Motion-corrected $^{23}$Na MRI of the human brain using interleaved $^{1}$H 3D navigator images

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Purpose: To evaluate the feasibility of motion correction for sodium ($^{23}$Na) MRI based on interleaved acquired 3D proton ($^{1}$H) navigator images.

Methods: A 3D radial density-adapted sequence for interleaved $^{23}$Na/$^{1}$H MRI was implemented on a 7 Tesla whole-body MRI system. The $^{1}$H data obtained during the $^{23}$Na acquisition were used to reconstruct 140 navigator image volumes with a nominal spatial resolution of (2.5 mm)$^3$ and a temporal resolution of 6 s. The motion information received from co-registration was then used to correct the $^{23}$Na image dataset, which also had a nominal spatial resolution of (2.5 mm)$^3$. The approach was evaluated on six healthy volunteers, whose motion during the scans had different intensities and characteristics.

Results: Interleaved acquisition of two nuclei did not show any relevant influence on image quality (SNR of 13.0 for interleaved versus 13.2 for standard $^{23}$Na MRI and 176.4 for interleaved versus 178.0 for standard $^{1}$H MRI). The applied motion correction increased the consistency between two consecutive scans for all examined volunteers and improved the image quality for all kinds of motion. The SD of the differences ranged between 2.30% and 6.96% for the uncorrected and between 2.13% and 2.67% for the corrected images.

Conclusion: The feasibility of interleaved acquired $^{1}$H navigator images to be used for retrospective motion correction of $^{23}$Na images was successfully demonstrated. The approach neither affected the $^{23}$Na image quality nor elongated the scan time and can therefore be an important tool to improve the accuracy of quantitative $^{23}$Na MRI.

KEYWORDS
$^{23}$Na MRI, 7 Tesla, brain MRI, interleaved dual-nuclear MRI, motion correction, ultrahigh field strength
1 | INTRODUCTION

Sodium ($^{23}$Na) MRI has been established as a noninvasive technique to determine the tissue sodium concentration in the human brain, and numerous studies have revealed new metabolic information for many diseases such as stroke, tumors, or multiple sclerosis. Due to the important role of sodium in the metabolism of human cells, $^{23}$Na MRI is used as a versatile tool in biomedical research.

Although improved hardware capabilities and increased magnetic field strengths have established $^{23}$Na MRI in clinical research, the considerably lower in vivo concentrations, fourfold lower gyromagnetic ratio, and very fast relaxation compared to proton ($^{1}$H) MRI still cause substantial restrictions in clinical research applications. To reach a reasonable SNR and a spatial resolution of about 2.5–3.5 mm, acquisition times of about 10–15 min are necessary. Furthermore, for every patient additional scan time is usually required for adjustment measurements as well as for anatomical $^{1}$H MRI scans, which are used in the image postprocessing (e.g., partial volume correction). Over such long scan times, at least small motion of the head is very likely to occur and can lead to quantification errors, which may, if unnoticed, result in false clinical conclusions. Furthermore, repeating scans due to motion artifacts is costly.

To correct such inaccuracies resulting from head motion, a method that does not prolong scan time and that does not depend on additional devices is desirable. Whereas many approaches have been suggested for $^{1}$H MRI for $^{23}$Na MRI of the brain, only Lu et al. proposed a 3D $^{23}$Na navigator-based retrospective motion correction, which uses a second echo with lower spatial but higher temporal resolution. However, as a result of the intrinsically low SNR of $^{23}$Na MRI and the fast signal decay, the proposed 8 mm isotropic spatial resolution is only sufficient to correct for relatively large motion amplitudes, and the temporal resolution of about 1 min further limits the correction capabilities. To improve both the spatial and temporal resolution of the navigator images and therefore the precision of the motion correction, performing $^{1}$H acquisitions instead of $^{23}$Na could be a promising approach due to the considerably higher MR signal.

Interleaved or simultaneous dual-nuclear MR acquisitions, where data of two different nuclei are acquired within the same sequence, have been performed at different magnetic field strengths, recently even at modern 3 Tesla clinical systems without the need of hardware modification. At 7 Tesla human scanners, however, interleaved dual-nuclear MRI has always been conducted on research systems using additional hardware modifications or a special software interface developed by the user. With the recent generation of clinically approved 7 Tesla platforms, interleaved MRI measurements have also been enabled by the manufacturer, which has been demonstrated recently with interleaved $^{31}$P/$^{1}$H MR spectroscopy.

The purpose of this work was to introduce an improved retrospective motion-correction method for $^{23}$Na MRI by interleaving it with $^{1}$H 3D navigator imaging. This was accomplished without applying any hardware modification and without the need of additional acquisition time.

2 | METHODS

All measurements were conducted on a whole-body 7 Tesla MR system (Magneton Terra, Siemens Healthcare, Erlangen, Germany) with multinuclear capability using a dual-tuned $^{23}$Na/$^{1}$H head RF coil (RAPID Biomedical, Rimpar, Germany), which consists of a dual-tuned $^{23}$Na/$^{1}$H quadrature transmit–receive birdcage coil and an additional 32 channel receive-only array for $^{23}$Na MRI. In vivo measurements were performed on six healthy volunteers (4 males, 2 females, 27 ± 7 years) who provided written informed consent before being scanned. The study was approved by the local ethical review board.

2.1 Interleaved $^{23}$Na/$^{1}$H density-adapted 3D radial projection pulse sequence

The motion-correction MRI measurements were conducted using a density-adapted 3D radial projection (DA3DRAD) pulse sequence, which was adapted to allow for interleaved dual-nuclear acquisition with different acquisition parameters for $^{23}$Na and $^{1}$H. The sequence scheme is shown in Figure 1. First, the $^{23}$Na signal is excited and acquired. The idle time before the next $^{23}$Na excitation pulse is then used to acquire $^{1}$H MRI data. The idle time before the next $^{23}$Na excitation pulse is then used to acquire $^{1}$H MRI data.

2.2 MR imaging and image reconstruction

At the beginning of every measurement, a $B_0$-shim based on $^{1}$H MRI using the standard brain $B_0$-shim provided by the vendor and a global flip angle calibration for $^{23}$Na MRI were performed. For interleaved $^{23}$Na/$^{1}$H MRI, the interleaved DA3DRAD sequence described above was conducted with the following parameters (for definition see Figure 1): $TR_{23Na} = 120$ ms, $TP_{23Na} = 0.6$ ms, $TE_{23Na} = 0.4$ ms, $TRO_{23Na} = 10$ ms, $FA_{23Na} = 87^\circ$, 7000 $^{23}$Na projections and $TR_{1H} = 3.9$ ms, $TP_{1H} = 1.5$ ms, $TE_{1H} = 1.3$ ms, $TRO_{1H} = 1$ ms, $FA_{1H} = 6^\circ$, 196,000 $^{1}$H
FIGURE 1 Scheme of the interleaved $^{23}\text{Na}/^{1}\text{H}$ sequence. During every $^{23}\text{Na}$ repetition time $TR_{23\text{Na}}$, the acquisition of one $^{23}\text{Na}$ projection is followed by the acquisition of $N$ $^{1}\text{H}$ MRI projections with a repetition time of $TR_{1\text{H}}$ such that the complete $^{23}\text{Na}$ recovery time is used. Both nuclei are excited using rectangular RF pulses with pulse lengths and flip angles of $TP_{23\text{Na}}$ and $FA_{23\text{Na}}$ for $^{23}\text{Na}$ and $TP_{1\text{H}}$ and $FA_{1\text{H}}$ for $^{1}\text{H}$. The signal of both nuclei is acquired by a DA3DRAD readout ($TE_{23\text{Na}}/TE_{1\text{H}}$ and readout duration $TRO_{23\text{Na}}/TRO_{1\text{H}}$ for $^{23}\text{Na}/^{1}\text{H}$). Every readout is followed by a rewinder gradient. Additionally, a spoiler gradient is used to dephase the remaining $^{1}\text{H}$ magnetization by $2\pi$ per voxel $^{1}\text{H}$, hydrogen; $^{23}\text{Na}$, sodium; DA3DRAD, density-adapted 3D radial projection projections (28 $^{1}\text{H}$ projections per $^{23}\text{Na}$ projection), nominal spatial resolution (2.5 mm)$^3$ for $^{23}\text{Na}$ and $^{1}\text{H}$, and 3D golden-angle projection scheme$^{22}$ for $^{23}\text{Na}$ and $^{1}\text{H}$. The $^{23}\text{Na}$ parameters were adapted from a measurement protocol currently used in clinical studies at our institution.

All images were reconstructed offline using a custom-written MatLab script (MatLab, MathWorks, Nat- ick, MA). The reconstruction was based on re-gridding on a Cartesian grid after density compensation using a Kaiser Bessel kernel with width 4.0 and a twofold oversampling. To increase the SNR and avoid Gibbs’ ringing artifacts, a Hamming filter was applied$^{23,24}$ The images were finally obtained by a Fast Fourier Transform of the $k$-space data, which were zero-filled to a spatial resolution of (1 mm)$^3$. Due to the acquisition with a 3D golden-angle projection scheme, an arbitrary number of projections could be used for image reconstruction.$^{22}$ The multichannel data were combined using an adaptive combination reconstruction.$^{25,26}$

2.3 Evaluation of multinuclear acquisition interactions

To evaluate a potential influence of the $^{1}\text{H}$ navigator acquisitions on the quality of $^{23}\text{Na}$ MRI, measurements of a spherical phantom (7.5% agarose gel, 100 mmol/L NaCl solution) were conducted. For the single-nuclear comparison measurement, reference data from each individual nucleus were acquired by running the interleaved sequence with either the $^{1}\text{H}$ or the $^{23}\text{Na}$ RF power and readout gradients turned off, without changing any of the other parameters (sequence parameters in subsection 2.2). In the following, images acquired with the interleaved dual-nuclear pulse sequence are labeled DA3DRAD$_{\text{IL}}$ when signals of both nuclei were excited and acquired, and DA3DRAD$_{23\text{Na}}$/DA3DRAD$_{1\text{H}}$ when only the $^{23}\text{Na}$ or $^{1}\text{H}$ signal was excited and measured. The SNR was calculated using one signal region of interest (signal intensity higher than 35% of the maximum signal intensity) and one noise region of interest (signal intensity lower than 15%/5% of the maximum signal intensity for $^{23}\text{Na}/^{1}\text{H}$) in the image.$^{27}$

2.4 Navigator images and motion correction

The navigator images were all co-registered to the first one ($I_1$) using the realign function in SPM12 (Wellcome Trust Centre for Neuroimaging, London, UK) assuming rigid-body motion. The estimated
transformation parameters
\[
T_{I_i \rightarrow I_1} = \begin{pmatrix}
T_{x_i, I_i \rightarrow I_1} \\
T_{y_i, I_i \rightarrow I_1} \\
T_{z_i, I_i \rightarrow I_1}
\end{pmatrix}
\]
for the translation and
\[
R_{I_i \rightarrow I_1} = R_{z_i, I_i \rightarrow I_1} R_{y_i, I_i \rightarrow I_1} R_{x_i, I_i \rightarrow I_1}
\]
for the rotation

(1)
of every navigator image \( I_i \) were then used to directly correct the k-space data of the corresponding \(^{23}\text{Na}\) projections acquired during the same time. First, the rotation was performed by rotating the \(^{23}\text{Na}\) k-space locations
\[
k_{S_i, \text{cor} r} = R_{I_i \rightarrow I_1} \cdot k_{S_i},
\]
and afterward the translation was achieved by applying a phase shift of the \(^{23}\text{Na}\) k-space samples
\[
S_{I_i, \text{cor}r} = S_i e^{-2\pi i (k_{S_i, \text{cor}r} \cdot T_{I_i \rightarrow I_1})}.
\]
(3)

Thereby, \( S_i \) denotes a \(^{23}\text{Na}\) complex-valued k-space sample acquired during the same time as the navigator image \( I_i \), and \( k_{S_i} \) is the corresponding k-space location. The motion-corrected \(^{23}\text{Na}\) image was then obtained by reconstructing the corrected k-space data.

### 2.5 In vivo motion-correction measurements

To demonstrate the capacity of the proposed motion-correction method, \(^{23}\text{Na}/^{1}\text{H}\) interleaved MRI measurements (sequence parameters in subsection 2.2) were performed on a total of six healthy volunteers. For every volunteer, two scans were performed with different motion during the measurement. The motion pattern and intensity were up to the volunteers and not further specified. All \(^{23}\text{Na}\) data sets were reconstructed with and without applying the motion correction, and the images were normalized to the mean signal intensity in a central region of interest in the vitreous humor of the eyes. To demonstrate the benefits of the motion correction, the consistency between the two scans of each volunteer was evaluated for the uncorrected as well as for the motion-corrected images by subtracting the two images after registration and visualizing the distribution of the differences in a histogram. Furthermore, the SD was calculated for the distributions of the uncorrected and corrected images to quantify the improvements.

Additionally, in order to find the optimal number of projections for navigator image registration for the chosen spatial resolution of (2.5 mm)\(^3\), for two volunteers the described approach was repeated for navigator image data sets reconstructed out of 700, 1400, 2800, and 3500 projections, which correspond to a temporal resolution of 3, 6, 12, and 15 s, respectively. These values result from the chosen parameters (7000 \(^{23}\text{Na}\) projections, 28 \(^{1}\text{H}\) projections per \(^{23}\text{Na}\) projection) because they provide integer values for the corrected \(^{23}\text{Na}\) projections per navigator image.

### RESULTS

### 3.1 Hardware and sequence evaluation

The results of the interleaved sequence evaluation are shown in Supporting Information Figure S1. The \(^{23}\text{Na}\) comparison measurements did not reveal any relevant influence of the additional \(^{1}\text{H}\) acquisitions on the \(^{23}\text{Na}\) image quality. Also, \(^{23}\text{Na}\) MRI had no relevant influence on \(^{1}\text{H}\) MRI. The SNR over the whole phantom obtained with the DA3DRAD\(^{1}\text{H}\) sequence was almost identical to the SNR obtained with the DA3DRAD\(^{23}\text{Na}\) and DA3DRAD\(^{1}\text{H}\) sequence (13.0 vs. 13.2 for \(^{23}\text{Na}\) and 176.4 versus 178.0 for \(^{1}\text{H}\)).

### 3.2 In vivo motion-correction measurements

Despite the relatively low SNR of the images acquired with the \(^{1}\text{H}\) birdcage of the dual-tuned \(^{23}\text{Na}/^{1}\text{H}\) head RF coil (see Supporting Information Figure S2), for both volunteer measurements conducted in order to find the optimal number of projections for navigator image registration, the motion correction improved the consistency between the two scans for all evaluated \(^{23}\text{Na}\) images reconstructed using different numbers of projections. The results are shown in Figure 2 (volunteer 1) and Supporting Information Figure S3 (volunteer 2). In both cases, the best consistency between the two scans was achieved with navigator images reconstructed out of 1400 projections. Therefore, in the following, this number of projections was used for the reconstruction of each 3D navigator image data set in all volunteer measurements. This leads to one navigator image data set every 6 s and 140 3D navigator image data sets in total during the interleaved \(^{23}\text{Na}/^{1}\text{H}\) MR acquisition. As a result, every single \(^{1}\text{H}\) navigator image data set was used to correct the corresponding 50 consecutive \(^{23}\text{Na}\) projections acquired during the same time.

Exemplary results of the determined motion parameters as well as the \(^{23}\text{Na}\) images with and without motion correction and the differences between uncorrected and corrected images are shown in Figure 3 for volunteer 3 and in Figure 4 for volunteer 4. Furthermore, the distributions of the differences between the uncorrected and
FIGURE 2 Results of the motion correction for volunteer 1 using 700, 1400, 2800, and 3500 projections for the reconstruction of the $^1$H 3D navigator image data sets. Exemplary navigator images for the different numbers of projections are presented in (A) to get a visual impression of the image quality. Furthermore, for each number of projections the distribution of the differences between the uncorrected as well as between the corrected images are shown (B). The motion correction improved the consistency between the two scans for all evaluated numbers of projections. The best results were achieved for navigator image data sets reconstructed out of 1400 consecutive projections.
FIGURE 3 Exemplary measurement results of volunteer 3 for the two consecutive scans (scan 1 (A) and scan 2 (B)). All $^{23}$Na images were normalized to the mean signal intensity in a central region of interest in the vitreous humor. In scan 2, clearly stronger movements were detected. In both cases, differences between the uncorrected and corrected image are visible. Especially for scan 2, the correction clearly improved the image quality and reduced motion artifacts. Furthermore, the difference between the uncorrected and the corrected images of both cases respectively were significantly reduced.
FIGURE 4 Exemplary measurement results of volunteer 4 for the two consecutive scans (scan 1 (A) and scan 2 (B)). All $^{23}$Na images were normalized to the mean signal intensity in a central region of interest in the vitreous humor. In scan 2, clearly stronger movements were detected. For scan 1, hardly any differences between the uncorrected and corrected image are visible. For scan 2, the correction clearly improved the image quality and reduced all kinds of motion artifacts as signal blurring and wrongly depicted anatomical structures. Furthermore, the difference between the uncorrected and the corrected images of both scans respectively were significantly reduced.
corrected images of the consecutive scans for all examined volunteers are presented in Figure 5. In all cases, the interleaved motion-correction approach reduced the differences between the uncorrected and the corrected $^{23}$Na images, which proves the increased image accuracy. In the measurements, motion of different intensity and characteristic were detected—abrupt as well as continuous movements. Especially for scans including strong movements with translations of up to 5 mm and rotations of up to 10°, which showed clear motion artifacts, the image quality was significantly improved, and the corrected images did not exhibit obvious artifacts such as washed-out anatomical structures anymore (in particular, see Figure 4B).

4 | DISCUSSION

In this work, a retrospective motion-correction method for $^{23}$Na MRI, based on interleaved acquired 3D $^1$H navigator images, was implemented. The navigator images were obtained without any hardware or software modification as well as without time penalty. The additional $^1$H acquisitions did not affect the image quality of $^{23}$Na MRI.

The presented approach corrected motion of different characteristics and intensities. For all examined volunteers, it reduced differences between two consecutive scans and therefore improved the $^{23}$Na MR image quality. For strong and abrupt movements persisting during the whole scan, parts of the motion artifacts, especially signal blurring, still remained in the corrected image. For continuous motion during the entire scan and for single strong abrupt movements, no obvious motion artifacts were observed in the corrected images. For all volunteers, apart from general signal blurring, all clear motion artifacts were removed from the images. Thus, the correction may prevent the necessity to repeat measurements.

In this study, the spatial resolution of the $^1$H navigator images was chosen identical to the $^{23}$Na resolution to get high spatial accuracy in the registration process. If the focus is more on the temporal resolution, the navigator acquisition time could be shortened by using lower $^1$H spatial resolutions. This tradeoff between spatial and temporal resolution can always be adapted to the specific needs of the study. With the parameters of this study, one navigator data set was used to correct only 0.7% of the acquired $^{23}$Na projections. This should result in sufficient correction capability for most types of motion, like occasional abrupt or continuous positional changes of the head.

Different methods to correct periodical movements such as cardiac$^{26,30}$ or respiratory$^{31}$ motion exist for $^{23}$Na MRI. However, these are not suited for correction of aperiodic motion such as motion of the head. Retrospective motion correction of random head movements for $^{23}$Na MRI has only been performed by Lu et al.$^{10}$ using a second $^{23}$Na echo with a long TE and was therefore mainly based on the $^{23}$Na CSF signal. Due to the significantly higher $^1$H MR signal compared to $^{23}$Na, the proposed interleaved $^{23}$Na/$^1$H approach, as applied in this work, provides a more than elevenfold higher temporal resolution while increasing the spatial resolution of the navigator images by a factor of $\sim 32$ compared to Lu et al.$^{10}$ The potential for motion correction is thereby clearly improved. Because a DA3DRAD acquisition with repeated sampling of the k-space center is used for $^{23}$Na MRI, a self-navigated motion-correction approach, as it is known for $^1$H MRI$^{32}$, would be another possibility based on $^{23}$Na MRI, which, however, also does not reach the potential of the interleaved $^{23}$Na/$^1$H method due to the significantly lower SNR and the resulting lower spatial and temporal resolution (see Supporting Information Figure S4).

The $^1$H navigator data were acquired using a DA3DRAD readout scheme with a golden-angle projection scheme, which offers some beneficial properties. The excitation and acquisition of the $^{23}$Na signal take about 12 ms (i.e., $3\text{TR}_{1H}$) with the sequence parameters used. Therefore, the $^1$H acquisition had to be segmented, and thus no uniform steady state is reached. Although this influenced the contrast of the $^1$H navigator images, no artifacts deteriorated the image quality for reliable image registration because the different intensities are distributed over the whole k-space due to the golden-angle projection scheme.$^{22}$ Non-segmented $^1$H acquisition (e.g., $\text{TR}_{1H} > 12$ ms) is unfavorable because this would reduce the number of $^1$H projections by at least a factor of 3, resulting in accordingly lower temporal resolution of the navigator images. To reach a uniform steady state for $^1$H, simultaneous acquisition of $^{23}$Na and $^1$H would be a possibility$^{18,33}$ which, however, requires additional hardware and therefore limits applicability. Furthermore, due to the golden-angle projection scheme, the temporal resolution of the navigator image data sets can even be adapted retrospectively, depending on the needs of the study. This has also been of advantage for showing the feasibility of the interleaved method because different parameters and approaches could have been evaluated retrospectively using the same dataset.

Despite these advantages, the $^1$H acquisition strategy has not been further optimized and may be improved in order to reach higher spatial or temporal resolution of the $^1$H navigator image data sets. Further improvements of the temporal resolution could be achieved by using a sliding-window reconstruction, which is often used for example in myocardial imaging.$^{34}$ Because this method, however, does not generally lead to improvements for
FIGURE 5  Distributions of the differences between the uncorrected and corrected images of the two scans for all six examined volunteers. The SD of the distribution of the uncorrected $^{23}$Na images lies between 2.3% and 6.96%, depending on the motion characteristic of the two scans. For all volunteers, the motion correction reduced the SD of the distribution of the corrected $^{23}$Na images and therefore improved the image quality by increasing the consistency between two consecutive scans.
the presented application (see Supporting Information Figure S5) without specific further optimization, this may be investigated in future work. Furthermore, other methods already used in 1H retrospective motion-correction approaches for improving navigator data could be tested. Improvements may, for example, be achieved by using other 1H excitation strategies such as fat navigators by using k-space sampling trajectories optimized for fast imaging such as EPI or by additionally taking into account projection moments of the 1H data. However, due to the predetermined timing of the 23Na acquisition and the additional SAR contribution in combination with the reduced quality of the 1H channel of dual-tuned RF coils compared to RF coils commonly used in 1H motion-correction studies, the applicability may be limited. In particular, the single channel constraint in our setting, which is a common restriction of dual-tuned RF coils, does not allow for parallel imaging techniques, and multichannel coils might diminish the accuracy of projection moment estimation. The proposed method in this work is expected to work with all kinds of dual-tuned RF coils, for example independent of the number of 1H channels and the resulting homogeneity.

A disadvantage of the interleaved 23Na/1H method is the need for a dual-tuned RF coil. Even though 23Na MRI with RF coils comprising an additional 1H channel benefit from advantages with localizer and adjustment measurements, single-resonant 23Na RF coils or dual-tuned X-nuclei RF coils (e.g., 23Na/35Cl RF coil) are often used, for which our approach is not applicable. Furthermore, interleaved measurements lead to a higher SAR. Even though this did not influence the choice of the acquisition parameters in the current 23Na MRI protocol, it may be a limiting factor for other studies using inversion recovery or triple-quantum filtered 23Na MRI that typically have higher SAR.

Our work successfully demonstrates the feasibility of the presented motion-correction approach on a typical subject group size for current technical development MR studies. Because the signal intensities were normalized to vitreous humor, relative deviations of the normalized signal intensity could be derived that should correspond to relative concentration deviations. However, for conclusions about the actual improvement in quantitative accuracy in future clinical 23Na MRI studies, which will also depend on the characteristics of the examined pathology, quantitative measurements including crucial postprocessing steps for concentration determination such as correction of the coil sensitivity and partial volume correction are required. Our method is particularly promising for applications in studies on patients with neurological disorders such as multiple sclerosis because motion is typically a limiting factor in these cohorts.

5 | CONCLUSION

We successfully demonstrated the feasibility of retrospective motion correction in 23Na brain MRI using interleaved acquired 23Na navigator images. The approach neither affected the 23Na image quality nor elongated the scan time. Especially regarding the advances in 23Na spatial resolutions with higher field strength, the presented results are promising for improving image quality in future 23Na MRI studies.

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SUPPORTING INFORMATION
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FIGURE S1. The $^{23}$Na images of the DA3DRAD$_{1H}$ sequence were compared to the ones acquired with the DA3DRAD$_{23Na}$ sequence (A) and the $^{1}$H images to the ones of the DA3DRAD$_{1H}$ sequence (B). All images were individually normalized to the maximum signal intensity of the data set. The mean differences over the whole phantom were $(-0.3 \pm 1.9)$ % of the maximum signal value for $^{23}$Na and $(-0.06 \pm 0.12)$ % for $^{1}$H. Therefore, in both cases no relevant deviation was observed.

FIGURE S2. The dual-tuned $^{23}$Na/$^{1}$H head RF coil is optimized for $^{23}$Na MRI measurements and the included $^{1}$H quadrature Tx/Rx birdcage so far has only been used for localization measurements and for performing B$_0$-shims based on $^{1}$H. To evaluate the quality of the $^{1}$H quadrature Tx/Rx birdcage, single-nuclear $^{1}$H in vivo measurements of a healthy volunteer were performed using the dual-tuned $^{23}$Na/$^{1}$H as well as an 1Tx/32Rx $^{1}$H head RF coil (Nova Medical, Wilmington, MA, USA) which is a commercial, frequently used 7T brain RF coil. These measurements were conducted using a common DA3DRAD sequence with, apart from the pause due to the $^{23}$Na acquisition, parameters identical to those of the DA3DRAD$_{1H}$ sequence: TR = 3.9 ms, TP = 1.5 ms, TE = 1.3 ms, TRO = 1 ms, FA = 6°, 3500 $^{1}$H projections, nominal spatial resolution 2.5 mm isotropic, 3D golden angle projection scheme. Furthermore, $^{1}$H MRI measurements of a spherical phantom (7.5% agarose gel, 100 mmol/L NaCl solution) were used to compare the SNR of both RF coils. The SNR was calculated using a ROI in the image and an additional noise scan. The comparison of the 1Tx/32Rx $^{1}$H head RF coil and the $^{1}$H birdcage of the dual-tuned $^{23}$Na/$^{1}$H head RF coil clearly showed that the image quality of the latter was lower, as expected. Over the whole agarose gel phantom, the 1Tx/32Rx $^{1}$H RF coil provided a five-fold higher SNR than the dual-tuned $^{23}$Na/$^{1}$H RF coil. In the in vivo measurements, the image quality of the 1Tx/32Rx $^{1}$H RF coil is mainly restricted by undersampling artifacts and not by the SNR for all considered numbers of projections. In contrast, in the images acquired with the $^{1}$H channel of the dual-tuned $^{23}$Na/$^{1}$H RF coil undersampling artifacts are strongly superimposed by noise.

FIGURE S3. Results of the motion correction for volunteer 2 using 700, 1400, 2800 and 3500 projections for the reconstruction of the $^{1}$H 3D navigator image data sets. Exemplary navigator images for the different numbers of projections are presented in (A) to get a visual impression of the image quality. Furthermore, for each number of projections the distribution of the differences between the uncorrected as well as between the corrected images are shown (B). The motion correction improved the consistency between the two scans for all evaluated numbers of projections. The best results were achieved for navigator image data sets reconstructed out of 1400 consecutive projections.

FIGURE S4. As a DA3DRAD acquisition with repeated sampling of the k-space center is used for $^{23}$Na MRI, a self-navigated motion correction approach which uses low spatial resolution image data sets with higher temporal resolution reconstructed out of the central part of the k-space. Such a self-navigation was evaluated for volunteer 2 and volunteer 3 and compared to the results of the interleaved $^{23}$Na/$^{1}$H method. The spatial resolution of the navigator images was empirically chosen to be 10 mm and for the temporal resolution with 6 s (50 $^{23}$Na projections per image) the same value as for the interleaved method was used. The reconstruction, co-registration, and evaluation have been done in exactly the same way as for the interleaved $^{23}$Na/$^{1}$H method. The determined motion parameters for the second scan of volunteer 3 are shown for the self-navigated approach (A) as well as the interleaved $^{23}$Na/$^{1}$H approach (B). The translation values of the self-navigated method are clearly noisier compared to the interleaved method, especially in H-F-direction. Furthermore, the rotations were not detected properly for the self-navigation and clear deviation compared to the interleaved method can be seen. Furthermore, the distribution of the differences between the uncorrected as well as between the corrected images are shown for the self-navigated and for the interleaved motion correction for volunteer 2 (C, D) and volunteer 3 (E, F). For both volunteers, the self-navigated approach performs clearly worse than the interleaved $^{23}$Na/$^{1}$H method.

FIGURE S5. The $^{1}$H navigator data was acquired using a density-adapted 3D radial projection pulse sequence with a 3D golden-angle projection scheme. As this allows for the reconstruction of an image out of an arbitrary number of consecutively acquired projections, the $^{1}$H navigator data also allow for a sliding window approach to determine the motion parameters. This was evaluated for volunteer 3 and the results were compared to the motion correction method without using the sliding window. Like for the motion correction without sliding window, 1400 consecutive projections were used for the reconstruction of one single 3D navigator image data set. The window step size was chosen to be 350 projections. This leads to 557 3D navigator image data sets in total during the interleaved $^{23}$Na/$^{1}$H MR acquisition. The reconstruction of the motion corrected $^{23}$Na images and their evaluation was done in...
exactly the same way as for the approach without sliding window.

The motion parameters derived by the co-registration of 3D navigator image data sets without using the sliding window are shown for scan 1 (A) and scan 2 (B) and the corresponding parameters obtained using the sliding window (C and D). Only very minor differences occur between the two methods. The main motion characteristic is the same in both cases. Furthermore, the distribution of the differences between the uncorrected as well as between the corrected images are shown for the motion correction without (E) and with sliding window (F). Again, hardly any differences between the two approaches can be found. Regarding the SD of the differences between the corrected images, the motion correction without sliding window performed negligibly better.

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