pre-scan settings, and a subtraction of one image from the other made to produce a noise-only image. SNR was then calculated using Equation (3), where S is the mean signal in one of the images as described above, SD is the standard deviation of signal measured from the same ROI location in the subtracted image and where the $\sqrt{2}$ factor accounts for the increase in noise power owing to the effective addition of two nominally identical noise patterns.

$$SNR = \frac{S}{(SD/\sqrt{2})} \quad (3)$$

To provide a comparison of the *in vivo* image quality between three clinically relevant RF coil arrangements, a healthy female volunteer was imaged after having obtained informed consent. All volunteer scans were obtained under the hospital's local rules, which had been written after a prior risk assessment had been carried out by the hospital's MRI safety advisor. The local rules prohibited the operation of the scanner outside of the United States Food and Drug Administration/European Economic Area CE marking (FDA/CE) regulatory safety limits, *i.e.* scanning was not performed in research mode or with unapproved equipment. A bespoke thermoplastic mask (Type-S™ system; Civco Medical Solutions, Kalona, IA) was made for the volunteer, which consisted of a 3.2-mm-thick aquaplast material and a combined locking frame. This was used together with a standard headrest (Silverman, size B; Oncology Systems Ltd, Shrewsbury, UK) and a flat tabletop for the study of the body and oncology coils. For the HNS coil, the volunteer was aligned as flat and as straight as the coil would permit. In each case,

Table 2. Percentage uniformity values as measured in a homogeneous aqueous phantom and in the three orthogonal imaging/anatomical planes shown for the head and body coils and corresponding data for the oncology (head and neck) coil without the use of signal uniformity (B_1) correction

Radiofrequency coil	Axial	Coronal	Sagittal
Head	0.97	0.96	0.95
Body	0.96	0.94	0.94
Oncology	0.71	0.73	0.83

Table 3. Percentage uniformity values as measured in a homogeneous aqueous phantom and in the axial imaging/anatomical plane for the head-neck-spine and oncology (head and neck) coils both before uniformity correction and after two types of uniformity correction [where phased array uniformity enhancement (PURE) is reference image-based and surface coil intensity correction (SCIC) is not]

Radiofrequency coil	Before correction	After PURE	After SCIC
Head-neck-spine	0.90	0.96	0.96
Oncology	0.71	0.91	0.95

the same clinical imaging sequence was used: axial T_1 weighted fast spin echo (FSE) with TE/TR=9/400 ms, 3/3-mm slice thickness/gap, echo train length=4, 384×256 matrix and a 25-cm FOV, BW=±27 kHz and a total of 45 slices in 6 min 17 s with two signal averages. SNR was measured as the mean signal in the same tissue type at the centre of the imaging volume divided by the standard deviation of the noise measured in the air background. Additionally, a fat-only and water-only chemical shift imaging sequence was run to qualitatively assess the fat suppression in the oncology and HNS coils: axial T_2 weighted FSE IDEAL (iterative decomposition of water and fat with echo asymmetry and least-squares estimation) TE/TR=100/8200 ms, echo train length=16, 3/3-mm slice thickness/gap, 320×192 matrix and 25-cm FOV with two signal averages, BW=±83 kHz and a total of 45 slices in 4 min 51 s. This scan used a sensitivity-encoding acceleration (ARC) factor of 2.

Finally, a realistic head and neck phantom was used to examine the effects of various protocols on the CT-MR fusion process. This phantom was a slight modification of the one described in a previous study [15], which was carried out to demonstrate the accuracy of our chosen fusion procedure. Briefly, the phantom consisted of several compartments that could be filled with tissue-mimicking solutions and that represent head, neck and spine regions plus an additional compartment open to air, which represents the mouth/trachea. The neck compartment additionally contained an annulus that was filled with oil to simulate fat. The remaining compartments were filled with tap water, apart from the spine, which was filled with a diluted solution of CT contrast agent (Omnipaque™; GE Healthcare) and nickel chloride to provide a different signal intensity. The phantom was further modified to incorporate small plastic bricks to be fixed at various positions to help assess image co-registration.

The phantom was first imaged on the CT scanner to provide the gold-standard data set and then scanned with MRI using a sequence based on the BW and RF coil set-up that was deemed to provide the best results from the earlier tests: axial T_2 weighted FSE, with TE/TR=102/15 000 ms, 3/0-mm slice thickness/gap, 480×256 matrix, BW ±32.5 kHz and a total of 60 slices in 4 min 15 s with one signal average. Imaging was performed using the oncology coil with intensity correction. A separate scan covering the whole phantom was acquired in the body coil for comparison. Images were transferred to the Xio® planning system (Elekta AB, Stockholm, Sweden) and fused using mutual information [18].

Figure 2. A selection of uniformity maps calculated for various radiofrequency coils and uniformity correction methods. Images are shown for the axial plane (xy, left-hand column), the coronal plane (middle column) and the sagittal plane (right-hand column); and for the head coil (top row), the body coil (second row) and the oncology head and neck coil both prior to uniformity correction (third row) and after the application of surface coil intensity correction (bottom row). In all images, three uniformity bins are used indicating $\pm 10\%$ from the mean, $\pm 20\%$ and $>20\%$.

RESULTS

Figure 1 shows an image of the distortion phantom along with maps of the difference between the known and measured distances at each grid location. The co-ordinates relate to the position with respect to the centre of the phantom. In the xz plane (*i.e.* the coronal plane for head and neck imaging), the largest deviation was 1.8 mm and 2.4 mm with BWs of ± 35 kHz and ± 9 kHz, respectively. These occurred at the periphery of the phantom at a distance of approximately 18 cm from the isocentre. Deviations were within 1.0 mm for measurements made within 10 cm of the isocentre. In the xy (*i.e.* axial) plane, the maximum deviation was 1.97 mm towards the anterior of the phantom, which was at a distance of 25 cm from the isocentre. Only three measurements exceeded 1.0-mm deviation, and these were at least 15 cm from the isocentre.

Table 1 gives the SNR values for the standard head, body and oncology coils. Both the body and the head coils have similar SNR values in each of the three imaging planes. The SNR in the body coil is considerably reduced, as would be expected, owing to its larger volume. SNR values for the oncology coil vary more widely between imaging planes with axial scans performing best. Table 2 gives fractional uniformity values for head, body and oncology coils without the use of intensity correction. Values in the body and head coils are all >0.94 . The oncology coil is considerably worse without correction. Table 3 compares axial plane values for the oncology and the HNS coil both before and after the intensity corrections using both available methods. Results show that the oncology coil has improved, post correction, to an acceptable level (>0.91). Figure 2 demonstrates some of these results in the form of uniformity maps for

Figure 3. Photographs comparing the volunteer set-up used for imaging with the head-neck-spine (HNS) coil (left) and the oncology head and neck coil (right). Both the oncology coil and the integrated body coil (set-up not shown) enable the subject to be imaged in the treatment position on the flat tabletop, with the immobilisation mask fixed firmly in place. This is not possible with the HNS coil that fits snugly into the curved top of the regular scanner bed.

various coils and intensity correction schemes. The excellent uniformity of the head and body coils in each plane is clearly evident, as is the reduced SNR of the body coil. The oncology coil without correction demonstrates significant sensitivity to spatial non-uniformities, as can be expected from two separated surface coil arrays. Following the application of the intensity correction, the uniformity has improved dramatically but it is slightly worse than the other coils.

In Figure 3, the volunteer scan set-ups are shown, and in Figure 4 various images of the same volunteer are shown to illustrate the image quality achieved *in vivo* for the three RF coils and set-up options. Localiser images acquired in the coronal plane are shown in Figure 4a and clearly demonstrate the different sensitivity profiles of the three coils. In Figure 4b, an axial image is shown at a slice location near the base of the tongue. In the case of the body and oncology coils, the slice location is identical owing to imaging in the treatment position, but the image from the HNS coil shows subtle differences in anatomy owing to differences in neck angulation. These images serve to illustrate the relative uniformity and SNR obtained with these coils. SNR for each coil was equal to 5.5, 59.5 and 134.6, respectively. Figure 5 shows water-only images acquired in the oncology and HNS coils, respectively. These images serve to illustrate the good quality of the fat suppression obtained with this sequence in both coils. Water-only images in the body coil were not acquired owing to the poor SNR and excessive scan time required without parallel imaging.

The screenshot in Figure 6 is taken following the CT–MRI fusion of the head and neck phantom using the optimum imaging protocol described earlier. Although a full array of registration metrics and tests was beyond the scope of this work, the quality of the co-registration using this anthropomorphic phantom was very good.

DISCUSSION

Incorporating MRI into treatment planning involves a degree of compromise and balance between image quality and anatomical coverage and the requirements of high geometric accuracy and the need to maintain a consistent imaging/treatment position.

This paper has quantified these factors for a new scanner and a new, radiotherapy-specific coil as compared with other pre-existing coils. Both SNR and uniformity are important RF coil characteristics, with the size of the sensitive volume affecting the former, and both the size and distribution of individual elements affecting the latter. The body coil, while providing excellent spatial uniformity and good automatic fusion, suffers from poor SNR. This is a significant disadvantage when dealing with subtle tumour boundaries and in SNR demanding techniques, such as rapid dynamic contrast-enhanced MRI. The lack of separately controlled coil elements also prevents parallel imaging to be utilised to reduce the scan times. The HNS coil provides much greater SNR but cannot accommodate imaging in the treatment position. The oncology coil array presents an excellent solution to both these constraints. Using this coil, system distortion has been shown not to be a limiting factor. Image distortion is negligible for scans acquired with a 30-cm in-plane FOV or less and with 22 cm of slice coverage or less—a geometry that adequately balances the trade-offs in coil sensitivity, coverage and uniformity.

One limitation of the oncology coil is coverage of the lower neck region, and we currently restrict the planning of tumour sites to those that we know from experience will be suited to this coil. A potential solution to extend the coverage would be to use the HNS coil and then apply a robust, non-rigid body registration algorithm capable of local deformations [19], and this is the subject of ongoing work. However, in future, it is expected that vendors will bring to market more dedicated RF coils that will allow the acquisition of treatment position images across a number of sites. This will allow the radiotherapy community to build on the experience of other groups that have been so far restricted by diagnostic devices [20–22]. In the context of the current study, the addition of an anterior chest element to the oncology coil would extend coverage of the neck lower down and therefore remedy its current shortcoming. This could then be tested in the manner described earlier, perhaps using a cylindrical rather than spherical phantom.

CONCLUSION

We have outlined a series of tests that should be considered as part of the specific QA on any new or existing scanner prior to

Figure 4. T_1 weighted images of the same volunteer acquired in the body coil (left-hand column), the oncology coil (middle column) and the head-neck and spine (HNS) coil (right-hand column). (a) Coronal localiser images that demonstrate the spatial variation in coil sensitivities for each coil. (b) Axial fast spin echo images acquired at approximately the same superior-inferior slice location at the centre of the imaging volume, which demonstrates the signal-to-noise ratio (SNR) variation between the coils. The region of interest used in the SNR measurements is indicated on the middle image.

its application as a radiotherapy planning modality. This has included tests of distortion and image quality evaluations of the RF coil set-up. We have also demonstrated the utility of an

anthropomorphic phantom, which we have found useful in putting these issues into context of an “end-to-end” check of the entire process.

Figure 5. Axial T_2 weighted water-only iterative decomposition of water and fat with echo asymmetry and least squares estimation images at the same superior-inferior slice location as shown in Figure 4b. The results obtained can be compared for the oncology head and neck coil (left) and the head-neck-spine coil (right).

Figure 6. A photograph (top right) of the simulated head and neck phantom used to assess the accuracy of CT-MRI co-registration as illustrated by the screenshots of MRI-CT fused images from the treatment planning system. The primary CT image data are shown in greyscale with MRI data shown as an overlay. Images through the centre of the phantom are shown for each orthogonal plane (axial top left, coronal bottom left and sagittal bottom right).

We have used the findings of this work to implement our head and neck MRI planning service with the following recommendations with regard to image protocol: first, we restrict the slice coverage to a maximum of 20-cm coverage as a compromise between spatial distortion in the z-direction, coil sensitivity and registration performance. This is significantly shorter than the 30-cm coverage that has been suggested by the manufacturers in a general diagnostic setting. To anticipate this reduced coverage, we position the lateral coils so that they touch the patient's shoulders. We currently acquire the 12-s

calibration scan for PURE and retrospectively apply intensity correction so as to retain the original images. We maintain receiver BW at ± 35 kHz to further limit distortions and water-fat chemical shifts to within 1 mm, while maintaining a sufficient SNR.

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REFERENCES

1. Tein RD, Robbins KT. Correlation of clinical, surgical, pathologic and MR fat suppression results for head and neck cancer. *Head Neck* 1992;14:278–84.
2. Philips CD, Gay SB, Newton RL, Levine PA. Gadolinium-enhanced MRI of tumours of the head and neck. *Head Neck* 1990;12:308–15.
3. Emami B, Sethi A, Petruzelli GJ. Influence of MRI on target volume delineation and IMRT planning in nasopharyngeal carcinoma. *Int J Radiat Oncol Biol Phys* 2003;57:481–8.
4. Sievers KW, Greess H, Baum U, Dobritz M, Lenz M. Paranasal sinuses and nasopharynx CT and MRI. *Eur J Radiol* 2000;33:185–202.
5. Ahmed M, Schmidt M, Sohaib A, Kong C, Burke K, Richardson C, et al. The value of magnetic resonance imaging in target volume delineation of base of tongue tumours—a study using flexible surface coils. *Radiother Oncol* 2010;94:161–7.
6. Geets X, Daisne JF, Arcangeli S, Coche E, De Poel M, Duprez T, et al. Inter-observer variability in the delineation of pharyngolaryngeal tumour, parotid glands and cervical spinal cord: comparison between CT-scan and MRI. *Radiother Oncol* 2005;77:25–31.

7. Beavis AW, Gibbs P, Dealey RA, Whitton VJ. Radiotherapy treatment planning of brain tumours using MRI alone. *Br J Radiol* 1998; 71:544–8.
8. Houweling AC, van der Meer S, van der Wal E, Terhaard CH, Raaijmakers CP. Improved immobilization using an individual head support in head-and-neck cancer patients. *Radiother Oncol* 2010;96:100–3.
9. Hanvey S, Glegg M, Foster J. Magnetic resonance imaging for radiotherapy planning of brain cancer patients using immobilization and surface coils. *Phys Med Biol* 2009;54: 5381–94. doi: [10.1088/0031-9155/54/18/002](https://doi.org/10.1088/0031-9155/54/18/002).
10. O'Donnell M, Edlestein WA. NMR imaging in the presence of magnetic field inhomogeneities and gradient field nonlinearities. *Med Phys* 1985;12:20–6.
11. Petersch B, Bogner J, Fransson A, Lorang T, Pötter R. Effects of geometric distortion in 0.2T MRI on radiotherapy treatment planning of prostate cancer. *Radiother Oncol* 2004;71:55–64.
12. Mah D, Steckner M, Palacio E, Mitra R, Richardson T, Hanks GE. Characteristics and quality assurance of a dedicated open 0.23 T MRI for radiation therapy simulation. *Med Phys* 2002;29:2541–7.
13. Fransson A, Andero P, Potter R. Aspects of MR image distortions in radiotherapy treatment planning. *Strahlenther Onkol* 2001;177:59–73.
14. Moerland MA, Beersma R, Bhagwandien R, Wijrdeman HK, Bakker CJ. Analysis and correction of geometric distortions in 1.5 T magnetic resonance images for use in radiotherapy treatment planning. *Phys Med Biol* 1995;40:1651–64.
15. Moore CS, Liney GP, Beavis AW. A quality assurance phantom for verification and registration of CT and MRI data sets for treatment planning of radiotherapy for head and neck cancers. *J Appl Clin Med Phys* 2004;5:25–35.
16. Brunt JN. Computed tomography-magnetic resonance image registration in radiotherapy treatment planning. *Clin Oncol (R Coll Radiol)* 2010;22:688–97. doi: [10.1016/j.clon.2010.06.016](https://doi.org/10.1016/j.clon.2010.06.016).
17. British Standards Institution. Magnetic resonance equipment for medical imaging part 1: determination of essential image quality parameters. BS EN 62464-1:2007. London, UK: British Standards Institution, 2007.
18. Wang X, Li L, Hu C, Qiu J, Xu Z, Feng Y. A comparative study of three CT and MRI algorithms in nasopharyngeal carcinoma. *J Appl Clin Med Phys* 2009;10:3–10.
19. Fallone BG, Rivest DR, Riauka TA, Murtha AD. Assessment of a commercially available automatic deformable registration system. *J Appl Clin Med Phys* 2010;11:3175.
20. van der Heide UA, Houweling AC, Groenendaal G, Beets-Tan RG, Lambin P. Functional MRI for radiotherapy dose painting. *Magn Reson Imag* 2012;30:1216–23. doi: [10.1016/j.mri.2012.04.010](https://doi.org/10.1016/j.mri.2012.04.010).
21. Verduijn GM, Bartels LW, Raaijmakers CP, Terhaard CH, Pameijer FA, van den Berg CA. Magnetic resonance imaging protocol optimization for delineation of gross tumor volume in hypopharyngeal and laryngeal tumors. *Int J Radiat Oncol Biol Phys* 2009; 74:630–6. doi: [10.1016/j.ijrobp.2009.01.014](https://doi.org/10.1016/j.ijrobp.2009.01.014).
22. McJury M, O'Neill A, Lawson M, McGrath C, Grey A, Page W, et al. Assessing the image quality of pelvic MR images acquired with a flat couch for radiotherapy treatment planning. *Br J Radiol* 2011;84:750–5.