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Synchronous Radial ^1H and ^{23}Na Dual-Nuclear MRI on a Clinical MRI System, Equipped With a Broadband Transmit Channel

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Abstract

The purpose of this work was to synchronously acquire proton (^1H) and sodium (^{23}Na) image data on a 3T clinical MRI system within the same sequence, without internal modification of the clinical hardware, and to demonstrate synchronous acquisition with $^1\text{H}/^{23}\text{Na}$ -GRE imaging with Cartesian and radial k-space sampling. Synchronous dual-nuclear imaging was implemented by: mixing down the ^1H signal so that both the ^{23}Na and ^1H signal were acquired at ^{23}Na frequency by the conventional MRI system; interleaving $^1\text{H}/^{23}\text{Na}$ transmit pulses in both Cartesian and radial sequences; and using phase stabilization on the ^1H signal to remove mixing effects. The synchronous $^1\text{H}/^{23}\text{Na}$ setup obtained images in half the time necessary to sequentially acquire the same ^1H and ^{23}Na images with the given setup and parameters. Dual-nuclear hardware and sequence modifications were used to acquire ^{23}Na images within the same sequence as ^1H images, without increases to the ^1H acquisition time. This work demonstrates a viable technique to acquire ^{23}Na image data without increasing ^1H acquisition time using minor additional custom hardware, without requiring modification of a commercial scanner with multinuclear capability.

Keywords

sodium; multinuclear imaging; metabolic imaging; radiofrequency coils; dual-nuclear mri

INTRODUCTION

Sodium (^{23}Na) is the second most abundant NMR nucleus in human body after ^1H . Sodium MRI is less sensitive than ^1H -MRI, due to the relatively low in vivo ^{23}Na concentrations,

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rapid bi-exponential signal decay¹, and a low gyromagnetic ratio². Despite these challenges, ²³Na-MRI is currently being pursued as a complement to clinical ¹H-MRI to enable better characterization and assessment of tumor viability^{3,4}, cartilage health^{5–8}, renal failure^{9,10}, tissue damage following stroke^{11,12}, and multiple sclerosis¹³.

A significant barrier to ²³Na-MRI in clinical settings is the need for expensive and specialized hardware to enable non-proton imaging, in addition to the increased time requirements for obtaining both ²³Na and ¹H data that are typically acquired sequentially in separate MRI sequences^{3–13}. By acquiring both datasets within a single pulse sequence, synchronous ¹H/²³Na imaging has the potential to obtain both ¹H-MRI and ²³Na information without increasing acquisition time, thereby improving the practicality of ²³Na-MRI.

Synchronous multinuclear imaging (**SMI**), where images from more than one nuclei are acquired in a single pulse sequence, has been investigated on clinical 1.5T¹⁴, 3T^{15–21}, and 7T^{17,22} MRI systems, imaging ¹H synchronously with ²³Na^{14–17}, ¹⁹F¹⁸, ³He¹⁹, and ³¹P^{20,22} and obtaining synchronous triple-nuclei ¹H/³He/¹²⁹Xe images²¹. Although SMI was demonstrated initially in 1986^{14,23}, only recently has there been renewed interest in SMI on human research systems, with the majority of publications using a software-only method on Philips systems^{15,17–19,21}, and additional hardware added to enable SMI on a Siemens system²².

The increased complexities in both hardware and pulse sequence have limited the application of SMI on clinical MRI systems, as typical clinical systems are capable of irradiating or receiving only at ¹H frequency. Any system that is equipped with a broadband amplifier and that can demodulate both sodium and proton frequencies has the basic hardware capability to do SMI through interleaving proton and non-proton pulses and acquisition²⁴. Even when a system has dual resonant ¹H/²³Na coils and multi-nuclear MR capability, SMI is still uncommon because: 1) non-proton imaging or spectroscopy on most human research systems is enabled by a switch that limits transmission to a single RF amplifier at any given time; and, 2) human research systems lack appropriate software and/or demodulation hardware that enables reception of two frequencies in a sequence. Our scanner is hard-coded such that it will not allow reception of two frequencies within the same sequence. However, the ability to *transmit* at two frequencies within a sequence by sequentially switching between RF transmission with the ¹H RF amplifier and broadband RF amplifier, which is commonly done during non-proton spectroscopy, allows us to add additional receive hardware to acquire synchronous data by manipulating both ²³Na and ¹H spin systems within the same pulse sequence.

Previous implementations of SMI have demonstrated excitation, readout, and encoding occurring for two nuclei at exactly the same time (“simultaneous”)^{15,18} or interleaved with rapid succession of ¹H and the non-proton steps (“interleaved”)^{14,19,21,23,24}. One implementation demonstrated both interleaved and simultaneous imaging with the use of additional hardware²². Many of the SMI techniques used *separate* ¹H and ²³Na demodulation hardware channels to acquire the ¹H/²³Na signals at each respective frequency either internal to the scanner or through additional custom transmit/receive hardware. In

order to avoid the need for two frequency demodulation channels, which require either completely custom acquisition software/hardware or internal modification of the MRI machine, our method down-converts the ^1H signal to the ^{23}Na frequency to acquire both nuclei on the system's unmodified hardware. Our novel SMI technique uses our MRI system's ability to do sequential $^1\text{H}/^{23}\text{Na}$ excitation, and enables dual-nuclear simultaneous receive by down-converting the proton signal to ^{23}Na frequency during $^1\text{H}/^{23}\text{Na}$ readout.

In vivo $^1\text{H}/^{23}\text{Na}$ -SMI has been reported in the brain¹⁴, breast¹⁶, and knee^{15,17}, with Cartesian gradient echo (GRE) imaging for both nuclei, which is inefficient for obtaining ^{23}Na images that have a short T_2^* (~15 ms). de Bruin has also shown SMI with radial sampling using a software-based interleaved imaging approach¹⁷. In this manuscript, we present a novel technique for SMI to obtain synchronous $^1\text{H}/^{23}\text{Na}$ -GRE images with Cartesian sampling. We have also performed in vivo $^{23}\text{Na}/^1\text{H}$ -SMI with a 3D GRE sequence and simultaneous radial sampling, which greatly improves ^{23}Na image quality by allowing for short TRs and sampling ^{23}Na before significant T_2^* signal loss²⁵.

METHODS

Hardware

Custom add-on hardware shown in Figs. 1&2 was developed and used with a 3T clinical MRI system (Tim-Trio, Siemens Healthcare, Erlangen, Germany), equipped with multinuclear capability, to sample both ^{23}Na and ^1H MR signals within the same acquisition at the same carrier frequency. The stock multinuclear option equips the MRI system with an additional broadband RF amplifier, and a switch box that can switch between the ^1H and broadband RF amplifiers. The broadband RF amplifier is used for non-proton RF transmission and the main baseband ^1H RF amplifier is used for proton RF transmission. The baseband and broadband signals are combined into a single transmit channel that is supplied to the RF coil.

Transmit switching for both ^1H and ^{23}Na channels uses a custom dual-resonant TR switch with passive traps to separate the ^1H and ^{23}Na transmit frequencies²⁶. The TR switch contains two connectors for transmit/receive on both ^1H and ^{23}Na channels for standard non-synchronous single-frequency operation.

Because the dual-resonant TR switch is also intended for single-nuclear imaging of either nuclei, an additional ^1H TR switch is placed between the ^1H output of the dual-resonant TR switch (Fig. 1a) and coil. The ^1H TR switch is to direct the ^1H NMR signal toward a second ^{23}Na receive channel, down-convert and receive the ^1H signal at the carrier frequency of 32.6 MHz (the same frequency as ^{23}Na) through the scanner, without modifying the current system hardware. The ^1H TR switch incorporates standard TR switch circuitry, such as a serial PIN diode, quarter-wave cables that are connected by PIN diodes to ground for preamplifier protection, and an RF preamplifier (Fig. 1b). The host MRI system provides DC biasing that is software controlled to switch the ^1H PIN diodes. During ^1H transmission, a forward-bias current (+ 100mA) shorts the PIN diodes in the ^1H TR switch (Fig. 1b) so that the ^1H RF transmit can easily pass through, but is negatively biased (– 30V) during RF reception to ensure that the receive signal is directed to the ^1H preamplifier (Fig. 1b). The

inductance of the RF chokes (L in Fig. 1b) used to pass DC bias is 15 μH ; the capacitor values (C in Fig 1b) to block the DC bias are 15 nF.

During ^1H reception, the ^1H signal is, (i) amplified by a 24 dB preamplifier (Advanced Receiver Research, Burlington, CT, USA), (ii) converted to 32.6 MHz using an RF mixer (ZAD-1, Mini-Circuit Inc., Brooklyn, NY) by mixing a 90.6 MHz sine wave from a precision frequency synthesizer (SG382, Stanford Research Systems, Sunnyvale, CA), which is synchronized using a 10 MHz clock signal from the host MRI system, and, (iii) filtered using a low-pass filter to pass the 32.6 MHz signal only (Fig. 1a). The frequency of the synthesizer is determined by subtracting the ^{23}Na frequency from the ^1H frequency, which is explained in Appendix A. The resultant ^1H signal at the carrier frequency of 32.6 MHz is then (iv) fed into one of the ^{32}Na receive channels²⁰ of the MRI system.

Two custom-made $^1\text{H}/^{23}\text{Na}$ coils were used due to imaging samples with different dimensions. The first coil consists of a linear ^{23}Na loop concentric with a linear ^1H loop²⁶ (Fig. 2a) on a 133 mm diameter tube, with the loops decoupled from each other using passive resonant traps (Fig. 2a). A second coil designed for knee imaging consists of a quadrature ^{23}Na coil (a central circular loop that is spatially decoupled with an overlapped butterfly loop) and ^1H linear coil (a butterfly loop spatially decoupled with the ^{23}Na loops) (Fig. 2b).

Pulse Sequence

A dual-nuclei MRI pulse sequence (Fig. 3) was developed to manipulate both spin systems within the same pulse sequence, using the manufacturer's pulse sequence development software (IDEA, Siemens Medical Solution, Erlangen, Germany). Sodium is selected as the main measurement nucleus and ^1H as the second transmit nucleus.

Synchronous $^1\text{H}/^{23}\text{Na}$ imaging was performed with a standard gradient-echo sequence (GRE) for ^{23}Na with both ^1H and ^{23}Na transmit and receive occurring in the same sequence (Fig. 3). RF and gradient pulses for ^1H imaging are added prior to the ^{23}Na excitation RF pulse, so that the pulse sequence for ^{23}Na imaging is not altered. The zeroth moment of the gradient pulses for ^{23}Na slice-selection is compensated in the refocusing gradient of the ^1H slice-selection. Sampling for *both* ^1H and ^{23}Na at the carrier frequency of 32.6 MHz was acquired simultaneously on *two* ^{23}Na channels.

The pulse sequence in Fig. 3 had different k-space trajectories for either Cartesian or radial synchronous acquisition. During Cartesian acquisition, the sequence acquired a phase reference echo that is used to correct phase errors on the ^1H Cartesian FIDs, as detailed later.

MRI Experiments

SMI was implemented for 3D Cartesian and 3D ultra-short TE (UTE) radial $^1\text{H}/^{23}\text{Na}$ -GRE. The Cartesian studies were performed to test SMI feasibility and SNR changes with a phantom. The radial studies tested in vivo image quality on a healthy human knee. The acquisition parameters are listed in Table I.

The Cartesian phantom SMI studies were conducted with the linear and quadrature coil (Fig. 2) on a phantom that contains three cylindrical tubes with different concentrations of MnCl_2 and NaCl (0.4 mM MnCl_2 and 300 mM NaCl , 0.2 mM MnCl_2 and 150 mM NaCl , 0.1 mM MnCl_2 and 75 mM NaCl). Air surrounded the tubes when the quadrature coil was used, while liquid surrounded the tubes containing 0.2 mM MnCl_2 and no NaCl during the experiment with the linear coil. The phantom study parameters are listed in Table IA–D.

Three sets of a healthy human knee images were acquired with 3D UTE radial sampling: a single-nuclear ^1H acquisition, a single-nuclear ^{23}Na acquisition, and a synchronous $^1\text{H}/^{23}\text{Na}$ acquisition. Radial SMI data was acquired with the quadrature coil (Fig. 2b) that conformed to the knee geometry. Imaging was conducted after informed consent and with approval of the local institutional review board.

For both Cartesian and radial studies, identical acquisition parameters (Table IA–C, IE–G) for each nucleus were used to compare single-nuclear imaging (SNI) and SMI image quality.

Post-processing Using Phase Reference Echoes:

Although the external synthesizer is synchronized with the host MRI system using a 10 MHz reference signal, the single stage mixing introduces a systematic and incremental phase-error into the ^1H channel data, which may be due to the small difference in the frequency resolution between the scanner's synthesizer and the mixer's synthesizer. These small differences can be introduced during separate scan sessions when the sampling frequencies may occur after scanner calibration. For Cartesian imaging, this systematic phase-error is corrected using the ^1H phase reference echoes after gradient rewinding (G_{rw} in Figure 3), which are simultaneously acquired with each ^{23}Na FID. Note that the ^{23}Na phase reference echo is not useful for phase error correction because of its low SNR.

In order to correct this phase offset, the ^1H image data is transferred to and processed on an offline computer. For each ^1H imaging echo, the systematic phase error, θ , is estimated at the peak of the ^1H phase reference echo during Cartesian imaging or by the first data point of each radial line, and subtracted from the ^1H imaging echo by multiplying by $e^{-i\theta}$.^{27,28} Offline reconstruction of the ^1H radial data was done with a nearest-neighbor regridding algorithm, programmed using the Numpy package in Python programming language (Version 2.7, Python Software Foundation, <https://www.python.org/>). Nearest-neighbor was used due to the efficiency of implementation.

The single-nuclear ^1H data was reconstructed using the same off-line reconstruction software for accurate comparison with the SMI ^1H images. The single-nuclear ^1H images were also constructed with and without phase correction to verify that the phase-correction process did not adversely affect ^1H image quality. H image quality.

The ^{23}Na data is reconstructed by the scanner for all images, as the additional ^{23}Na hardware does not affect the scanner's ability to reconstruct ^{23}Na SNI or SMI radial data. The percent SNR difference is reported as the absolute value of the phase corrected image subtracted from the uncorrected image, all divided by the uncorrected image, and multiplied

by 100. The SNRs were normalized by the noise along a similar region of image-space in registered images.

RESULTS

Synchronous ^1H and ^{23}Na images were obtained in both a phantom and in vivo (Figs. 4–7).

Phantom Studies

Phase correction—The magnitude of a single-nuclear proton acquisition without and with phase correction is shown in Figs. 4(a) and 4(b), respectively. The percent difference between the single-nuclear data with and without phase correction was less than 10^{-9} , which is near the precision error of the computer.

Phantom Proton Results—The proton data acquired synchronously with the sodium data without and with phase correction is shown in Figs. 4(c) and 4(d), respectively. Only the first average is shown in Fig. 4(c), as all ten averages would create a less coherent image. The proton image has significant artifact Fig. 4(c) that is removed with phase correction as shown in Fig. 4(d).

The synchronous data was shifted by 11 voxels in the readout direction and by 5 pixels in the phase-encode direction when compared to the single-nuclear data. After adjusting for image registration, the percent SNR difference, shown in Fig. 4(e), was $9 \pm 3\%$ when measured over a similar region in either of the two brightest phantoms.

The data here demonstrates the effects of phase correction and the SNR difference using the different methods, although the proton images in Fig. 4 show susceptibility artifact.

Phantom Sodium Results—The sodium images acquired using both single-nuclear and synchronous acquisitions are shown in Figs. 5(a) and 5(b), respectively. The SNR difference is shown in Fig. 5(c), which is $6 \pm 3\%$ as measured in the phantom.

The image shown in 5(b) was acquired in the same acquisition as the image shown in 4(d), demonstrating the ability to acquire both proton and sodium data together for an overall reduced scan time. The full FOV is shown in Fig. 5(b), which is 3.8 times larger than the proton FOV. Because the ^1H and ^{23}Na signals are acquired using the same gradients for readout and phase-encoding, the ^1H image data has a 3.8 times higher resolution and smaller FOV due to the ratio of the ^1H and ^{23}Na gyromagnetic ratios ($= \gamma_{1\text{H}} / \gamma_{23\text{Na}}$).

A much better phantom image is shown in Fig. 6, which sampled k-space more fully and had MnCl_2 surrounding the tubes to reduce susceptibility artifact. This image was not acquired with the same SNR comparisons. The full FOV is shown in the ^1H images in all planes in Fig. 6, while the sodium images were interpolated and reduced to a similar FOV.

In Vivo Human Knee Study—Fig. 7 shows images from single-nuclear and synchronous $^1\text{H}/^{23}\text{Na}$ multi-nuclear radial acquisitions in a healthy human knee. The single-nuclear ^1H and ^{23}Na images were acquired in two 7-minute acquisitions, respectively, for a total

acquisition time of 14 minutes. The synchronous $^1\text{H}/^{23}\text{Na}$ images were acquired in a single 7-minute acquisition, i.e., a total acquisition time of 7 minutes.

There were $< 0.1\%$ SNR differences between the ^1H -SNI acquisition with and without phase correction, indicating that using phase correction of radial data by subtracting each k-space readout from its first data point remains accurate.

The SNR was similar for the synchronous and single-nuclear ^{23}Na acquisitions. The ^1H synchronous in vivo images had a $\sim 6\%$ SNR decrease when compared to the single-nuclear ^1H images.

Both SNI and SMI ^1H show the plastic structure that the coils were mounted on as a halo around the object. There is undersampling artifact in the lower portion of the knee image (Figs. 7(a,c)).

DISCUSSION

The overall result of this study is that *both* ^1H and ^{23}Na images were acquired in a *single* acquisition, demonstrating a novel method to acquire MRI/NMR data from an additional second nucleus without an increase of acquisition time, with minor to no loss ($\sim 6\%$) in ^{23}Na -SNR and minor losses ($\sim 9\%$) in ^1H -SNR. Without modification such as the method we presented, SMI would not have been possible on our system. This is similar to an approach used by Meyerspeer²² This study demonstrates a method for the substantial reduction of acquisition time for acquiring image data from both nuclear species. Typical single-nuclear acquisitions of both ^1H and ^{23}Na would require twice the acquisition time for two sequential acquisitions. As one of the major limitations of in vivo ^{23}Na -MRI is its long acquisition time, SMI may greatly improve the practical applications of ^{23}Na -MRI^{3–11,13,17,22}, without requiring additional acquisition time. By acquiring both ^1H and ^{23}Na within the same readout, TRs can be kept low for efficient imaging. With future developments, SMI could make ^{23}Na more attractive for clinical imaging by allowing it to be acquired simultaneously without impacting imaging times.

Our solution for SMI was performed without modifying the internal components of our MRI scanner by adding hardware to our non-proton setup, requiring an additional TR switch, a mixer, a frequency synthesizer, and a synchronous multi-nuclear pulse sequence. This work did not involve acquiring or transmitting both the $^1\text{H}/^{23}\text{Na}$ signals simultaneously, and therefore did not require significant scanner modifications required by previous work¹⁸. Our method relies on the unmodified scanner hardware for reception of both nuclear species, by down-converting and receiving the ^1H signal through a second sodium channel. Interleaved methods that use software-defined switching of the RF amplifier and receiver frequencies were not possible here, so a modification to commercial hardware was required. This method could be permanently implemented within the system by modifying RF receive channel pathways. In this respect our implementation of SMI was different from previously described setups that modified internal system components^{14–16,18–21,23}.

The introduction of SMI resulted in minor losses to the ^1H -SNR ($\sim 9\%$) and minor losses ($\sim 6\%$) to ^{23}Na -SNR. The ^{23}Na signal pathways and sequence did not change, even with the

introduction of SMI. Sufficient SNR was obtained to illustrate the benefits of obtaining simultaneous data from two nuclei in the same sequence, reducing the need for two separate imaging sessions. We attribute the minor losses on the ^1H images during SMI (~10%) due to using a separate add-on preamplifier, from the possible introduction of noise in the mixing process, and from the very minor change in k-space acquisition. This work suggests that many ^1H sequences could implement $^1\text{H}/^{23}\text{Na}$ -SMI without degradation of ^1H imaging ability and obtain ^{23}Na imaging data without image acquisition time cost.

We demonstrated that correcting data with a phase reference acquisition at the center of k-space allows the data to be reconstructed with near zero errors in a single-nuclear phantom experiment.

We demonstrated SMI with two k-space trajectories using similar GRE pulses, for an SNR comparison using Cartesian acquisition and for efficient ^{23}Na sampling for in vivo imaging. This is a demonstration of ^{23}Na SMI with truly simultaneous radial acquisition for obtaining high quality ^{23}Na images.

Completely interleaved imaging allows more flexible sequences, such that the proton sequences could be nearly completely independent from the sodium sequences. However, truly simultaneous imaging offers the ability to reduce the readout time and TRs due to combined sampling.

As a result of *simultaneous* $^1\text{H}/^{23}\text{Na}$ acquisition, the ^{23}Na -FOV is 3.8 times larger than the ^1H -FOV, and the ^1H voxel size is 3.8 times smaller than the ^{23}Na resolution. Currently, our MRI system limits our FOV to a maximum of 500 mm, which means that the largest ^1H -FOV that can be acquired without aliasing is 130 mm during down-sampled SMI. This FOV is sufficient for imaging moderate sized knees and breasts, or imaging with FOV-limiting local coils, but would require implementation of ^1H slice selection or increased oversampling for imaging larger FOVs. This restriction is primarily software-related, as standard MRI systems have a much broader range of selectable FOVs that make use of higher gradient strengths than we used here. However, the difference in FOVs can be very practical and useful, as ^1H imaging occurs at a typically higher resolution than ^{23}Na imaging due to ^{23}Na -SNR being much lower than the ^1H -SNR. Increasing the ^{23}Na -FOV by a factor of four doubles ^{23}Na -SNR compared to the original FOV. The constraint on the ratio of ^{23}Na to ^1H FOV and resolution thus fits very naturally into the requirement for ^{23}Na imaging to increase its SNR through use of larger voxel sizes than ^1H imaging.

Although the current work has demonstrated flexibility in the selection of imaging parameters, some parameters must be identical using this setup, such as receiver bandwidth and FOV-ratio, because both NMR signals are simultaneously sampled within the same acquisition window. The FOVs can be completely independent by applying additional gradients prior to ^{23}Na excitation, which adds or subtracts the phase-encoding gradient appropriate for the ^1H -FOV. In addition, by using a ^1H pulse before the ^{23}Na pulse, accurate slice selection of the ^1H data would be possible that could not occur with simultaneous $^1\text{H}/^{23}\text{Na}$ pulses. The restriction in other imaging parameters may be completely removed by interleaving the acquisitions between two nuclei with a minor cost in time efficiency.

In our current implementation, we use two of the ^{23}Na receive channels of our MRI system. One channel receives the ^{23}Na signal directly from a ^{23}Na coil, while a second channel receives the ^1H signal mixed down to the ^{23}Na frequency. The SNR of the ^{23}Na images can also be improved by the use of a ^{23}Na phased array coil, as demonstrated in²⁶, without any change to the SMI system. B1 maps of the coil for accurate ^{23}Na quantitation could be acquired by altering the ^{23}Na flip angle²⁹ during SMI, without changing the ^1H flip angle. It is also straightforward to acquire $^1\text{H}/^{23}\text{Na}$ -SMI data with multiple ^1H channels, by duplicating the conversion box including the switching circuitry in Fig 1b.

The current mixing scheme for ^1H -data in the receive data caused a consistent non-random phase error along the increasing FID numbers. This systematic phase-error was small as the synthesizer frequency is set to match the down-converted ^1H signal to the carrier frequency. The systematic phase-error caused by down-conversion was straightforward to estimate and correct using the ^1H phase reference signal during the post-processing. A higher performance synthesizer would result in reduced phase errors but would result in additional cost.

During the lengthy acquisitions that often occur for ^{23}Na imaging, the subject may move, reducing the quality of ^{23}Na data. In SMI, the ^1H phase reference FID could be used to correct the phase error in ^{23}Na FIDs that may be induced by motion, which is not possible using ^{23}Na FIDs alone due to ^{23}Na 's low intrinsic SNR. $^1\text{H}/^{23}\text{Na}$ -FID data with severe motion corruption subject's motion can be identified using the ^1H -FID and excluded from processing. By using this ^1H phase reference echo in real-time self-gated ^{23}Na -MRI, the phase and magnitude of the phase reference echo peak could be calculated and then compared to those of the reference echoes to determine if the ^{23}Na and ^1H imaging echoes are corrupted by the subject's motion, which then would be rejected and reacquired³⁰. This method requires accurate estimation of the systematic phase increment. The specific absorption rate (SAR) was calculated automatically by the scanner, using predetermined values in the coil-control files. The sequences used in this study were not pulse intensive and so did not result in SAR limitations of TR or flip angle. SMI implementations with other types of sequences (eg. turbo-spin echo) would have to consider SAR limitations more fully.

Our synchronous technique could be extended to study the NMR characteristics of other nuclei (^{13}C , ^{19}F , ^{31}P), as well as extended to triple-nuclear imaging²¹ by converting and receiving multiple frequencies to a single frequency, provided that the scanner allows transmit on more than two frequencies in a sequence.

CONCLUSIONS

We have demonstrated synchronous ^1H and ^{23}Na MRI using an unmodified commercial scanner with multinuclear capability and additional hardware. We have shown in vivo synchronous acquisition of ^1H and ^{23}Na images of the knee in a 7 minute scan time, which is nearly equivalent to acquiring ^1H and ^{23}Na images with two separate conventional single-nuclear 7 minute scans. We have demonstrated the acquisition of synchronous ^1H and ^{23}Na GRE images with simultaneous Cartesian and simultaneous radial sampling, with a low and

equal TR for both sodium and proton acquisitions. Our method allows the addition of ^{23}Na imaging to conventional ^1H imaging with little or no increase in total scan time.

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APPENDIX

A. Phase relationship during mixing

The mixer acts as a frequency multiplier between the ^1H MR signal

$S_H(t) = S_{Ho} e^{-t/T_2^*} \cos(\omega_H t + \theta_H(t))$ with a time-varying phase $\theta_H(t)$ and the synthesizer signal $S_S(t) = S_{So} \cos(\omega_S t + \theta_{So})$ with an initial phase θ_{So} :

$$\begin{aligned} S_H(t) \cdot S_S(t) &= \\ &= S_{Ho} S_{So} e^{-t/T_2^*} \cos(\omega_H t + \theta_H(t)) \cos(\omega_S t + \theta_{So}) \\ &= \frac{1}{2} S_{Ho} S_{So} e^{-t/T_2^*} \left(\cos((\omega_H + \omega_S)t + \theta_H(t) + \theta_{So}) + \cos((\omega_H - \omega_S)t + \theta_H(t) - \theta_{So}) \right) \end{aligned} \quad (1)$$

After low-pass filtering, the converted signal becomes,

$$S'_H(t) = \frac{1}{2} S_{Ho} S_{So} e^{-t/T_2^*} \cos((\omega_H - \omega_S)t + \theta_H(t) - \theta_{So}) \quad (2)$$

Equation (1) indicates that the new mixed frequency will be

$$\omega'_H = \omega_{Na} = \omega_H - \omega_S. \quad (3)$$

Equation (2) also indicates that *the converted ^1H signal retains the same phase $\theta_H(t)$* as in the original ^1H signal added to the synthesizer phase θ_{So} , and the ^1H signal is also modulated by the amplitude of the reference signal.

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Fig. 1.

(a) Transmit/receive hardware for both ^1H and ^{23}Na . Arrows indicate transmit / receive pathways. The host system provided both ^1H and ^{23}Na transmit pulses through a single channel, which passed through a commercial $^1\text{H}/^{23}\text{Na}$ TR switch. The ^{23}Na coil connected directly to the $^1\text{H}/^{23}\text{Na}$ TR switch; the ^1H channel runs through a second ^1H TR switch before connecting to the coil. The second ^1H TR switch was necessary to redirect the ^1H RF for mixing and down-conversion to 32MHz for simultaneous $^1\text{H}/^{23}\text{Na}$ reception. An attenuator is necessary due to the significant signal difference between ^{23}Na and ^1H . (b) TR switch schematic for ^1H . The ^1H signal is amplified at 123MHz before down-conversion to 32MHz.

Fig. 2.
(a) Linear sodium coil and linear proton coil concentric with each other and decoupled using traps. (b) Quadrature sodium coil (central circular loop and butterfly loop) and linear proton coil, each overlapped to minimize mutual inductive coupling.

Fig. 3.

Pulse sequence diagram for synchronous GRE imaging of ^{23}Na and ^1H . Although the two frequencies share gradients, they are received on separate channels. RF and gradient pulses for ^1H imaging are applied prior to the ^{23}Na excitation RF pulse. All phase-encoding gradients are rewound (G_{rw}) just before the acquisition of the phase reference echo. The ^1H gradients before the ^{23}Na pulse do not affect the ^{23}Na signal, so that the ^1H slice and phase encodes can be independent from ^{23}Na slice and phase encodes. The method shown was used for both Cartesian and radial sampling. The phase reference pulse did not occur during GRE imaging, as there was sufficient ^1H signal to enable phase correction without a reference pulse.

Fig. 4.

(a,b) Single-nuclear and (c,d) synchronously acquired proton images using Cartesian GRE reconstructed (a,c) without and (b,d) with phase correction from the reference echo. The images in (c,d) were obtained synchronously with sodium images in Fig. 5. (e) The SNR difference is shown between a registered single-nuclear image and phase-corrected synchronous image shows less than 10% SNR difference in the phantoms.

Fig. 5.

(a) Single-nuclear and (b) synchronously acquired sodium images using Cartesian GRE. The sodium image in (b) was obtained synchronously with proton images in Fig. 4. (c) The SNR difference is shown between the shifted, registered single-nuclear image and the phase-corrected synchronous data shows less than 6% SNR difference in the phantoms. The full reconstructed sodium FOV is shown here, which is 3.8 times smaller than the proton FOV.

Fig. 6.
(a-c) Proton and (d-f) sodium SMI images acquired during a single synchronous 21-minute 3D GRE Cartesian acquisition. Three orthogonal slices from a full 3D data set are shown for both (a-c) proton and (d-f) sodium images. Images are shown at the same physical scale showing the entire ^1H FOV and $\frac{1}{4}$ of the ^{23}Na FOV.

Fig. 7.

Single-nuclear (a,b) and synchronous multinuclear (c,d) knee images of a healthy volunteer. (a) and (b) were obtained in two 7 minute acquisitions for a total of a 14 minute scan time to acquire both nuclear images. (c) and (d) were obtained in a single 7 minute acquisition to obtain both nuclear images. The structure at the right of the proton images (a,c) is a plastic coil element that is visible because of the ultra-short TE acquisition.

TABLE I**Acquisition Parameters**

Column Name	A	B	C	D	E	F	G
Figure	4a,b	5a	4(c,)d,5b	4,5,6a–f	7a	7b	7c,d
Sample	Phantom	Phantom	Phantom	Phantom w/ MnCl	Knee	Knee	Knee
Nuclei	^1H	^{23}Na	$^1\text{H}/^{23}\text{Na}$	$^1\text{H}/^{23}\text{Na}$	^1H	^{23}Na	$^1\text{H}/^{23}\text{Na}$
Sampling	Cartesian	Cartesian	Cartesian	Cartesian	Radial	Radial	Radial
Coil	Quad	Quad	Quad	Linear	Quad	Quad	Quad
TR (ms)	50	50	50/50	50/50	40	40	40/40
TE (ms)	9.51	2.45	9.51/2.45	7.27/2.25	0.70	0.27	0.70/0.27
Flip Angle	25	70	25/70	25°/70°	25°	70°	25°/70°
FOV (mm)	125	500	125/500	118/448	105	400	105/400
Voxel Size (mm ³)	0.98 iso	3.9 iso	0.98/3.9 iso	0.9/3.5 iso	0.8 iso	3.1 iso	0.8/3.1 iso
Averages	10	10	(1) 10	5/5	1	1	1
Acq. Matrix	256×128×16	256×128×16	256×128×16	256×128×40	256×10240	256×10240	256×10240
Image Time	17 min	17 min	17 min	21 min	7 min	7 min	7 min
Readout Sampling	0.52	0.52	0.52	6/8	-	-	-

Acquisition parameters for the single and synchronous dual-nuclear GRE images shown in this paper. When two values are listed in a box, the first parameter is for the ^1H parameters and the second for ^{23}Na parameters in the pulse sequence.