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Characterization of a Low-Profile, Flexible, and Acoustically Transparent Receive-Only MRI Coil Array for High Sensitivity MR-Guided Focused Ultrasound

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ABSTRACT Magnetic resonance guided focused ultrasound (MRgFUS) is a non-invasive therapeutic modality for neurodegenerative diseases that employs real-time imaging and thermometry monitoring of targeted regions. MRI is used in guidance of ultrasound treatment; however, the MR image quality in current clinical applications is poor when using the vendor built-in body coil. We present an 8-channel, ultra-thin, flexible, and acoustically transparent receive-only head coil design (FUS-Flex) to improve the signal-to-noise ratio (SNR) and thus the quality of MR images during MRgFUS procedures. Acoustic simulations/experiments exhibit transparency of the FUS-Flex coil as high as 97% at 650 kHz. Electromagnetic simulations show a SNR increase of 13× over the body coil. In vivo results show an increase of the SNR over the body coil by a factor of 7.3 with 2× acceleration (equivalent to 11× without acceleration) in the brain of a healthy volunteer, which agrees well with simulation. These preliminary results show that the use of a FUS-Flex coil in MRgFUS surgery can increase MR image quality, which could yield improved focal precision, real-time intraprocedural anatomical imaging, and real-time 3D thermometry mapping.

INDEX TERMS Magnetic resonance imaging, coils, ultrasonic transducer arrays.

I. INTRODUCTION

Magnetic resonance guided focused ultrasound (MRgFUS) has emerged as a non-invasive treatment modality in a number of applications, such as essential tremor [1]–[3], Parkinson's disease [4]–[6], neuropathy [7], [8], epilepsy [9], blood-brain barrier opening [10]–[13], and Alzheimer's disease [14]–[16].

MRgFUS systems use helmet-shaped transceivers with a large number of ultrasound (US) transducers (for instance, the

INSIGHTEC ExAblate system comprises 1024 transducers) concentrating acoustic energy on a millimetric-sized focal point in the brain. In order to efficiently couple acoustic energy, a degassed water bath is placed between the ultrasound transducer and the skull. This water bath also serves as a cooling mechanism. The frequency and intensity of the acoustic energy can vary (220 kHz - 720 kHz) depending on the application.

To localize the sonication target, structural MRI is used [17], [18]. MR thermometry [19]–[22] is employed to monitor temperature/energy delivery in the target and healthy tissue during intervention. Furthermore, diffusion tensor

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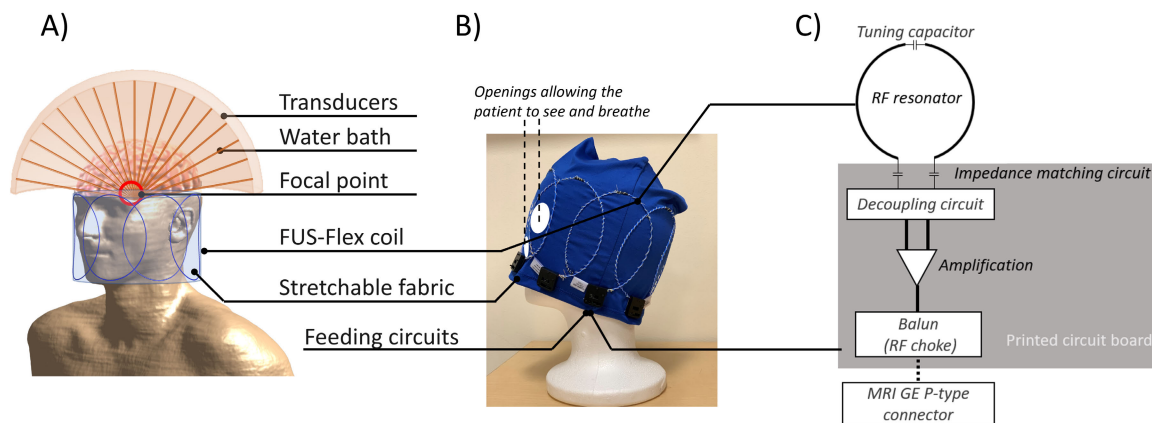


FIGURE 1. A) Drawing showing the FUS-Flex coil and transducer placed around a human head model. B) Photograph of the 8-channel FUS-Flex coil. C) Schematic of one RF resonator along with the main components of the feeding circuit.

imaging (DTI) aids the selection of ablation sites in preprocedural and intraprocedural planning [23].

However, poor imaging quality in many current MRgFUS exams precludes effective and fast image acquisition. First, a typical birdcage-like head receive coil cannot be used to achieve signal-to-noise-ratio (SNR) typically observed in MRI because the transducer does not leave adequate space. As a result, most MRgFUS techniques currently use the much larger and less efficient, vendor built-in, body-sized coil for both transmission and reception. Second, the high-permittivity water bath, together with the conductive transducer surface, causes significant B_1 inhomogeneities that produce the unwanted low-signal band artifacts [24] observed in MRgFUS images at the region of interest. This artifact tends to occur at the locations of the thalamus and hippocampus [25], [26], which are regions of interest for essential tremor and Alzheimer's disease.

Different receive coil arrays have been designed in order to achieve better SNR [20], [24], [27]–[33]. Bitton *et al.* proposed a 3T dual-channel receive coil integrated into the MRgFUS silicone sealant membrane [32]. The upper portion of the coil is submerged in the water bath, while the lower part sits outside, providing a SNR increase by a factor of 4 compared to the body coil. Watkins *et al.* proposed a volume coil design for 3T MRI that can be placed partially inside the water-filled transducer. This interior portion of the coil is inductively coupled to the portion of the coil that is located outside the transducer [28].

However, the evaluation of the acoustic footprint has not been tackled in great detail in MRgFUS related coil design, with the exception of the work of Corea *et al.*, in which the printed capacitor-based coil design exhibits experimentally evaluated, acoustic, shoot-through, transparency of up to 89.5% at 650 kHz and 80.5% at 1 MHz, allowing the coil to be placed in the acoustic path [30]. Phantom results show an increase in MRI SNR by a factor of 2 at the center of the phantom using a 4-channel printed receive coil.

In this paper, we aim to improve both imaging sensitivity and acoustic transparency in one apparatus by presenting a very thin, low-profile, receive-only 8-channel head coil (FUS-Flex) operating at 3T. The design is inspired by stretchable [34], [35], flexible [36]–[40] and lightweight [41] coil technologies, offering a coil array with full conformity to the head. The novelty of our work lies in the use of very thin (~ 1 mm) RF elements (providing low interaction with the acoustic field), and the use of higher channel count than currently available in the literature, increasing the available imaging SNR, the sensitivity of the coil and improving/enabling parallel imaging. Better receive SNR in the region of the low-signal band artifact can also indirectly reduce the associated sensitivity problems.

II. METHODS

A. COIL GEOMETRY

The proposed receive-only FUS-Flex coil consists of an 8-channel array using receive architecture inspired by highly flexible and thin coil technology [42], [43]. Each element has a diameter of 110 mm. The coil is designed to be placed conformally, and in a close-fitting fashion, around the circumference of the patient's head (Figure 1). The RF elements consist of a thin malleable conductor construction [36], [39], [42]–[46] comprising two parallel conductor wires encapsulated and separated by a dielectric material, the two parallel conductor wires maintained separate by the dielectric material along the entire length of the loop portion between terminating ends thereof (INCA, integrated distributed capacitors - thickness = 0.6 mm) with a polytetrafluoroethylene (PTFE) jacket (outer diameter ~ 1 mm) (GE Healthcare, Waukesha, WI, USA). The RF element is created from a flexible link resonator structure with the length of each resonator being no greater than 1/10th of the wavelength of the resonant RF field [47]. This design ensures tuning stability when loaded due to uniform charge distribution and internally confined irrotational electric fields

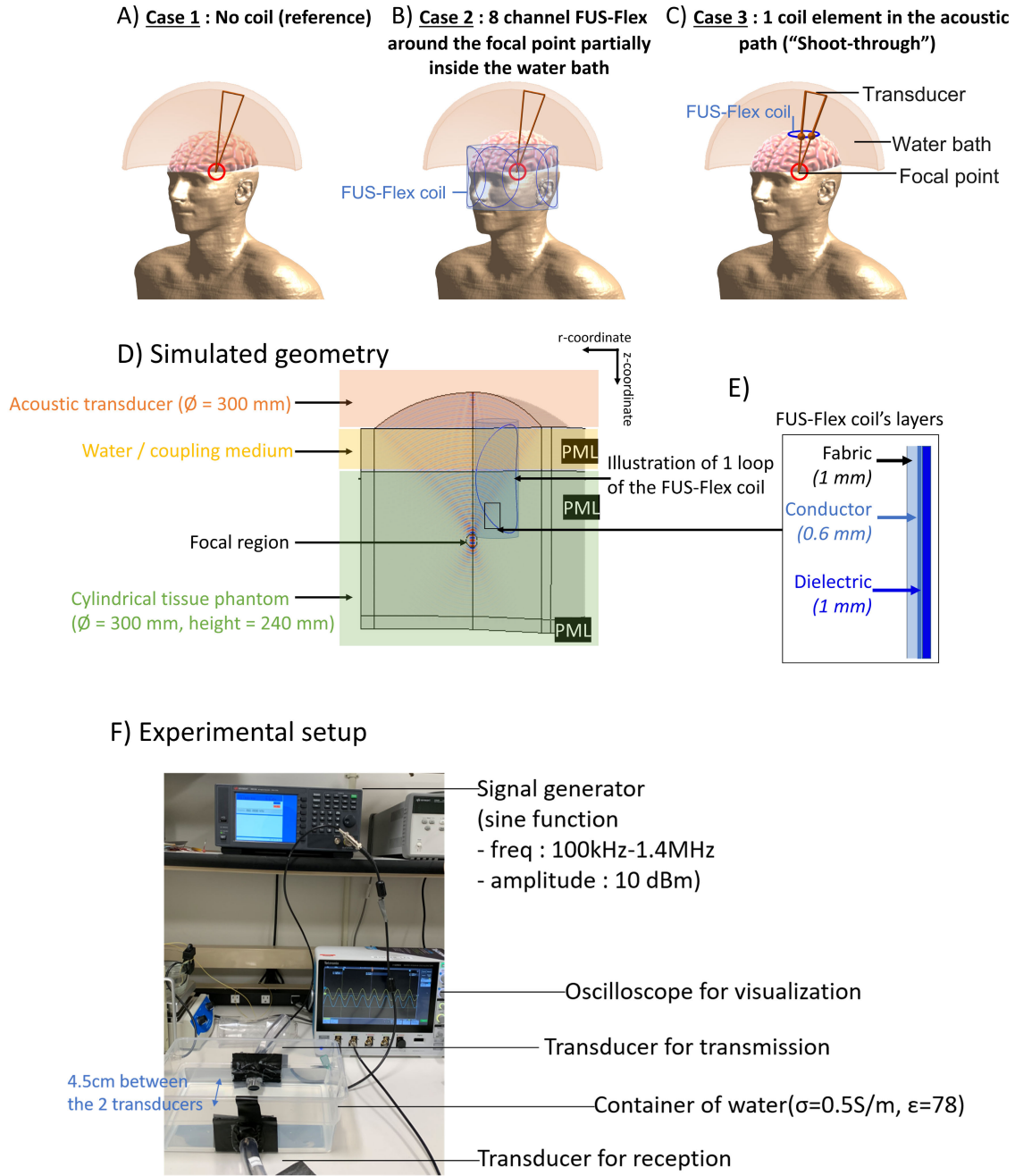


FIGURE 2. Illustration of the transducer A) without a coil (case 1); B) with an 8-channel FUS-Flex coil placed around the focal point (case 2); and C) 1 channel FUS-Flex coil "shoot-through" (case 3). The 3 cases were simulated using a cylindrical phantom to mimic tissue. D) 3D simulation model with cylindrical phantom. The 30cm transducer is represented by the top dome in orange. The cylindrical phantom used is shown in green. The black lines at the exterior of the phantom/water represent perfectly matched layers used to absorb outgoing waves. Different orientations/positions of the coil (blue line, shown enlarged for better illustration, not to scale) were simulated as illustrated in Figure 2A-C. E) A magnified view of the different layers of the FUS-Flex coil. F) Experimental bench setup to measure the acoustic attenuation incurred due to the FUS-Flex coil.

within the resonator [48]. The smaller diameter size conductor lends itself to its application in MRgFUS due to substantially decreased acoustic scattering. The conductor, whose

resistance measures 10Ω with head loading, is attached to a feedboard utilizing a custom preamplifier with a noise figure of <0.5 dB, a gain of 28 dB at 127.7 MHz, and an input

impedance of $<3 \Omega$. Coil elements were placed with a fixed overlap of 30 mm in a 2D planar configuration. The effective preamplifier decoupling impedance is sufficiently robust ($>1 \text{ k}\Omega$) to facilitate element-to-element overlap beyond that of conventional critical coupling to accommodate the conforming of the array or different head sizes [44], [45]. The conservative electric field is strictly confined within the small cross-section of the two parallel wires and dielectric filler material. In the case of two RF coil loops overlapping, the parasitic capacitance at the cross-overs is greatly reduced in comparison to two overlapped copper traces of traditional RF coils. RF coil thin cross-sections allow better magnetic decoupling and reduce or eliminate critical overlap between two loops in comparison to two traditional trace-based coil loops [44], [45]. In the RF transmit phase a hybrid decoupling scheme is utilized [44].

The array is sewn on a quasi-acoustic transparent polyester fabric often used in loudspeaker designs (shown in blue in Figure 1B) (Guilford of Maine, ME, USA). The light weight of the FUS-Flex coil and the breathability of the polyester fabric help improve patient comfort and allow patients to see and breathe normally during procedures.

B. ACOUSTIC SIMULATIONS AND EXPERIMENTS

The acoustic transparency of the FUS-Flex coil was evaluated by investigating the attenuation of the acoustic signal as well as the shift of the focal point in different coil placements using numerical simulation. To this goal, we studied the influence of the FUS-Flex coil material (conductor, dielectric, and fabric) on the acoustic focal point emitted by a 30 cm-diameter transducer. Case 1: the transducer was simulated without the RF coil present for reference (Figure 2A). Case 2: the 8-channel coil was placed around the focal point at a distance of 80 mm, mimicking the position of the coil around the patient's head (Figure 2B). Case 3: one RF element was placed directly in front of the acoustic source ("shoot-through") to study the acoustic transmission/attenuation directly through the coil and thus to quantify the attenuation from one coil element (Figure 2C). Simulations were performed using COMSOL Multiphysics® (COMSOL, Burlington, MA). Figure 2D, E show a model of a transducer (focal length 232 mm, radius 150 mm), water bath, and a cylindrically shaped tissue phantom to mimic the head (radius 150 mm, length 240 mm) [49], [50]. The thicknesses of the fabric, conductor, and coil dielectric were 1, 0.6, and 1 mm, respectively. The transducer was driven at typical low and high frequencies used in FUS treatment, i.e., 220 kHz and 650 kHz. For each case, the intensity magnitude, in W/m^2 , was plotted along the z-coordinate through the focal point. The spatial resolution used in this simulation was approximately 0.01 mm.

The acoustic attenuation of the coil was also evaluated on the bench using 2 immersion transducers (500kHz, 00-011923_NF, Sensor Networks, Inc) in a container of water as shown in Figure 2F. The acoustic transmission attenuation was measured for the FUS-Flex coil and was compared to the

INSIGHTTEC membrane that was used to seal the 2-channel coil in the study by Bitton *et al.* [32]. This membrane is often used in MRgFUS settings when an acoustically transparent sealant material is required. We therefore included it in our acoustic tests as a known reference standard. The transducers were separated by 4.5 cm, and the material under test was positioned centrally between the two transducers.

C. ELECTROMAGNETIC SIMULATIONS

We hypothesized that the proposed coil provides increased MR imaging SNR in (1) a non-MRgFUS exam compared to a conventional head coil (given its conformity and close proximity), and (2) in an MRgFUS exam in comparison to the vendor built-in body coil.

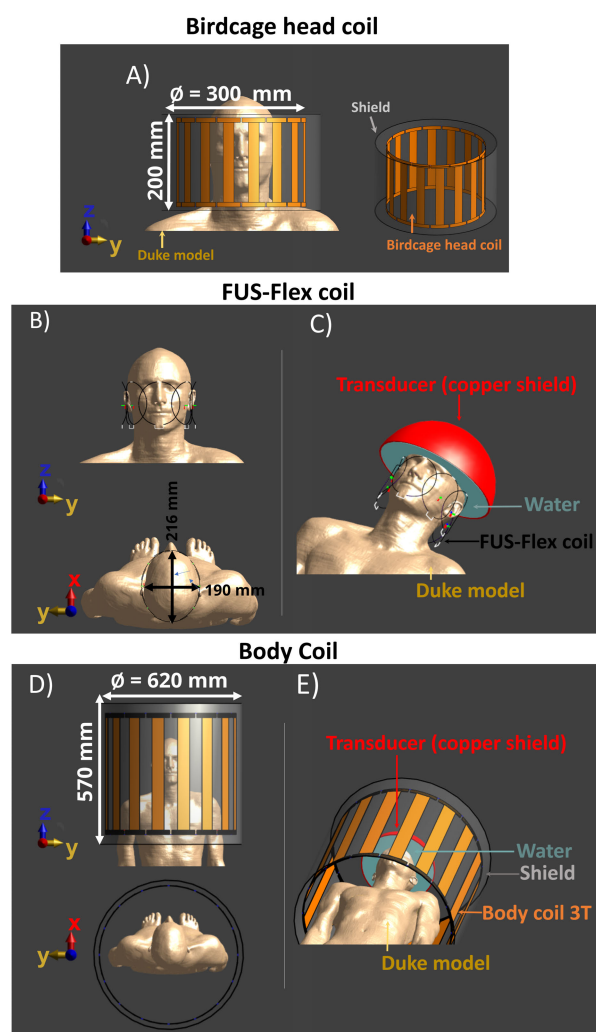


FIGURE 3. Illustrations of the head of duke in several scenarios: A) a standard birdcage head coil geometry; B) a FUS-Flex coil, and D) a body coil geometry. The transcranial focused ultrasound transducer was modeled for use with C) a FUS-Flex coil E) a body coil geometry.

Numerical simulations were performed to analyze coil performance in both applications. SNR improvement was determined using the B_1^- field magnitude.

1) COMPARISON OF FUS-FLEX COIL TO CONVENTIONAL BIRDCAGE HEAD COIL

After an MRgFUS procedure, a head coil is often used for a control scan without the transducer. Often the standard head birdcage is used. The conformal fit of the FUS-Flex coil could outperform the commercially available head coil even in a normal, non-MRgFUS exam as used at the end of an MRgFUS procedure and could also outperform a less flexible phased array due to its increased distance from the skull. To investigate on this hypothesis, electromagnetic simulations of the 8-channel receive-only FUS-Flex array using an element diameter of 110 mm were performed using Sim4Life (Zurich MedTech, Zurich, Switzerland). Its performance was compared to a 16-leg conventional birdcage head coil (diameter: 300 mm; length: 200 mm), Figure 3A, B. For a realistic *in silico* scenario, a body model, Duke (IT'IS Foundation, Zurich, Switzerland), was used. The FUS-Flex coil array was considered to be of oval shape (semi-minor axis of 190 mm, semi-major axis of 216 mm). The conductors were chosen to be perfect electric conductors (PEC). Matching and tuning capacitors were used to tune the coil elements to 128 MHz and ensure a 50 Ω -match. Each RF element was driven by a 1V gaussian excitation signal with sequential phase increments of 45 degrees. In order to provide an estimation of the SNR with the receive-only FUS-Flex coil, we plotted the rotational component of the magnetic field B_1^- .

2) FUS-FLEX COIL WITHIN ULTRASOUND TRANSDUCER AND COMPARISON TO BODY COIL

First, we replicated the low-signal bands that stem from the influence of the transducer on the transmit field by modeling an MRgFUS transducer of 30 cm diameter using a semi-spherical water-filled copper-coated geometry, placed over Duke's head (Figure 3C, E). We then evaluated the receive SNR of the proposed FUS-Flex coil and compared it to the commonly used 16-leg body coil (diameter: 620 mm; length: 570 mm) in order to quantify imaging performance increases.

D. COIL CHARACTERIZATION ON THE BENCH

Each loop of the 8-channel coil was subsequently tested on the bench using a single-loop pickup coil and a network analyzer. The transmission coefficient (quantified by S_{21}) between the coil element connected to an industry test fixture (port 1) and a pickup loop (port 2) was measured. The fixture allows active decoupling through biasing of the diode and allows connection to DC power supply. The RF response was evaluated for each RF element separately and within the array. The feedboard including the preamplifier was included in the measurements.

E. IN VIVO MR IMAGING

We hypothesized improved imaging SNR and evaluated the imaging signal. As such, we validated the improvement of the SNR with and without the presence of the water-filled transducer at the thalamus region. A GE Healthcare Discovery MR750 system was used. *In vivo* MR images with the FUS-Flex receive coil were acquired with institutional

review board approval (IRB protocol number 20-03021574) and informed consent on healthy volunteers without (setup 1) and with the transducer (setup 2). Images were compared with the body coil in receive mode. A water-filled transducer (INSIGHTTEC ExAblate neuro) was placed around the head of the two volunteers using the INSIGHTTEC sealant membrane. GE's T1 weighted volume imaging (3D Bravo) sequence (TE = 3 ms, TR = 7.4 ms, FA = 12° and Pixel bandwidth = 244.1 Hz/px) was used. The FUS-Flex coil was used in receive-only mode and the body coil was used as an RF transmitter. SNR was determined according to the NEMA MS 1-2008 standards publication (R2014, R2020) [51].

Note the coil was placed outside the water bath in the *in vivo* experiment to ensure electrical safety in this first, unsealed, feasibility evaluation.

III. RESULTS

A. ACOUSTIC TRANSPARENCY

Figure 4A, F show 2D maps of the acoustic field pressure for low and high frequencies and the interaction of the acoustic field with the coil in cases 2 and 3. The acoustic field magnitude is shown in Figure 4B-E, G-J; results along the z - and r - direction were normalized to the case without a coil (reference). The results along the z -direction (parallel to the wave propagation direction) for case 2 exhibit an attenuation of the peak intensity at the focal point at $z = 221$ mm by 16% and 11% for 220 kHz and 650 kHz, respectively, and the displacement of the focal point was around 1.59 mm and 0.11 mm at 220 kHz and 650 kHz, respectively. In the third case, minor signal fluctuations were observed (<5%) with a shift of the focal point by less than 0.39 mm for both frequencies. Focal point locations along the r -direction (in plane/perpendicular to the direction of the wave propagation) were less affected, a negligible shift was observed at $r = 0$ mm), and the highest attenuation was observed for case 2: about 6% and 3% for the 220 kHz and 650 kHz frequencies, respectively.

The experimental measurements show that the relative acoustic attenuation (normalized to the case without a coil) due to the single-channel FUS-Flex coil varies from about 1% to 5% in the frequency range from 100 kHz to 700 kHz (Figure 5), which confirms the simulated results (case 3). The acoustic attenuation due the INSIGHTTEC membrane varies from about 10% to 30% in the frequency range from 100 kHz to 700 kHz. In summary, the FUS-Flex coil outperforms the INSIGHTTEC sealing membrane, which is specifically made to be acoustically transparent by the vendor.

B. ELECTROMAGNETIC SIMULATIONS

1) COMPARISON OF FUS-FLEX COIL TO CONVENTIONAL BIRDCAGE HEAD COIL

The use of the FUS-Flex coil improves the simulated B_1^- values, and therefore the SNR by a factor of 4 \times in the sagittal plane and 9 \times in the coronal plane over a standard birdcage head coil in the thalamus region (Figures 6A, B), demonstrating significantly improved performance even in a non-MRgFUS brain exam.

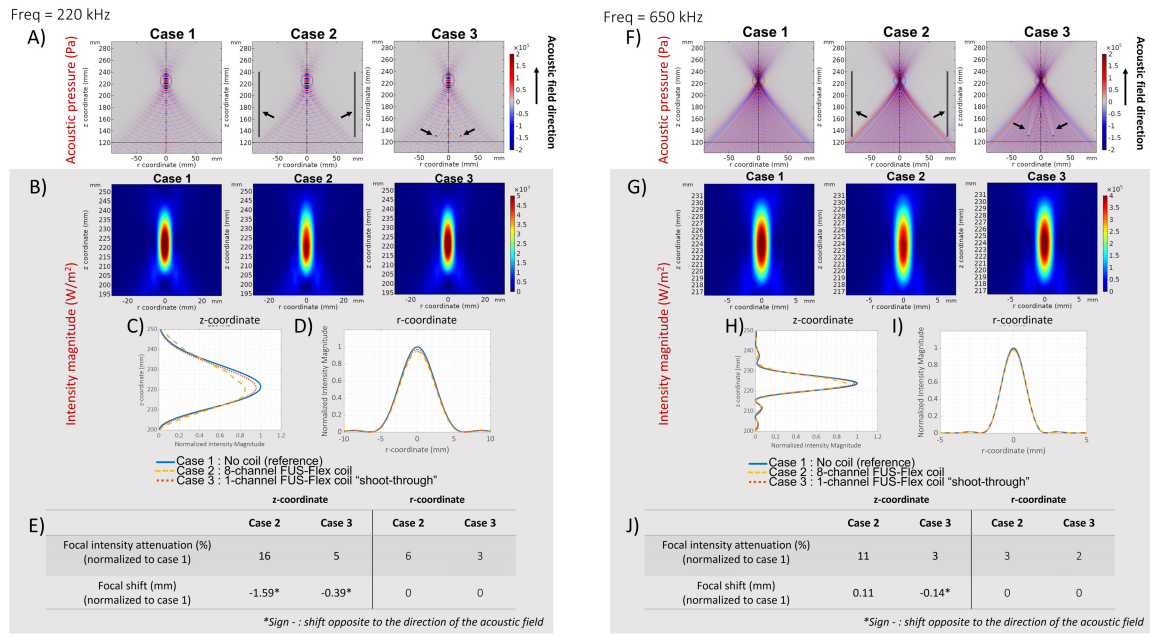


FIGURE 4. 2D map of total acoustic pressure showing the effects of the RF coil on the acoustic field: (A) 220 kHz; (F) 650 kHz. 2D map of intensity magnitude: (B) 220 kHz; (G) 650 kHz. First column: case 1 without RF coil. Second column: case 2 with coil around the focal point. Third column: case 3 FUS-Flex coil placed in the acoustic path - “shoot-through”. The black arrows show the positions of the FUS-Flex coil for cases 2 and 3. Normalized radial acoustic intensity magnitude for (C, D) 220 kHz and (H, I) 650 kHz along the dotted line passing through the focal point along the z-coordinate. Tables showing the acoustic attenuation and the focal point shift for cases 2 and 3 normalized to the reference case without a coil (case 1) for (E) 220 kHz and (H) 650 kHz.

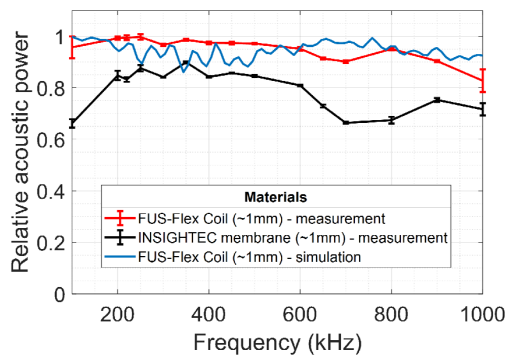


FIGURE 5. Relative acoustic power transmitted through the FUS-Flex coil and the INSIGHTEC sealant membrane. Error bars show the standard deviation. Note that the measurement and simulated curves are not representing the exact same scenario. Measurement: single immersion transducer - simulation: 30 cm diameter focused ultrasound transducer.

2) FUS-FLEX COIL WITHIN ULTRASOUND TRANSDUCER AND COMPARISON TO BODY COIL

The RF signal reflection from the copper-coated transducer produces E -field minima and causes a typical low-signal band in MRgFUS images along with a significant reduction in B_1 magnitude (Figure 6).

Figures 6C, D show the simulated B_1^- maps for FUS-Flex and body coils, denoting a SNR improvement at the position of the thalamus of $\sim 13\times$ and $\sim 15\times$ with and without the transducer, respectively, in both sagittal and coronal planes.

C. COIL CHARACTERIZATION ON THE BENCH

We confirmed that the magnetic coupling between the coil elements was minimized through overlapping (Figure 7). The measured quality factor ratio ($Q_{unloaded}/Q_{loaded}$) was approximately 4.5 [46], [52], indicating sample dominant losses.

D. IN VIVO MR IMAGING

Images acquired using the FUS-Flex coil in Figure 8 depict the position of the thalamus in a healthy volunteer with high sensitivity and show clear improvement of the low-signal band. At this location, the SNR gain is 7.3-fold and 7.6-fold compared to the body coil, with and without the MRgFUS transducer present, respectively. Note that for a 2-fold acquisition time (t_{acq}), the experimental SNR increase factor (7.3 and 7.6) can be multiplied by $\sqrt{2}$ and equal ~ 11 , which agrees with the simulation results.

We would also like to note that the position and intensity of the low-signal band artifact is the result of complex electromagnetic field interferences and reflections and strongly depends on a number of parameters, such as the positioning of the head, the amount of water used, and other factors. Since the volunteer in Figure 8 was not part of an actual MRgFUS surgical treatment, we did not use the typical mounting screws and frame for reasons of volunteer comfort. The head is slightly tilted and located off-center, resulting in a shift of the low-signal band to the frontal upper region of the brain, partially extending into the water bath. Overall, the simulated increase in SNR is a close match to the in vivo results for both

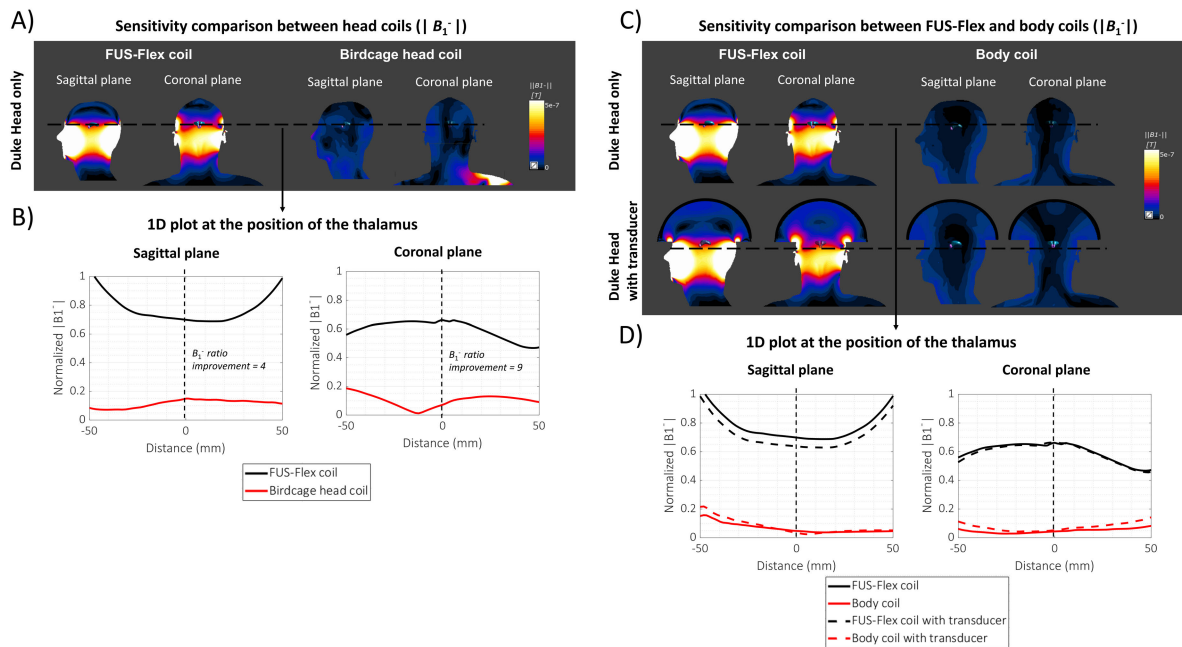


FIGURE 6. A) Sagittal and coronal plane of the B_1^- sensitivity map for receive-only 8-channel FUS-Flex (first column) and standard birdcage head coils (second column). The origin of the simulated coordinate system is located at the center of the thalamus (blue and purple spot in the midbrain). B) 1D plot of B_1^- along the thalamus region. C) Sagittal and coronal plane of the B_1^- sensitivity map for receive-only 8-channel FUS-Flex (first column) and body coils (second column) without (first row) and with the transducer (second row). D) 1D plot of B_1^- along the thalamus region with and without the transducer.

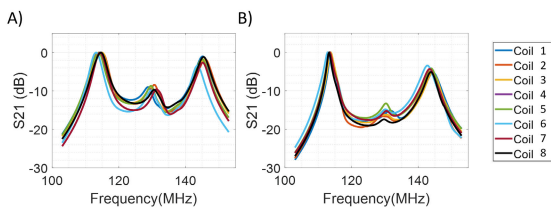


FIGURE 7. Sensitivity measurements of each coil element separately (A) and within the array (B).

volunteers, confirming the potential of FUS-Flex technology to yield improved MRgFUS imaging quality.

IV. DISCUSSION

In the above, we proposed the FUS-Flex concept, a new acoustically transparent 8-channel coil geometry, for use in MRgFUS neurosurgery. This is the first 8-channel coil built for transcranial MRgFUS applications. Choosing a coil array of 8 channels or more allows to not only increase the quality of the image, but to accelerate acquisition to provide fast, high-resolution imaging with accurate detection of the region of interest (ROI) and temperature monitoring, especially when parallel imaging is used. Increasing the number of channels can be easily achieved using coil technology with the heavy overlapping characteristic of RF elements beyond that of critical coupling [44], [45]. Current procedures often involve the coarse localization of the thalamus using the poor MR signal from the body coil. Non-ablative temperatures

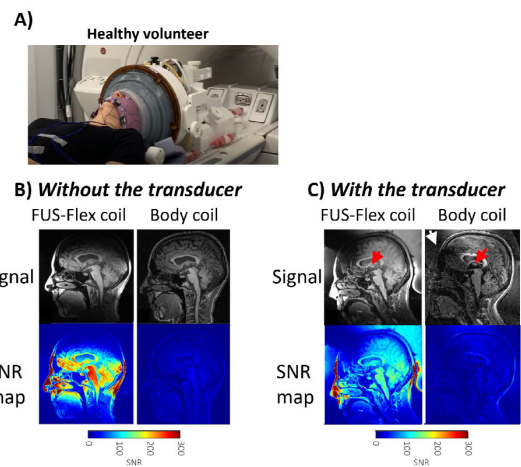


FIGURE 8. A) Setup of the FUS-Flex coil around a healthy volunteer without the transducer. Coronal MR images and SNR maps acquired with FUS-Flex and body coils of a healthy volunteer B) in absence of the transducer and C) in presence of the transducer. In vivo images were acquired using a T1 weighted volume imaging (3D Bravo) sequence (TE = 3 ms, TR = 7.4 ms, FA = 12°, and pixel bandwidth = 244.1 Hz/px). The red and white arrows show the positions of the thalamus and the low-signal band, respectively.

are then used to produce reversible sonication observable in the awake subject, thus providing a means to fine-tune the focal point at sub-millimetric accuracy. Our proposed coil array may avoid this tedious, risky, and uncomfortable calibration by providing suitable SNR and thus improved

spatial resolution, directly usable to precisely locate the target region.

Current T2-weighted intraprocedural imaging can require a scan time of 3 min [2], [23] and is carried out late in the protocol when cooling time already requires a halt of the procedure. Allowing for acquisition times <1min could benefit real-time intraprocedural imaging and hence confirmation of energy delivery and measurement of the ablation site. Moreover, diagnostic intraprocedural imaging could be useful when considering timing to conclude the treatment. Allowing 3D thermometry maps in real time, combined with active fusion to the DTI imaging, could help overcome the limitation of the body coil and improve the intraprocedural imaging utility.

Due to the severely degraded imaging performance, patients are often imaged without the transducer, using a standard birdcage head coil, after their treatment to obtain a high-resolution image of the target region. With the proposed FUS-Flex concept, it becomes attainable to provide such images at any time during the exam, interprocedurally, at a resolution that is potentially even higher than that of the birdcage head coil due to its decreased distance to the anatomy.

Highly flexible RF coil arrays are an emerging field of research even in applications that do not use MRgFUS. The fact that the coil array can be situated as conformally and as closely as possible with respect to the skin/skull (while obeying safety limits) maximizes the received MR signal and therefore the SNR in the MR image. Our proposed coil array is lightweight and flexible, allowing significant bending without performance decrease from geometry-dependent decoupling and resonance shifts that are normally observed in warped/stretched coil array designs [42].

It is to be noted that the FUS-Flex surface receive coil will not directly/completely solve the low-signal band artifact. While the FUS-Flex concept is a receive-only solution, the coil is located directly around the area of the brain with the infamous low-signal band, thus increasing SNR in the affected region (Figure 6C). Its increased receive SNR suggests feasibility to produce a significant increase of MRgFUS image quality over the body coil.

A large hindrance to the success of specific coil designs for MRgFUS has been their acoustic footprint and thus the distortion of ultrasound signal, which ultimately results in a physical shift, signal loss, and/or broadening of the focal point. The presented RF coil array is comprised of ultra-thin wiring mounted on acoustically transparent fabric. We simulated the presence of the FUS-Flex coil in an MRgFUS system using COMSOL Multiphysics and demonstrated the transparency of the coil when it is placed in the acoustic path. Our results indicate that the acoustic footprint of the coil is very small compared to the attenuation/aberration caused by the skull (70% of skull attenuation [53] versus 3% (650kHz) and 5% (220kHz) (shoot-through) as well as 11% (650kHz) and 16% (220kHz) (coil array around the head) of attenuation from our coil). Note that while lower frequencies are generally

attenuated to a lower degree than their higher counterparts, they also propagate deeper into the tissue, potentially causing a larger scattered field and therefore a more pronounced interaction with the focal point. These findings are in line with Fig. 3a in [30]. In comparison, Köhler *et al.* showed that using a thin rod ($\varnothing = 0.5$ mm) placed in the path of the acoustic beam (shoot-through) decreases the acoustic pressure by only 1.6%, which means that the focal spot remains unaffected [54]. These results are consistent with those of the FUS-Flex coil presented here and demonstrate the importance of using thin wire coils in MRgFUS procedures. Moreover, the attenuation incurred here is reduced compared to the screen-printed design in [30]. In addition, the transducer elements can be selectively deactivated thus avoiding interaction with the coil elements. As a result, we do not expect a major need to refocus the acoustic target location beyond what is already employed when correcting for the skull. The possibility to take into account the coil in the correction of the phase aberration, in a similar way to the skull, will be studied to further improve the acoustic attenuations.

Our final clinical goal for this work is to use the coil entirely (or sometimes partially, depending on anatomy) inside the water bath. At this proof-of-concept stage, we do not yet incorporate a fully sealed, waterproof, design. Inserting the coil into the water bath requires additional work with regard to transparency, air bubbles, and water permeability. Electrical safety is a big concern when working with in vivo subjects as well as costly MRI systems. This is outside the scope of this feasibility study and part of current and future work. Yet, we show that even with this low-profile 8-channel coil placed outside the water bath, we improve SNR significantly. The acoustic evaluation (experiment/simulation) along with the RF investigation (simulation partially/fully inside water bath, experiment outside water bath) performed in this paper suggest feasibility of full immersion once practical details of safe coil sealing are accomplished.

Future work will involve the use of higher channel counts to further increase the SNR and shorten acquisition time. A possible tradeoff between the number of channels and acoustic performance of the coil will be investigated. Along with increasing the number of channels, we can further optimize sequences to fall below the one-minute mark and thus allow for optimized intraprocedural acquisition. The FUS-coil can allow sequences such as DTI and 3D thermometry to achieve better results compared to the body coil in terms of image resolution and scan time and thus efficient monitoring of target and surrounding tissue. Future work will also include acoustic evaluation using the INSIGHTEC transducer as well as potential degassing of the coil fabric to remove air bubbles.

V. CONCLUSION

The proposed FUS-Flex coil is lightweight, stretchable, ultra-thin, and can potentially be adjusted to different head sizes and shapes without adding extra weight to the head while allowing the patient to see and breathe normally during procedures. When placing the FUS-Flex coil outside the water

bath, the SNR is improved a factor of 7.3 with $2\times$ acceleration (equivalent to $11\times$ without acceleration), leading to a higher SNR efficiency. Acoustic simulations and experiments show a negligible influence of the coil on the position of the focal point and acoustic signal for deep target applications (98% transparency simulated/measured).

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