On-demand field shaping for enhanced magnetic resonance imaging using an ultrathin reconfigurable metasurface

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The signal-to-noise ratio (SNR) is the main figure of merit that assesses the quality of magnetic resonance imaging (MRI). Existing studies mainly focus on improving the magnetic field intensities of the constant homogenous $B_0$ field from the main coil or the oscillating $B_1$ field from the radio frequency (RF) coil. In addition to these options, SNR also depends on the coupling between the imaging subject and the RF coil during the signal reception, which has been largely ignored. Here, we provide a different route toward enhancing the SNR of MRI by improving this coupling during the signal reception. We elucidate a theoretical design of an ultrathin metasurface with micrometer thickness and high flexibility. This metasurface is reconfigurable; it can selectively boost the SNR at a desired imaging region with any arbitrary shapes. Our design has shown that this metasurface can enhance SNR by up to 28 times in the region of interest. At the same time, the metasurface is designed to minimally disturb the excitation fields by less than 1.6%, thus maintaining the uniformity of the excitation, important to achieve a high-quality MR image without artifacts.

KEYWORDS
magnetic resonance imaging, medical imaging instrumentation, metamaterials, reconfigurable

Magnetic resonance imaging (MRI) has been widely used in staging tumors, imaging musculoskeletal systems, and monitoring brain functions.1,2 In MRI, the contrast among different types of tissue originates from the different decay rates of precession of nucleus spins.3 The typical resolution of MRI is in micrometers, and the scanning time is about tens of minutes; both the resolution and the scanning time depend heavily on the signal-to-noise ratio (SNR) of the imaging area.2 Therefore, the SNR has been a critical parameter associated with the imaging quality of MRI.

As the static magnetic $B_0$ field in MRI governs the magnetization of the nucleus spins, the signal increases with a stronger $B_0$ field intensity.4 On the other hand, the noises of MRI include Johnson noise of the imaging subject, Larmor frequency shift due to subject-receiving coil induction, resistive noise of the receiving coil, and circuit noise of the preamplifier.5 The major source of noises is Johnson noise caused by the Brownian motion of the electrons, which does not increase with the $B_0$ field intensity. As a result, when $B_0$ field increases from 3T to 7T, the SNR can be...
enhanced by approximately 5.4 times. However, the safety of a strong magnetization on biological subjects remains unclear; the financial cost to generate such a strong magnetic field is also substantial.\textsuperscript{5}

Another approach to enhance SNR is to increase the oscillating \(B_0\) field emitted and/or received by the receiving coil. Different surface coils have been designed for various imaging targets, such as eyes,\textsuperscript{7} knees,\textsuperscript{8} and heads.\textsuperscript{9} While a specific designed surface coil can improve the SNR by 2.7\textsuperscript{th}–5\textsuperscript{th} fold, it needs to be redesigned for different organs due to the constraints in shapes and rigidity; once developed, it does not provide the flexibility to image the specific region of interest within the imaging target. Recent research has shown great promise of applying the concept of metamaterials in MRI to further enhance the SNR without modifying the imaging system.\textsuperscript{10,11–13} Metamaterials are subwavelength periodic structures that can achieve anomalous field effects that are not available in natural materials.\textsuperscript{14} Because of the advantages of independently controlling the localized fields, metamaterials have been widely applied to boost or reshape electromagnetic fields in imaging applications.\textsuperscript{15} Notably, a metamaterial consisting of an array of copper-wires has shown to enhance the SNR of MRI by 2.7 times\textsuperscript{12} and a metamaterial with spiral arrays has shown to enhance the SNR by 4.2 times.\textsuperscript{11} However, one remaining challenge of this concept is the nonuniformity of the excitation field caused by the metamaterials.

To enhance the signals at the receiving coil, while preserving the uniformity of the excitation field, metamaterials with nonlinearities\textsuperscript{13} have been developed to tune the resonance frequencies, so it can be switched on/off during the reception/excitation.\textsuperscript{13} The nonlinear metamaterial can enhance SNR up to 15.9 times without interrupting the excitation but requires a thickness over 5 cm, which creates challenges to fit in the limited space between the patient and the receiving coil.\textsuperscript{5} Further enhancing the SNR based on this concept is limited because the energy is distributed in a large region. Very different from the previous work, here, we employ an ultrathin dynamically reconfigurable metasurface.\textsuperscript{16} Although resonance tuning is also feasible with our metasurface design, our working principle does not rely on tuning the resonances, but rather on reforming the eigenmodes of the metasurface for an on-demand enhancement region. We show that a maximum 28-fold enhancement of the SNR in a user-defined area can be achieved, simultaneously overcoming the limitation of the spatial constraints with an ultrathin profile in micrometers.

In MRI, the RF coil generates an oscillating magnetic field \(B_0\) to excite the proton-spin, and the decay is captured by the receiving coil. Here, we use \(B_1^+\) and \(B_1^-\) to represent the excitation signal and the reception signal, respectively. Because metasurfaces/metamaterials are oftentimes made of high-index dielectric materials or metal, it can locally enhance the electromagnetic fields but at the same time also create phase disruptions. The phase disruptions are not desired during the excitation, because it will create a nonuniform three-dimensional MR image as well as complicate the image reconstruction. To avoid phase disruptions during the excitation, we reform the eigenmodes of the metasurface only during the reception for an on-demand enhancement region, therefore, specifically enhance the \(B_1^-\) field without influencing the \(B_1^+\) field to ensure a uniform and optimal excitation. As a result, the receiving coil can measure a stronger decay-signal and thus achieve an improved SNR in the resulting image. With the contribution of the metasurface, the excitation \(B_1^+\) field becomes \(B_{1m}^+\), and the reception \(B_1^-\) field becomes \(B_{1m}^-\) which includes both the original \(B_1^+/B_1^-\) field and the scattered field from the metasurface. The objective of this study is to achieve a high enhancement ratio of the SNR, defined as \(B_{1m}^+/B_1^-\), at the same time, keep \(B_{1m}^+/B_{1m}^-\) as close to 1 as possible.

To shape a user-defined distribution of the \(B_{1m}^-\) field, we need to tailor the eigenmode of the metasurface. The eigenmode is described by the normalized distribution of \(a_n\), \(a_n = \sqrt{\frac{L}{2}}I_ne^{i\omega_0t}\), where \(L\) and \(I_n\) are the self-inductance and current of each unit cell. \(\hat{L}\) is normalized as 1 H. \(\omega_0\) is the angular frequency of the \(B_1^+\) and \(B_1^-\) field. Note that although the signal detected by the receiving coil \(B_1^-\) is generated by the imaging subject, based on the principle of reciprocity, the same field \(B_1^-\) can be generated by the receiving coil with a unit current at a specific location of the imaging subject. This concept is used to design our metasurface.

Our metasurface is composed of periodic arrays of resonators with significant mutual coupling among the neighboring unit cells. According to the coupled mode theory,\textsuperscript{12} \(a_n\) is related to the coupling among the unit cells \((\kappa_{kn})\) of the metasurface, the intrinsic loss \((\Gamma_n)\), and the driving force \((F_n(t))\) of each unit cell,

\[
\frac{da_n(t)}{dt} = (i\omega_0 - \Gamma_n)a_n(t) + i\sum_{k=1}^{m}\kappa_{kn}a_k(t) + F_n(t), \quad (1)
\]

where \(m\) is the total number of the unit cells of the metasurface. The intrinsic loss \(\Gamma_n\) is directly associated with the impedance \(Z_n\) of each unit cell as \(\Gamma_n = \frac{Z_n}{2L}\). \(\kappa_{kn}\) is the coupling coefficient between the \(k\)th and \(n\)th unit cell. The driving force, \(F_n(t)\), excites the eigenmodes of the metasurface but does not define the eigenmodes; therefore, we ignore this term in our design.

By expanding Equation (1), we deduce the relation among the intrinsic loss \(\Gamma_n\), the coupling coefficient \(\kappa_{kn}\),
FIGURE 1  Configuration of the metasurface. (A) Left: The unit cell of the metasurface is composed of a Hg-Al$_2$O$_3$-Cu tri-layer structure with a capacitive layer in between the Hg and Cu layers. The capacitance can be tuned by the overlap area between the top and bottom layers. The radius $r$ of the Hg and Cu rings is 4 mm, the width $w$ is 1 mm, and the thickness $h_1$, $h_2$, and $h_3$ is 500, 1, and 1.2 $\mu$m, respectively. Right: Schematic illustration of a reconfigurable metasurface composed of $3 \times 3$ unit cells, $d$ is the in-pane periodicity, $\kappa_{01}$ refers to the coupling coefficient between the 0th and 1st unit cells. (B) $\text{Im}[\Gamma]$ as a function of the arc angle ($\theta$) of the Hg layer. The analytical results are calculated using the RLC lumped circuit element model. (C) Comparison between the simulated and analytical coupling coefficients as a function of distance between coupled resonators by assuming the periodicity $d$ of 1 cm. For example, $\kappa_{01}$ denotes the coupling coefficient between the 0th and 1st unit cells as shown in (A). The inset shows the error map between the simulated and analytical coupling coefficients with various angles $\theta$ for unit cell 0 and unit cell 1.

and the targeting eigenmode $a_{nt}$ (Supporting Information, Section I),

$$\Gamma_n = \frac{i \sum_{k=1}^{m} \kappa_{kn} a_{kt}}{a_{nt}}.$$  \hspace{1cm} (2)

To shape the eigenmode, one can either tune the coupling coefficients $\kappa_{kn}$ or the intrinsic loss $\Gamma_n$ of each unit cell. As $\kappa_{kn}$ depends on the relative positions of the unit cells, changing the position of one unit cell will alter its relative position to the rest of the unit cells. Therefore, $\kappa_{kn}$ can hardly be controlled independently. On the other hand, $\Gamma_n$ is only related to the impedance of each unit cell, thus, can be readily controlled independently. In our metasurface design, we fix the coupling coefficient $\kappa_{kn}$ (i.e., the periodicity of the metasurface), and tune $\Gamma_n$ to achieve the desired eigenmode. The real part of $\Gamma$ represents the intrinsic resistive loss of the unit cell that leads to attenuation of the $B_{lm}^-$ field, therefore, lossless/low-loss materials at the operation frequency needs to be employed for constructing the metasurface to minimize heating. The imaginary part of $\Gamma$ ($\text{Im}[\Gamma_n]$) comes from the capacitance and inductance of each unit cell constituting the metasurface, with the capacitance being the dominant variable at the given frequency. Therefore, we choose to tune the capacitance in our metasurface design.

Here, we focus on designing a metasurface operating at 127 MHz that corresponds to the magnetic resonance frequency of proton in the 3T MRI, as commonly used in current clinical scanner. As shown in Figure 1A, our metasurface is composed of a metal-insulator-metal tri-layer structure. The top layer is a microfluidic circular channel with a radius $r$. The channel is partially filled with a liquid metal (such as eutectic gallium–indium alloy or mercury) with an arc angle $\theta$ to form a C shaped metal layer. The additional liquid metal can be pumped in and out to change the arc angle $\theta$, and thus tune the $\text{Im}[\Gamma_n]$. We chose mercury (Hg) as a model liquid metal to demonstrate our theory since its optical properties are well known. The spacer layer is aluminum oxide (Al$_2$O$_3$), and the bottom layer is a fixed Cu ring-resonator. The
periodicity (d) of the metasurface is 1 cm, the radius (r) is 4 mm, and the width (w) is 1 mm. The thickness of the top, spacer, and bottom layers of the metasurface (h₁, h₂, and h₃) is 500, 1, and 1.2 μm, respectively.

We first explore the relation between $\theta$ and $\text{Im}[\Gamma]$ and simulate $\text{Im}[\Gamma]$ as a function of various $\theta$ using the finite element method (CST studio suite 2019), and compare these results with analytical calculations. The analytical results (Figure 1B, black curve) are generated using the lumped circuit model. Our simulation matches well with the analytical solution with an error less than 1.35% (Figure 1B). To reduce the computational burden, we use the analytical solution in our later design. Our results show that as the arc angle $\theta$ changes from 20° to 180° (Figure 1B), $\text{Im}[\Gamma]$ can be tuned from −29.3 (at 20°) to 2.94 (at 180°), sufficient to switch the desired eigenmode “off” and “on” during the excitation and reception.

There are two approaches to calculate the coupling coefficients $\kappa_{kn}$ among each unit cells, one is the analytical approach based on Neumann’s theory of mutual inductance, and the other is the full wave numerical simulation. The analytical approach is more computational efficient thus more readily adaptable in the field. Note that the Neumann’s theory assumes the inductors’ width is negligible comparing to their distances, so the analytical approach will lead to increasing errors when the distances between coupling unit cells reduce. We compare the analytical results of $\kappa_{kn}$ with numerical simulations among six nearby unit cells (unit cells 0–5) by assuming the metasurface periodicity $d = 1$ cm (Figure 1C). As expected, when the distances between the coupling unit cells increase, the coupling coefficients $\kappa_{kn}$ decay, and the absolute errors of the analytical results also decrease. When the distance between two unit cells is fixed (i.e., between unit cell 0 and unit cell 1), the error between the analytical result and the simulation is dependent on the arc angle $\theta$. The smaller the arc angle $\theta$ of each unit cell, the smaller the error will be.

A reconfigurable metasurface allows for selectively turning “off” and “on” the eigenmode in the excitation and reception of the MRI by changing the arc angle $\theta$. During the excitation, a uniform $B_{1m}^+$ field distribution is desired, so we turn “off” the eigenmode of the metasurface by setting the angle $\theta$ of all unit cells at 20°. As shown in Figure 2, $B_{1m}^+$ field remains mostly uniform, and the difference between $B_{1m}^+$ field and $B_{1m}^-$ field is less than 1.6%, demonstrating the minimal influence of the metasurface during the excitation.

During the reception, to enhance the $B_{1m}^-$ field thus the SNR in a region of interest, we reset the $\text{Im}[\Gamma]$ distribution to switch “on” the eigenmode. To demonstrate this on-demand regional enhancement, we choose a target region from a brain tumor MRI scan as shown in Figure 3A. The goal is to create the enhanced $B_{1m}^-$ field that matches the shape of the brain tumor and minimize the $B_{1m}^+$ field elsewhere. To do so, ideally, the amplitude of the eigenmode $a_{nt}$ outside the tumor region should be 0 and inside the tumor should be 1. However, the desired eigenmode distribution is formed by the entire field both outside and inside the region of interest. Therefore, if the field amplitude outside the tumor is set to be 0, one cannot achieve the desired enhancement inside the tumor region. We choose a target distribution with the field amplitude outside the tumor region to be 0.1, and inside the tumor to be 1.

To achieve $a_{nt}$ as 1 for the tumor region and 0.1 elsewhere, we design the impedance map of the metasurface as shown in Figure 3B. Our calculation shows that the $B_{1m}^- / B_{1m}^+$ ratio can be enhanced 6.2 and 3.4 times, respectively, in the imaging planes at 1 and 2 cm above the metasurface (Figures 3C and D). Although the enhancement
ratio decays gradually as the imaging plane moves away from the metasurface, the enhancement ratio remains at least 1.5 times at 5 cm above the metasurface.

Because of energy conservation, the enhancement ratio inevitably decays. By taking advantage of the reconfiguration of the metasurface, it is possible to redirect the energy and extend the enhancement region further with the cost of a relatively smaller enhancement ratio near the metasurface. This can be achieved by selectively “focusing” the field with a metasurface supporting nonuniform eigenmode, such as a Gaussian distribution. The intensity distribution of a Gaussian field is

\[ I(x, y) = \exp \left( -\frac{4 \ln(2) \sqrt{(x-x_0)^2 + (y-y_0)^2}}{w^2} \right), \]  

where \( w \) is the full width at half maximum (FWHM), \( x_0 \) and \( y_0 \) are the coordinates of the “focal point.” Figures 4A and C show various \( \text{Im}[\Gamma] \) distributions of the metasurface, which targets a Gaussian eigenmode with an FWHM of 4 and 8 cm, respectively, with a focus location at (0 cm, 0 cm). When translating the \( \text{Im}[\Gamma] \) distribution, we can move the “focal point” in the transverse plane as seen in Figure 4B, where the 4 cm-FWHM enhancement region is relocated to (0 cm, 4 cm).

The enhancement ratio at the focus depends on the FWHM of the field. We calculate the enhancement ratios of Gaussian fields with FWHM of 2, 4, 8, and 12 cm at various distances away from the metasurface to mimic the imaging penetration. As shown in Figure 4G, at a distance of 5 mm, the enhancement ratio at the “focal point” of 2 cm-FWHM-Gaussian field is 28; for 4-, 8-, and 12 cm-fields, the enhancement ratios are 21, 15, and 12, respectively, which is 2–11 fold higher compared to other metamaterial designs at the same imaging depth. At such a distance, the enhancement ratio of a 2 cm-Gaussian field is shown to be 28 at the center of the field. Using this typical distance between the metasurface/metamaterial and the imaging subject, the enhancement ratio of the wire-metasurface is \( \sim 2.5; \) the magnetic metamaterials is \( \sim 4; \) and the nonlinear magnetic metamaterials is \( \sim 16. \)

As seen in Figure 4G, the decay rate of the signals along the z-direction is related to the designed FWHM of the Gaussian field. A broader targeted field will lead to a slower decay rate but a lower initial intensity right above the metasurface, because the energy is distributed to a larger area. To optimize the enhancement ratio at a certain depth \( z \), one can choose the field-width by considering both the...
FIGURE 4  Gaussian field with controllable field-width and position. (A)–(C) Im[Γ] distribution of a Gaussian field with field-width of 4, 4, and 8 cm and central location of (0 cm, 0 cm), (4 cm, 0 cm), and (0 cm, 0 cm), respectively. (D)–(F) The vertical cross-section (x–z plane) of the magnetic field with 1 cm above the metasurface corresponding to (A)–(C). The blue curves represent the enhancement ratio at z = 1 cm. (G) Enhancement ratio at the focal point as a function of z for 2-, 4-, 8-, and 12 cm-FWHM Gaussian field. The insets show x–y cross-section at 2.5 cm above the metasurface of the four Gaussian field, respectively.

initial value and the decay rate. As compared in Figure 4G, at 2.5 cm above the metasurface, the 8 cm-Gaussian field can achieve a relatively higher enhancement ratio than the 4 cm-Gaussian field by 12.17% due to its slower field decay, and a higher enhancement ratio than 12 cm-Gaussian field by 9.49% due to its higher initial intensity. In this way, one can control the desired SNR-enhanced region in three dimensions.

In many scenarios, the imaging region is not close to the surface of the subject. To receive the signal with a
lower attenuation in depth, the RF coil is often chosen as a birdcage coil instead of a surface coil. The excitation field is circularly polarized instead of linearly polarized, and the gap between the coil and the subject is cylindrical. 

As Al$_2$O$_3$ becomes flexible with micrometer thickness, the metasurface can be bendable and fit into the curved gap between the birdcage coil and the imaging subject. To test the performance of the metasurface with different bending curvature and circularly polarized $B_1^-$ field, we simulate a planar, half-cylindrical, and cylindrical metasurface as shown in Figures 5A–C to achieve a desired imaging area that shapes as the letter “I.” Figures 5D and E show the calculated $\text{Im}[\Gamma]$ map of the metasurface that remains relatively unchanged across the three different curvatures. It is because for certain targeted eigenmodes, $\text{Im}[\Gamma]$ is mostly determined by the coupling coefficient ($\kappa$) between the neighboring unit cells; the unit cell is much smaller than the metasurface with each taking only 0.23% of the area, so the coupling coefficient between the neighboring unit cells is insensitive to the curvature. The imaginary part of $\Gamma_n$ is much larger than its real part, as shown in Equation (3), the eigenmode of the metasurface can only be linearly polarized ($\alpha_{nt}$ needs to be in-phase). On the other hand, the $B_1^-$ field is circularly polarized (left handed), which mismatches with the eigenmode polarization. This mismatch results in the distortion of $B_1^-$ field as shown in Figures 5G–I, which increases with the curvature of the metasurface. Moreover, as the shape “I” is nonuniformly distributed among the rotational angle $\alpha$, the $B_1^\text{nt}/B_1^-$ field is asymmetric with respect to $\alpha$, this relationship will be reversed with an opposite handedness $B_1^-$. Despite the distortion induced by the polarization mismatch, the enhancement ratio at a certain part of the “I” shape remains high. So, the selective enhancement is still achievable for the scenarios using a birdcage coil.

In summary, here we introduced a design of reconfgurable metasurface for enhancing the receiving imaging signals on demand. In our demonstration, we have assumed a 3T $B_0$ field MRI, our design principle can be extended to other static $B_0$ fields by slightly changing the
unit cell parameters. Our designed metasurface can boost the reception fields, while poses little disruption to the excitation fields; notably, the designed enhancement ratio in a 3T-MRI is up to 28 times, while the disturbance of the excitation field is less than 1.6% across various imaging depths. These results demonstrate that we can achieve an on-demand enhancement in the MR images without the need of increasing the local magnetic fields. By shaping the uniformity of the enhancement region, we also show the tunability of the decay rate of the enhanced field across various imaging depths, highlighting the flexibility in enhancing MR imaging in three dimensions.

CONFLICT OF INTEREST
The authors declare no conflict of interest.

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REFERENCES
1. a) S. Ogawa, T. M. Lee, A. R. Kay, D. W. Tank, Proc Natl Acad Sci USA 1990, 87, 9868; b) D. Weishaupt, V. D. Kitchi, B. Marineck, How does MRI Work? An Introduction to the Physics and Function of Magnetic Resonance Imaging, Springer Science & Business Media, 2006; c) R. W. Brown, Y.-C. N. Cheng, E. A. Hong, Magnetic Resonance Imaging: Physical Principles and Sequence Design, Wiley-Blackwell, 2014; d) N. Petridou, D. Plenz, A. C. Silva, M. Loew, J. Bodurka, P. A. Bandettini, Proc Natl Acad Sci USA 2006, 103, 16015; e) J. Duyn, A. P. Koresky, Nat Clin Pract Cardiovasc Med. 2008, 5, 571; f) A. Barandov, B. B. Bartelle, C. G. Williamson, E. S. Loucks, S. J. Lippard, A. Jasanoff, Nat Commun. 2019, 10, 897.

2. Z.-P. Liang, P. C. Lauterbur, Principles of Magnetic Resonance Imaging: A Signal Processing Perspective, SPIE Optical Engineering Press, 2000.

3. D. G. Nishimura, Principles of Magnetic Resonance Imaging, Stanford University, 2010.

4. V. D. Schepkin, F. C. Bejarano, T. Morgan, S. Gower-Winter, M. Ozambela Jr., C. W. Levenson, Magn Reson Med. 2012, 67, 1159.

5. V. Hartwig, G. Giovannetti, N. Vanello, M. Lombardi, L. Landini, S. Simi, Int J Environ Res Public Health 2009, 6, 1778.

6. A. Nowogrodzki, Nature 2018, 563, 24.

7. F. Schenck, A. Hart, H. Foster, A. Edelstein, A. Bottomley, W. Redington, American Journal of Roentgenology 1985, 144, 5.

8. D. Burk, E. Kanal, J. Brunberg, G. Johnstone, H. Swensen, G. Wolf, Am J Roentgenol 1986, 147, 293.

9. J. A. d. Zwart, P. J. Ledden, P. Kellman, P. v. Gelderen, J. H. Duyn, Magn Reson Med. 2002, 47, 1218.

10. a) A. Hurshkainen, A. Nikulin, E. Georget, B. Larrat, D. Berraou, A. L. Neves, P. Sabouroux, S. Enoch, I. Melchakova, P. Belov, S. Glybovskiy, R. Abdeddaim, Sci Rep. 2018, 8, 9190; b) E. I. Kretov, A. V. Shchelokova, A. P. Slobozhanyuk, Appl Phys Lett. 2019, 115, 061604; c) R. Schmidt, A. Slobozhanyuk, P. Belov, A. Webb, Sci Rep. 2017, 7, 1678; d) R. Schmidt, A. Webb, ACS Appl Mater Interfaces 2017, 9, 34618; e) A. V. Shchelokova, A. P. Slobozhanyuk, S. Saha, I. Sotiriou, M. Koutsoupidou, G. Paliaras, E. Kallos, P. A. Belov, A. Webb, “In vivo magnetic resonance imaging of human knee with metasurface”, presented at 2017 Progress in Electromagnetics Research Symposium - Spring (PIERS), 2017.

11. G. Duan, X. Zhao, S. W. Anderson, X. Zhang, Commun Phys. 2019, 2, 35.

12. A. P. Slobozhanyuk, A. N. Podduhny, A. J. E. Raaijmakers, C. A. T. v. d. Berg, A. V. Kozachenko, I. A. Dubrovina, I. V. Meldakova, Y. S. Kvishar, P. A. Belov, Adv Mater. 2016, 28, 1832.

13. X. Zhao, G. Duan, K. Wu, S. W. Anderson, X. Zhang, Adv Mater. 2019, 31, 1905461.

14. a) R. A. Shelley, D. R. Smith, S. Schultz, Science 2001, 292, 77; b) J. B. Pendry, D. Schurig, D. R. Smith, Science 2006, 312, 1780; c) J. Valentine, S. Zhang, T. Zentgraf, E. Ulm-Avila, A. A. Genov, G. Barta, X. Zhang, Nature 2008, 455, 376; d) N. I. Zheludev, Y. S. Kvishar, Nat Mater. 2012, 11, 917; e) A. Boltasseva, H. A. Atwater, Science 2011, 331, 290; f) S. Linden, C. Enkrich, M. Wegener, J. Zhou, T. Koschnik, C. M. Soukoulis, Science 2004, 306, 1351; g) O. Hess, J. B. Pendry, S. A. Maier, R. F. Oulton, J. M. Hamm, K. L. Tsakmakidis, Nat Mater. 2012, 11, 573; h) A. Alù, M. G. Silverinaha, A. Saldarriino, N. Engheta, Phys Rev B 2007, 75, 155410; i) Y. Zhao, M. A. Belkin, A. Alù, Nat Commun. 2012, 3, 870; j) N. Engheta, R. W. Ziolkowski, Metamaterials: Physics and Engineering Explorations, John Wiley & Sons, Inc., Hoboken, NJ 2006; k) N. Mohammadi Estakhri, B. Edwards, N. Engheta, Science 2019, 363, 1333; l) B. Lu’yanychuk, N. I. Zheludev, S. A. Maier, N. J. Halas, P. Nordlander, H. Giessen, C. T. Chong, Nat Mater. 2010, 9, 707; m) N. Liu, H. Guo, L. Fu, S. Kaiser, H. Schweizer, H. Giessen, Nat Mater. 2008, 7, 31; n) W. Cai, U. K. Chettiar, A. V. Kildishev, V. M. Shalaev, Nat Photonics 2007, 1, 224; o) H. Alaeian, J. A. Dionne, Phys Rev A 2014, 89, 033829.

15. a) C. M. Soukoulis, M. Wegener, Nat Photonics 2011, 5, 523; b) W. Fan, B. Yan, Z. Wang, L. Wu, Sci Adv. 2016, 2, e1600901; c) Z. Jacob, L. V. Alekseyev, E. Narimanov, Opt Express, 2006, 14, 8247; d) X. Zhang, Z. W. Liu, Nat Mater. 2008, 7, 435; e) A. Podubny, I. Iorsh, P. Belov, Y. Kvishar, Nat Photonics 2013, 7, 948; f) C. M. Soukoulis, S. Linden, M. Wegener, Science 2007, 315, 47; g) Smolyaninov, II, J. Y. Hung, C. C. Davis, Science 2007, 315, 1699; h) J. Zhu, J. Christensen, J. Jung, L. Martin-Moreno, X. Yin, L. Fok, X. Zhang, F. J. Garcia-Vidal, Nat Phys. 2011, 7, 52; i) M. Khorasaninejad, F. Capasso, Science 2017, 358, eaam8100; j) N. Yu, F. Capasso, Nat Mater. 2014, 13, 139; k) F. Aieta, P. Genevet, M. A. Kats, N. F. Yu, R. Blanchard, Z. Gaburro, F. Capasso, Nano Lett. 2012, 12, 4932.

16. a) A. V. Kildishev, A. Boltasseva, V. M. Shalaev, Science 2013, 339, 1232009; b) G. X. Zheng, H. Muhlenbernd, M. Kenney, G. X. Li, T. Zentgraf, S. Zhang, Nat Nanotechnol. 2015, 10, 308; c) C. L. Holloway, E. F. Kuester, J. A. Gordon, J. O. Hara, J. Booth, D. R. Smith, IEEE Antennas Propag Mag. 2012, 54, 10; d) Y. Zhao, X. X. Liu, A. Alu, J. Opt. 2014, 16, 123001; e) H. T. Chen, A. J. Taylor, N. F. Yu, Rep Prog Phys. 2016, 79, 076401; f) Q. He, S. Sun, L. Zhou, Research 2019, 2019, 16; g) W. Zhang, Q. Song, W. Zhu, Z. Shen, P. Chong, D. P. Tsai, C. Qiu, A. Q. Liu, Adv Phys X,
2018, 3, 1417055; h) W. Zhu, Q. Song, L. Yan, W. Zhang, P.-C. Wu, L. K. Chin, H. Cai, D. P. Tsai, Z. X. Shen, T. W. Deng, S. K. Ting, Y. Gu, G. Q. Lo, D. L. Kwong, Z. C. Yang, R. Huang, A.-Q. Liu, N. Zheludev, Adv. Mater. 2015, 27, 4665; i) Y. Hui, J. S. Gomez-Diaz, Z. Qian, A. Alù, M. Rinaldi, Nat. Commun. 2016, 7, 11249.

17. a) A. Kurs, A. Karalis, R. Moffatt, J. D. Joannopoulos, P. Fisher, M. Soljačić, Science 2007, 317, 83; b) H. A. Haus, W. Huang, Proc. IEEE 1991, 79, 1505.

18. P. C. Wu, W. Zhu, Z. X. Shen, P. H. J. Chong, W. Ser, D. P. Tsai, A.-Q. Liu, Adv. Opt. Mater. 2017, 5, 1600938.

19. The ISPN Guide to Pediatric Neurosurgery, 2020.

20. R. Schmidt, A. Slobuzhanyuk, P. Belov, A. Webb, Sci. Rep. 2017, 7, 1.

21. N. Münzenrieder, L. Petti, C. Zysset, T. Kinkeldei, G. A. Salvatore, G. Tröster, IEEE Trans. Electron Devices 2013, 60, 2815.

22. A. D. Falco, Y. Zhao, A. Alù, Appl. Phys. Lett. 2011, 99, 163110.

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