Bioinspired cilia arrays with programmable nonreciprocal motion and metachronal coordination

Xiaoguang Dong1*, Guo Zhan Lum2*, Wenqi Hu1, Rongjing Zhang3, Ziyu Ren1, Patrick R. Onck3, Metin Sitti1,4,5†

Coordinated nonreciprocal dynamics in biological cilia is essential to many living systems, where the emergent metachronal waves of cilia have been hypothesized to enhance net fluid flows at low Reynolds numbers (Re). Experimental investigation of this hypothesis is critical but remains challenging. Here, we report soft miniature devices with both ciliary nonreciprocal motion and metachronal coordination and use them to investigate the quantitative relationship between metachronal coordination and the induced fluid flow. We found that only antiplectic metachronal waves with specific wave vectors could enhance fluid flows compared with the synchronized case. These findings further enable various bioinspired cilia arrays with unique functionalities of pumping and mixing viscous synthetic and biological complex fluids at low Re. Our design method and developed soft miniature devices provide unprecedented opportunities for studying ciliary biomechanics and creating cilia-inspired wireless microfluidic pumping, object manipulation and lab- and organ-on-a-chip devices, mobile microrobots, and bioengineering systems.

INTRODUCTION

Cilia broadly exist in a wide range of organisms on the micrometer scale and can induce notable net fluid flows at very low Reynolds number (Re; ~0.001 to 0.01). Their ability of producing net fluid flows plays an essential role in many living systems, such as the self-propulsion of Paramecium (1); inducing vortical fluid flows for enhancing the feeding function in coral reefs (2); and pumping biofluids in the respiratory (3), brain ventricle (4), and reproductive systems (5) in the human body. In addition to the nonreciprocal dynamics of a single cilium, metachronal coordination of cilia arrays is another fascinating behavior of cilia observed in various biological systems (6–9), which has been hypothesized to be critical to enhance net fluid flows (10–15). However, experimental validation is still missing, and the fundamental relationship between metachronal coordination and the induced fluid flow has not yet been resolved. This is because biological ciliary systems are not fully viable for controlled experiments, limiting the flexibility of investigating biological hypotheses.

One promising solution is to use small-scale soft devices with programmable motions (16–20) as a bioinspired platform to model and study their biological counterparts by more controlled experiments at a larger length scale while keeping the essential nondimensional parameters (e.g., Re) the same (21–23). These bioinspired soft devices can also enable unprecedented microfluidic functionalities for lab-on-a-chip (24, 25) and other bioengineering applications (26). So far, many small-scale artificial cilia (27–33) have been proposed for emulating motions produced by biological cilia arrays. However, at small-length scales, designing and controlling both the complex individual cilary motion and the overall coordinated dynamics within collective artificial cilia arrays pose an enormous conceptual challenge. Metachronal coordination has been mostly investigated in numerical studies (10–15, 27). For example, Khaderi et al. (27) did not experimentally create metachronal waves in artificial cilia, but numerically studied metachronal waves and experimentally demonstrated pumping water by a paramagnetic artificial cilia array with synchronized motions. This may be due to the challenge of controlling locally distributed magnetic fields at small scales. Recent works have proposed several ways to experimentally create metachronal coordination including using the length-dependent buckling motions of paramagnetic sheets (30) and magnetic anisotropy in paramagnetic rods (31) or ferromagnetic rod arrays (33). However, these existing works have focused on the fabrication methods and only demonstrated simple fluid pumping with non-optimized designs. In contrast, we developed design and fabrication methodologies to encode different nonreciprocal kinematics profiles into single-cilium motions and arbitrary phase shifts in a cilia array, even with two-dimensional (2D) metachronal waves and on curved surfaces like their biological counterparts. Moreover, we can optimize both the single-cilium dynamics and cilia array coordination independently to achieve the highest pumping performance by artificial cilia arrays with optimal metachronal coordination. We systematically investigate the quantitative relationship between metachronal waves and their induced fluid flows. These findings also enable us to create cilia-inspired fluidic soft devices with unique capabilities of pumping and mixing viscous synthetic and biological fluids at low Re. For example, we can pump even viscous syrup, synthetic mucus, and mouse blood in narrow channels.

RESULTS

Design artificial cilia arrays with programmable nonreciprocal motion and metachronal coordination

Here, we present a class of soft miniature devices consisting of multiple submillimeter-scale ferromagnetic-elastic sheets [see fig. S1 (A to C) and the “Fabrication of ferromagnetic cilia,” “Simultaneous encoding of cilia magnetization profiles,” and “Assembly of artificial cilia
arrays” sections in Materials and Methods] with identical geometries and nonidentical magnetization profiles. Figure 1 and movie S1 show that these magnetically actuated artificial cilia arrays are capable of creating both programmable nonreciprocal cilia motion and collective metachronal coordination. The complex nonreciprocal oscillating dynamics of a ferromagnetic-elastic sheet (referred as “cilium” later on) in viscous fluids emerges from the nonlinear relations between internal macroscopic elastic stress, hydrodynamic forces, and exerted external magnetic torques at low $Re$, which emulates the regulation of molecular motors in a real cilium. By proper design of the material elastic properties and individual cilium magnetization profiles $M(s)$ ($s$, material coordinate) as well as the time-varying external actuation magnetic field $B(t)$ ($t$, time), ferromagnetic-elastic cilia arrays can produce complex but still deterministic ciliary dynamics (Fig. 1A). In particular, with an encoded $M(s)$ and a rotating $B(t)$ (see fig. S2 and the “Magnetic actuation of ferromagnetic cilium” section in Materials and Methods for creating such a magnetic field), complex nonlinear oscillations emerge from a rotational buckling instability (30) when a cilium undergoes a so-called phase-locking motion (14) until it reaches a critical deformed state ($t/T$, 0.3 to 1; $T$, beating cycle; Fig. 1B) and then rapidly snaps back in the reverse direction ($t/T$, 0 to 0.3; Fig. 1B). This buckling-based motion with two strokes i.e., the power and recovery strokes, results in a nonzero swiping area $S_A = \int_0^T r_{\text{tip}}(x) \ dy$, which is the contour area (the gray

Fig. 1. Artificial magnetic cilia with programmable nonreciprocal motions and metachronal waves. (A) Schematics of 2D rotational buckling motion of a ferromagnetic-elastic cilium (sheet). $M(s)$, cilium magnetization profile; $B(t)$, external rotating magnetic field; $s \in [0, L]$, material coordinate in the cilium length direction ($L = 1$ mm); $\alpha$, cilium initial body angle. (B) Illustration of ciliary nonreciprocal dynamics obtained by numerical simulation. $S_A$ represents the swiping area of the cilium tip within a beating cycle $T$. (C) Video snapshots of an artificial cilia array with metachronal waves pumping viscous fluids (glycerol). $B(t)$: frequency $f = 1$ Hz and magnitude $B_m = 38$ mT. (D) Illustration of metachronal waves by the shifts in $r_{\text{tip}}$ ($i = 1$ to 6) obtained by numerical simulation. (E) The magnetization phase profiles $\phi(i)$ for the cilia in (C) and (D). The neighboring cilia have the same constant magnitude in their $M(s)$. (F) The linear mapping from $\Delta \phi_x$ to $\Delta \psi_x$ in the metachronal waves. (G) Snapshots and overlapped trajectories of transporting neutrally buoyant particles by an artificial cilia carpet with 2D metachronal waves. (i), top view; (ii), side view. In (G), $B(t)$: $f = 2.5$ Hz and $B_m = 40$ mT. Scale bars, 1 mm. Photo credit: Xiaoguang Dong, Max Planck Institute for Intelligent Systems.
area in Fig. 1B) of a single cilium’s tip trajectory \( r^{\text{tip}} \). Specifically, Fig. 1B illustrates a cilium dynamics with a positive swiping area by programming a specific \( \mathbf{M}(s) \). Please note that the net fluid flow is independent of the timing of cilium motion at low \( Re \) but rather depends on the trajectory of an artificial cilium (8).

By encoding different \( \mathbf{M}(s) \) of a cilium (see Fig. S3 and the “Simultaneous encoding of cilia magnetization profiles” section in Materials and Methods for encoding a specific magnetization profile), cilia dynamics with either a positive or negative swiping area can be produced as shown in Fig. S4, which leads to a net fluid flow in different directions. The swiping area is a common metric to quantify the nonreciprocil motion and the induced net fluid flow (11, 27, 32). We define the swiping area ratio (SAR) as a metric to quantify the effectiveness of producing a net fluid flow within a unit time by a single cilium of length \( L \), where \( \text{SAR} = \frac{\Delta \phi}{0.5 L^2} \) SAR is the normalized swiping area per unit time. Therefore, as SAR increases, the motion becomes more nonreciprocil, leading to increased fluid flows at low \( Re \). A computer-aided method based on a fluid-structure interaction (FSI) model (see the “Summary of the used coordinate systems” and “FSI model of single cilium dynamics” sections in Materials and Methods) was developed to optimize SAR. The single-cilium dynamics mainly depends on the material elastic modulus, \( \mathbf{M}(s) \), and \( \mathbf{B}(t) \). We specifically optimized \( \mathbf{M}(s) \) and \( \mathbf{B}(t) \) to maximize SAR, as shown in Fig. S5 and described in the “Optimizing a single cilium by maximizing the SAR” section in Materials and Methods. The advantage of our design method and system is that the individual cilium beating dynamics can be designed separately from their metachronal waves, giving substantial flexibility and viability of the collective motion.

Figure 1 (C and D) shows that arbitrary 1D metachronal waves can be further produced within an array of magnetic artificial cilia, by encoding phase shifts \( \Delta \phi_i = \phi_i - \phi_{i-1} \) in \( \mathbf{M}(s) \) (Fig. 1E) and applying a rotating \( \mathbf{B}(t) \). The dynamics of each cilium remains the same for one period (see the “Mechanics and design methodology of metachronal waves” section in Materials and Methods for the mechanics). These 1D metachronal waves can be quantified by the lags in the motions of neighboring cilia, such as the lag time \( \Delta t \) in the periodic beating dynamics when each cilium tip reaches the maximum position as shown in Fig. 1D. The phase shift in motions can be defined as \( \Delta \psi_x = \frac{\Delta \phi_i}{L} \cdot 2 \pi (\Delta t \in [0, T]) \). The \( x \) component of the wave vector is thus given by \( k_x = \Delta \psi_x/d \), for quantifying the spatio-temporal coordination. Consequently, Fig. 1F shows that \( \Delta \phi_i \) can be linearly mapped to \( \Delta \psi_x \) and \( k_x \), which suggests that arbitrary 1D metachronal waves can be encoded into neighboring cilia. Figure 1F also shows the so-called symplectic (\( \Delta \psi_x > 0 \)) and antiplectic (\( \Delta \psi_x < 0 \)) metachronal waves (7), where the direction of the wave propagation and the power stroke are in the same and opposite directions, respectively.

Similar to biological cilia, Figure 1G and movie S2 show that our method also allows designing and fabricating artificial cilia arrays to emulate the propagation of 2D metachronal waves in both the \( x \) and \( y \) and \( x-z \) planes with arbitrary 2D wave vectors, by encoding both the phase \( \Delta \phi_x \) and \( \Delta \phi_y \) into a 2D array. Different from the emergence of metachronal waves due to hydrodynamic interaction (12) and distal coupling (34) in biological systems, the metachronal waves in our system emerge from the rotational shift in the magnetization phase profile of each magnetic cilium subject to a rotating uniform magnetic field. Nonetheless, the resulting metachronal waves resemble those of their biological counterparts and can be used for investigating the function of metachronal coordination in producing fluid flows at low \( Re \).

**Optimal metachronal waves for pumping fluid flows efficiently**

To find the optimal metachronal waves for pumping directional fluid flows, Fig. 2 (A to C) compares the directional fluid flow produced by our artificial magnetic cilia arrays with different \( k_x \) (or \( \Delta \psi_x \)) (Fig. 2B) and intercilium spacing \( d_i \) (Fig. 2C), based on the particle image velocimetry (PIV) measurements inside glycerol (dynamic viscosity, 0.876 Pa·s; see the “Fluid flow measured by PIV” section in Materials and Methods for more details). Because of a small number of cilia at low \( Re \), fluidic boundary/wall effects (11, 35) could be substantial if the cilia are placed inside confined fluidic channels or other confined spaces. Here, we avoided such effects by using nearly semi-infinite boundary configuration as shown in Fig. 2A and Fig. S2F. To quantify the performance of pumping directional fluid flows, we use the widely used average volume flow rate \( Q_x (11-13) \) in the direction of pumping (+\( x \) direction). In Fig. 2B, \( k_x \) is varied from \(-2\pi/(3L)\) to \( 5\pi/(9L) \) when \( \Delta \psi_x \) is varied from \(-\pi \) to \( 5\pi/6 \), while \( d_i \) is equal to 1.5L and other parameters are kept identical. It is found that antiplectic waves produce a maximal directional fluid flow when \( \Delta \psi_x \in [-\pi/2, -\pi/6] \). For example, artificial cilia arrays with \( \Delta \psi_x = -\pi/3 \) achieve a \( Q_x \) of 4 mm$^3$·s$^{-1}$, which is about 1.6 times of that produced by arrays with synchronized motions and 4 times of the minimal net fluid flow produced by arrays with a symplectic wave (\( \Delta \psi_x = \pi/3 \)). In addition, Fig. 2C shows that artificial cilia arrays with denser spacing yield a larger \( Q_x \), by varying the intercilium spacing \( d_i \) from \( L \) to \( 2L \) while keeping \( \Delta \psi_x = -\pi/3 \) and other parameters identical. The quantitative relationship between \( k_x \) and \( Q_x \) is evident from Fig. 2C. Our experiments are in good agreement with computational fluid dynamics (CFD) simulations (see Fig. S6 and the “CFD simulations in comparison with experimental data” section in Materials and Methods) in terms of the overall trends, confirming the effect of the phase shift \( \Delta \psi_x \) and intercilium spacing \( d_i \) on the observed fluid flow. Our experiments prove the key role of antiplectic metachronal waves in enhancing fluid transportation. The finding is in accordance with that in biological systems (7, 36) and numerical works (12, 13) and introduces extra insights into the special function of antiplectic waves by reporting the specific range of wave factors for producing optimal net fluid flows.

While \( Q_x \), quantify the fluid transportation ability of artificial cilia arrays in the Eulerian space, we further investigate the effect of different metachronal waves and intercilium spacing on transporting and attracting particles in the Lagrangian space. In Fig. 2 (D to H) and movie S3, we compare the average transportation velocities \( \bar{u}_{px} \) of a tracer particle in the +\( x \) direction (Fig. 2D). Tracer particles (300 to 330 \( \mu \)m in diameter; density, 1.2 g·cm$^{-3}$) are transported in the +\( x \) direction by artificial cilia arrays with \( k_x \) varying from \(-2\pi/(3L)\) to \( 5\pi/(9L) \) when \( \Delta \psi_x \in [-\pi, 5\pi/6] \) and \( d_i \) equaling to 1.5L (see the “Particle transportation using cilia arrays” section in Materials and Methods for experimental details). The Lagrangian coherent structures of the local time-periodic fluid flow guarantee consistent trajectories of neutrally buoyant tracer particles (37). Figure 2E shows that the artificial cilia arrays with \( k_x \in [-\pi/(3L), -\pi/(9L)] \) when \( \Delta \psi_x \in [-\pi/2,$
−π/6] could pump particles with a maximal average speed (~0.45 mm/s), agreeing with the trend in Fig. 2B. At the same time, artificial cilia arrays with denser spacing result in larger particle transportation speeds as shown in Fig. 2C, consistent with the results in Fig. 2C.

Comparing the particle transportation trajectories shows that another function of the optimal antiplectic waves is attracting free particles. Figure 2G shows that metachronal waves with Δψx, varying from −π/6 to −π/6 attract particles closer to the ciliary surfaces compared with other designs. Figure 2H shows a negative correlation between dx and the distance of particles to the boundary wall dy. Such a result could support the finding that biological cilia provide a key functionality of predation by attracting specific microbiomes toward their surfaces (38).

**Mechanism of antiplectic waves enhancing fluid flows**

The reason of the superior ability of enhancing fluid flows by antiplectic waves is further investigated. Figure 3 shows that metachronal waves can modulate local fluid flows by producing different time-varying boundary conditions to the fluid, thereby varying the overall fluid transportation performance. This observation is obtained by the PIV data of artificial cilia arrays with representative metachronal waves. The PIV results in Fig. 3A show that the local fluid flow around cilium no. 3 is strongly influenced by its neighboring cilia nos. 2 and 4. Compared with an artificial cilia array showing a symplectic wave, such as Δψx = π/6, the fluid flow induced by an artificial cilia array with an antiplectic wave, such as Δψx = −π/6, is less blocked from neighboring cilia during the power stroke (t1 = 0.127; Fig. 3A). This leads to an increased and more continuous local flow in the power stroke direction. Meanwhile, the flow is more blocked from neighboring cilia during the recovery stroke (t2 = 0.45T; Fig. 3A), resulting in a reduced and blocked local fluid flow in the recovery stroke direction.

To quantitatively see such a difference in local flows, Fig. 3B compares the instant volume flow rate Qx passing through a selected cross section within a full period. The observation is that although an artificial cilia array with synchronized motion induces a large and single peak value of Qx due to the superposition effect of all cilia, artificial cilia arrays with metachronal coordination yield smaller but multiple peak values of Qx. Therefore, to make a fair comparison, we compare Qx produced by two artificial cilia arrays with opposite phase...
shifts ($\Delta \psi_x = \frac{\pi}{6}$ and $\Delta \psi_x = -\frac{\pi}{6}$) to see the difference of their induced fluid flows. In these two representative cases, the artificial cilia array with an antiplectic wave ($\Delta \psi_x = -\frac{\pi}{6}$) has a larger value of $Q_x$ during the power stroke and a smaller value of $Q_x$ during the recovery stroke, resulting in a larger net fluid flow within a period.

Simple metrics are further proposed to quantify the boundary conditions and predict the net fluid flow by artificial cilia arrays with different metachronal waves. To compare different instant boundary conditions, in Fig. 3C, we choose the instant inter-tip spacing $d_{3 \text{tip}}(t)$ ($i = 2$ to 5) between a selected cilium and its two neighbors to measure the no-slip boundary conditions. Figure 3D shows that antiplectic waves yield a larger $d_{3 \text{tip}}(t)$ during the power stroke and a smaller $d_{3 \text{tip}}(t)$ during the recovery stroke, compared with synchronized motions. In contrast, symplectic waves have a smaller $d_{3 \text{tip}}(t)$ during the power stroke and a larger $d_{3 \text{tip}}(t)$ during the recovery stroke. To compare different average boundary conditions and predict the corresponding average net fluid flows, we define the average $d_{3 \text{tip}}(t)$ in the synchronized case is caused by imperfect synchronization due to fabrication errors. (E) Experimental data of $\bar{Q}_x$ and its predicted value. $\bar{Q}_x = p \left[ \left( \frac{d_{\text{power}}}{L} \right)^3 + \left( \frac{1}{d_{\text{recovery}}} \right)^3 \right]$. $p = 0.25$ is a fitting variable. Error bars indicate SDs for $n = 4$ cilia (index, 2 to 5). In all experiments, glycerol (dynamic viscosity, 0.876 Pa·s) is used. B(t): $f = 2.5$ Hz and $B_m = 40$ mT. Scale bar, 1 mm.
proportional law given by \( Q_\varsigma \propto \left[ \left( \frac{d_{\text{w}}}{L} \right)^3 + \left( \frac{L}{d_{\text{e}}} \right)^3 \right] \). This metric correlates the average volume flow rate to the intercilium spacing of artificial cilia with different metachronal waves.

**Functional artificial cilia as fluidic devices by encoding optimal coordination**

The established quantitative relationship and mechanism between fluid flows and metachronal coordination further allow creating bioinspired cilia arrays as versatile fluidic devices with unique and important functionalities at low \( Re \). We demonstrate multiple functional fluidic devices by encoding diverse metachronal coordination and nonreciprocal motions.

In Fig. 4 and movie S4, we first demonstrate a bioinspired cilia array with optimal metachronal waves propagating on curved surfaces inspired by coral reefs (2). Existing artificial cilia are only capable of pumping on flat surfaces (27–29, 32, 33), which does not resemble their biological counterparts in terms of boundary morphologies (7).

In contrast, our developed artificial cilia array is capable of transporting particles along arbitrary curved surfaces, such as an S-shaped surface, as shown in Fig. 4A. To create bioinspired cilia planted on a boundary wall with much more complex curvatures, we generalize our method to include the cilium-attached coordinates \( \{ L_{si} \} \) and the individual dynamics of each cilium as two extra sets of design variables (see the “Encoding metachronal waves in cilia arrays on curved surfaces” section in Materials and Methods). The key design rule is to compensate the rotation of the cilium-attached coordinates when encoding the magnetization profile of each cilium such that a desired constant phase shift is kept between neighboring cilium even on curved surfaces (fig. S7).

With this design methodology, we develop a bioinspired cilia array with optimal metachronal waves, which can pump viscous fluids along curved surfaces. Figure 4B shows that such a bioinspired cilia array could transport particles tangentially to the attaching surface more efficiently, compared with an array with the same boundary wall but with synchronized motions due to the locally enhanced vortical flow, as visualized by dye and PIV in Fig. 4 (C and D, respectively). More bioinspired cilia arrays could be constructed with various complex boundary morphologies, such as large-polyed, mounding, and encrusting colony morphologies (39). Important scientific questions, such as how geometrical details of colony morphology affect the transportation performance of coral reefs (2) and similar cilia-covered organisms, can be potentially answered in the future to help researchers better understand the active mass transportation in ambient aquatic environments.

Pumping viscous fluids in narrow channels at low \( Re \) is challenging because of the no-slip boundary conditions (40) but is important in many biological systems, such as tubal transportation of ova in female fallopian tubes, where ciliary motion contributes to efficient transport in addition to muscular contractility (5). In Fig. 5 and movie S5, we demonstrate efficient transportation of particles in narrow channels filled with glycerol, by mimicking the natural cilia in tubal transportation. As shown in Fig. 5A, we jointly designed two cilia arrays with opposite SARS (see fig. S4), as their pumping directions are in the same direction (in \( \{ L_{si} \} \) subject to the same rotating \( \mathbf{B}(t) \)). Within each array, an optimal antiplectic metachronal wave is further encoded. In this way, the device transports fluids and particles faster (\( \bar{v}_{px} = 350 \mu m/s \)) than single cilia arrays only on the top (\( \bar{v}_{px} = 320 \mu m/s \)) and bottom (\( \bar{v}_{px} = 350 \mu m/s \)) side of the channel wall (Fig. 5B).

We also demonstrate pumping synthetic mucus (\( \bar{v}_{px} = 188 \mu m/s \); Fig. 5C) and syrup (\( \bar{v}_{px} = 491 \mu m/s \); Fig. 5D) in narrow channels using such a device, which could potentially help understand tubal transport of viscous biofluids in physiologically

---

**Fig. 4. Bioinspired cilia arrays with optimal metachronal waves propagating on curved surfaces.** (A) Illustration of encoding metachronal waves in artificial cilia on curved surfaces. The key design rule is to compensate the rotation of the cilium-attached coordinate of each cilium by designing their magnetization phase profiles (see fig. S7). The red and green lines represent the +X and +Y axes of the cilium-attached coordinates, respectively. Cilia magnetization profiles: \( \phi_i(t) = (i - 1) \Delta \phi + \pi/4 \) (i), where \( \Delta \phi = \pi/4 \). Boundary wall positions: \( x_j = \ell \sin \left( \frac{2j\pi}{8} \right) \) (8), where \( \ell_j = 1, 2, \ldots, 8 \) is chosen with an equal spacing from \(-4L \) to \( 4L \). (B) Comparison of the particle transportation performance between bioinspired cilia arrays with an (i) optimal metachronal wave (\( \Delta \phi _{w} = -\pi/4 \)) and (ii) synchronized motion (\( \Delta \phi _{w} = 0 \)). (C) Video snapshots of the fluid flow induced by an artificial cilia array (\( \Delta \phi _{w} = -\pi/4 \)) visualized by a food dye. Each column in an array has three identical cilia in the z direction. (D) Sequence of fluid flow vorticity and velocity distributions produced by the bioinspired cilia array with an optimal metachronal wave within a full period. Fluid flow data were measured using PIV. \( \mathbf{B}(t) = 2.5 \text{ Hz and } B_{zm} = 40 \text{ mT} \). Scale bars, 1 mm. Photo credit: Xiaoguang Dong, Max Planck Institute for Intelligent Systems.
optimized environment for fertilization and early embryonic development (41).

In addition to the functionality of transportation, in Fig. 6 and movie S6, we demonstrate that optimal metachronal coordination can be encoded into two arrays of artificial cilia for efficient fluid mixing function. Mixing adjacent laminar flow completely and rapidly at low Re is difficult while important for many applications (25). At low Re (<0.1), passive mixing devices usually require specific design of channel geometries, using chemical reactions, limiting the applicable scenarios. Other active mixing devices, such as magnetic micropillars, have limited mixing thoroughness between vertical layers due to their cone-shaped stirring motions (28). Here, we demonstrate a mixing device capable of mixing viscous fluids thoroughly and rapidly by encoding optimal coordination into two artificial cilia arrays. The cilia in the bottom array in Fig. 6A have an optimal metachronal wave \( \Delta \psi = -\frac{\pi}{3} \) for inducing a directional flow. Meanwhile, the cilia in the top array are producing a transverse flow by periodically beating opposite to their faced cilia in the bottom array. Visualized by a dye, the viscous glycerol can be rapidly mixed with a coefficient of variation (25) less than 0.05 within 35 s in a channel across a volume of 43 mm\(^3\) at a local Re < 0.03, as shown in Fig. 6 (B and C).

Last, the wireless magnetic actuation of our fluidic devices combined with ultrasound medical imaging could allow remote pumping and imaging of body fluids in enclosed small spaces with no physical contact, potentially applicable in the human body in the future for biomedical applications. As a proof of concept, in Fig. 7 and movie S7, we demonstrate the ability of pumping body fluids, such as fresh whole mouse blood in narrow channels (see the “Preparation and characterization of fluids” section in Materials and Methods), showing the promising function of pumping biofluids ex vivo and in vivo in future biomedical applications. They could potentially be deployed by a medical catheter or locomote by remote actuation (16) to target sites in the human body for delicate and minimally invasive noncontact fluidic or object/cell manipulation (42). These devices have potential to be applied in health care for patients with specific ciliary dysfunctions, such as assisting in pumping mucus in human respiratory systems and transporting ovum in female fallopian tubes for improving fertilization (5).

**DISCUSSION**

We have developed bioinspired ciliary devices, which not only provide remarkable insights of how the time-varying boundaries of ciliary surfaces by different metachronal coordination could lead to different cooperatively generated ciliary flows but also use these findings to enable unprecedented optimized engineering applications. The
Fig. 6. Mixing viscous fluids efficiently and completely at low $Re$. (A) Mixing of viscous fluids by producing directional and transverse flows visualized by the dye. Cilia in the bottom array (cilia in columns c1 to c6, $\Delta \phi_x = \frac{-\pi}{3}$) have an optimal metachronal wave for creating a directional flow. Cilia in the top array (cilia in columns c7 to c12, $\Delta \phi_x = \frac{\pi}{3}$) are creating a transverse flow by beating oppositely to the faced cilia in the top array. Magnetization phase profiles: c1 to c6, $\phi_i(s) = \phi_0(s) + (i - 1) \cdot (-\frac{\pi}{3})$, $\alpha_i = \frac{\pi}{8}, i = 1 - 6$; c7 to c12, $\phi_i(s) = \phi_0(s) + (i - 7) \cdot \frac{\pi}{3} + \frac{\pi}{8}$, $\alpha_i = \frac{9\pi}{8}, i = 7 - 12$. In both arrays, $\phi_0(s) = -\frac{3\pi}{8} + 1.75s$. Photo credit: Xiaoguang Dong, Max Planck Institute for Intelligent Systems. (B) Snapshots of the mixing process in a sample region. The sample region is enclosed with yellow dashed lines marked in (A). $B(t)$: $f = 2.5$ Hz and $B_m = 40$ mT. 

Fig. 7. Pumping and imaging biofluids in enclosed channels. (A) The cilia array pumping in enclosed channels. (i) Schematics of the enclosed channel. (ii) Pumping fresh mouse blood visualized by ultrasound imaging. The color indicates flow velocity of mouse blood flow ($\bar{v}_x = 1.96$ mm/s and $\bar{Q}_x = 9.7$ mm$^3$/s). The red dots mark the same position in (i) and (ii). $B(t)$: $f = 7$ Hz and $B_m = 40$ mT. Contrast enhancement particles/microbubbles are used for enhancing ultrasound imaging quality. (iii) Pumping water visualized by dye ($\bar{v}_x = \sim 15$ mm/s and $\bar{Q}_x = 74.2$ mm$^3$/s). The yellow dots mark the same position in (i) and (ii). $B(t)$: $f = 2$ Hz and $B_m = 40$ mT. (B) Photo of an ultrasound imaging system (Vevo 3100, VisualSonics Inc.). (C) Photo of the experimental setup for actuating the artificial cilia and imaging fluid flow via an ultrasound probe. Ultrasound gel is used between the ultrasound probe and the top surface of the fluidic channel for enhancing the imaging signal. Photo credit: Xiaoguang Dong and Wenqi Hu, Max Planck Institute for Intelligent Systems.
reported design methodology is applicable in principle to scale the artificial cilia down to the size of real cilia (~10 μm in length). A scaling analysis (see the “Scaling analysis of single cilium dynamics” section in Materials and Methods) shows the feasibility of scaling down the overall size of ferromagnetic sheets to the micrometer scale. Microfabrication of ferromagnetic materials using recent advance of fabrication methods (43) could potentially allow manufacturing smaller artificial cilia arrays with metachronal waves in the future, although challenges of weaker magnetization, limited geometry, and limited material properties still need to be overcome.

Biological cilia are densely packed rod structures capable of non-reciprocal 2D or 3D motions (44), with a length scale of ~10 μm, beating at 5 to 30 Hz (7). In our system, instead of mimicking the biological cilia, we aim at developing artificial cilia with metachronal coordination similar to their biological counterparts, by keeping the critical dimensionless number (e.g., Re) similar. As the fundamental physical law is the same, we can study a scientific question in cilia mechanics: the function of the metachronal coordination in inducing fluid flows at low Re. With such a platform, we find that antiplectic waves can indeed enhance fluid flows within the scope of our investigation space. This finding agrees with that in biological systems (7, 36) and numerical studies reported before (12, 13), bringing extra insights into the ciliary biological systems.

In addition, the developed artificial cilia array with programmable nonreciprocal motion and metachronal coordination could be used as a scientific tool to further investigate many interesting scientific questions, such as how the fluid flow patterns would be different when varying the cilia beating dynamics (45), boundary geometries (38), location of topological defects (6), and the conditions of the surrounding fluidic environment (46). We anticipate our design method and developed soft miniature devices that could open a wide range of unprecedented opportunities for studying ciliary biomechanics and creating cilia-inspired wireless microfluidic pumping, object manipulation and lab- and organ-on-a-chip devices, mobile micro-robots, and bioengineering systems.

MATERIALS AND METHODS

Fabrication of ferromagnetic cilia

The ferromagnetic-elastic sheet–based artificial cilia were made of composite materials of silicone rubber (Ecoflex 00-30, Smooth-On Inc.) and NdFeB hard magnetic microparticles (average diameter, 5 μm; MQFP-15-7, Neo Magnequench) with a weight ratio of 1:1. The two materials were mixed together and then poured on to a poly(methyl methacrylate) substrate with tapes as spacers for controlling the thickness to be 100 μm. The top surface of the mixed material was scraped by a single-edge razor blade (fig. S1A). The composite material was cured in room temperature for about 4 hours or cured in 1 hour by putting it on a hot plate at 60°C. After curing, as illustrated in fig. S1B, the ferromagnetic cilia were cut by a laser machine (LPKF ProtoLaser R3, LPKF Laser & Electronics AG) according to a designed dimension of 1 mm by 550 μm by 100 μm (L × w × t0) marked in fig. S1C. Other dimension parameters in fig. S1C are given by L1 = 300 μm, w1 = 850 μm, and w2 = 200 μm. The material has a density of (2.00 ± 0.056) × 10³ kg·m⁻³ and an average Young’s modulus of 144 kPa measured by a tensile testing machine (5940 series, Instron GmbH). In the experiments, we observed that the performance of the device is quite robust without any degradation issue. The dynamics of the artificial cilia array do not change even several months after its first use, when actuated with the same control signals in the same fluids. According to a previous study, structures made of Ecoflex 00-30 can maintain their mechanical properties after being subjected to cyclic loads for over 10⁶ cycles (47).

Simultaneous encoding of cilia magnetization profiles

We used a jig-assisted method reported in our previous work (18) but extended it to simultaneously encode the desired different M(s) of multiple ferromagnetic cilia using one magnetizing jig. Figure S1D illustrates the principle and process to encode the desired different M(s) on multiple cilia simultaneously. To obtain a desired magnetization profile, multiple ferromagnetic cilia are prebent into a jig with multiple cutout parts in designed shapes. They are then magnetized by applying a large magnetizing B (magnitude, 1.2 T) in the +x direction inside a vibrating sample magnetometer (EZ7 VSM, MicroSense LLC). The shape of the cutout part corresponding to each cilium is parameterized with a local slope angle profile φ(i)(s), which is the tangent angle at each element along the middle line in the cutout part (fig. S1D). To encode φ(i)(s) = φ(i)0(s) + (i − 1)Δφ for a given cilium, we have θ(i)0(s) = θ(i)φ(s). For multiple cilia, to encode desired M(s) in an array simultaneously, the inverse design of the cutout part of the magnetizing jig is thus given by

\[
x_i(s) = x_0 + \int_0^1 \cos(\phi_i(s) + (i - 1) \Delta \phi) \, ds
\]

\[
y_i(s) = y_0 + \int_0^1 \sin(\phi_i(s) + (i - 1) \Delta \phi) \, ds
\]

Figure S3 shows that we encoded the desired M(s) by comparing the measured and predicted magnetic fields produced by a magnetized single cilium and array of cilia at their surfaces.

Assembly of artificial cilia arrays

Figure S1 (E and F) illustrates the procedure of assembling ferromagnetic cilia into an array. The magnetized cilia were manually attached to an assembly fixture and fixed by glue (Loctite 401, Henkel AG) under a stereomicroscope (Stemi 508, Carl Zeiss AG). Motorized micromanipulators could be potentially incorporated for automating such a task in the future.

Magnetic actuation of ferromagnetic cilia

As shown in fig. S2 (A and B), a rotating Halbach array composed of 12 cubic ferromagnets (12 mm by 12 mm by 12 mm; N42, Super-magnet.de) was used for generating a rotating uniform magnetic field. The rotational motion was actuated by a dc motor (Parallax Continuous Rotation Servo, Parallax Inc.) and controlled by an embedded controller (Arduino Uno, Arduino.cc). B(t) was characterized in fig. S2 (C to E). The magnetic field strength was about 40 mT (at 15 mm above the surface of the magnet array). On this plane, the uniformity of the magnetic field in the workspace was ~95% within a circular area of 10 mm in diameter, which covered the whole artificial cilia array in the experiments. The artificial cilia array was fixed in a container (43 mm by 32 mm by 11 mm; fig. S2F) filled with glycerol (99.8% volume ratio) and supported on a customized sample holder above the Halbach array. We used glycerol as a viscous liquid in the experiments because it has a high viscosity to create a low Re environment and does not cause swelling of the elastic cilia. The fluid had a density of 1.257 × 10³ kg·m⁻³ and a dynamic viscosity of 0.876 Pa·s at the room temperature 25°C.
General design methodology

Summary of the used coordinate systems

Three coordinate systems in this work are presented in fig. S2 (A and G) and summarized as below. First, the global coordinate system \([G_b]\) \((x_b, y_b, z_b)\) is located at the center of the Halbach array for expressing the global actuation magnetic field and location of the artificial cilia arrays. Second, the array-attached coordinate system \([L_a]\) \((x, y, z)\) is located at the base of the artificial cilia array, which is used to express the dynamics of artificial cilia, fluid flow, and particle transportation. Last, the cilium-attached coordinate system \([L_c]\) \((X, Y, Z)\) of the \(i\)-th cilium in an array is located on the body of the cilium, which is used to describe its relative orientation and position within an array and to express the magnetization profile of each cilium.

FSI model of single-cilium dynamics

An FSI model in 2D is presented to guide the design of single ferromagnetic-elastic cilia. The motion of a given cilium is modeled using the Euler-Bernoulli beam theory for large deflections \(18\) while considering fluid drags. At quasi-static states, the governing equations are given below for a cilium with one end fixed to the boundary wall.

The cilia motions are assumed to be planar by imposing a large width-to-length ratio \((w/L = -0.6)\) to minimize the out-of-plane twisting and bending motions. The single-cilium oscillating beating dynamics under a rotating \(B(t)\) can be predicted using the above model as shown in fig. S8.

Optimizing a single cilium by maximizing the SAR

The SAR is optimized on the basis of the discussed FSI model in the above section. The dynamics of a single cilium in viscous fluids is a function of \(M(s)\), its physical dimensions, and \(B(t)\). The 2D magnetization profile along the \(i\)-th cilium with a length \(L\), is given by

\[
M_i(s) = M(s)\cos\phi_i(s)
\]

We first fix the magnitude of \(M(s)\), as \(40 \text{ kA/m}^{-1}\) and assume that the magnitude of external magnetic field is \(B_m = 40 \text{ mT}\). Then, for simplicity, \(\phi_i(s)\) is represented in a general continuous polynomial function with a first-order approximation as \(\phi_i(s) = \phi_i(0) + \phi_i(0) - \phi_i(L) \left( \frac{s}{L} \right)\) since the high-order terms are less important in determining the cilia dynamics. As shown in fig. S5 (A and B), we carried out a systematic parameter sweeping for the total phase span \(\phi_i(0) - \phi_i(L)\) of a single cilium. We found that a maximal positive SAR could be obtained when \(\phi_i(0) - \phi_i(L) = 1.75\pi\) (used in all experiments not explicitly mentioned) and a maximal negative SAR could be obtained when \(\phi_i(0) - \phi_i(L) = 0\) (used in Fig. 5). These two magnetization profiles are optimal within a practical range \((-1.75\pi\) to \(1.75\pi\)) constrained by the template-based magnetizing method.

The magnitude \(B_m\) and frequency \(f\) of \(B(t)\) are another two control inputs of the dynamics of a single cilium. Increasing \(f\) will increase SAR, but the maximal \(Re\) will also increase (fig. S5C). Therefore, we constrained \(f\) to ensure that our cilia arrays operate at the low \(Re\) regime in the experiments. In addition, a larger \(B_m\) yields a larger SAR (fig. SSD) until the cilia’s deformation is limited by the boundary walls.

In addition, the dimensions of the artificial cilium also affect the motion. We defined two ratios, the thickness-to-length ratio \(r_{th}\) and the width-to-length ratio \(r_{wl}\), to explain their effect based on our models. \(r_{th}\) has a major effect on the magnitude of the cilium motion. With the same actuation magnetic fields, artificial cilia with a larger \(r_{th}\) will exhibit a smaller maximal bending angle and a smaller SAR, as shown in fig. S9, while the buckling motion is more evident, yielding a larger output force during the power stroke after being deformed.
Mechanics and design methodology of metachronal waves

Each member within an array of ferromagnetic cilia is a distributed oscillator subject to a rotating magnetic field \( B(t) = R_2(t)B(0) \). The phases of these oscillators can be represented by their body slope angles \( \theta_i(s, t) \). For two neighboring cilia, we have the following moment-balancing equations:

\[
-EL\frac{\partial^2 \theta_i(s, t)}{\partial s^2} = \tau_{\text{mag}}^{\text{st}}(\theta_i(s, t)) + \left[ F_i(s, t) \cos \theta_i(s, t) - F_i(s, t) \sin \theta_i(s, t) \right]
\]

(12)

\[
-EL\frac{\partial^2 \theta_{i+1}(s, t)}{\partial s^2} = \tau_{\text{mag}}^{\text{st}}(\theta_{i+1}(s, t)) + \left[ F_i(s, t) \cos \theta_{i+1}(s, t) - F_i(s, t) \sin \theta_{i+1}(s, t) \right]
\]

(13)

The temporal differences in the time-varying and spatially distributed external magnetic torques \( \tau_{\text{mag}}^{\text{st}}(s, t) \) create the metachronal waves. To be more specific, the external magnetic torques applied on the \( i \)-th and \( (i+1) \)-th cilium are given by

\[
\tau_{\text{mag}}^{\text{st}}(s, t) = n_z \cdot A_{\text{cross}} \left[ R_d(\theta_i(s, t)) M_i(s) \right] \times \left[ R_z(\omega t) B(0) \right],
\]

(14)

\[
\tau_{\text{mag}}^{\text{st}}(s, t) = n_z \cdot A_{\text{cross}} \left[ R_d(\theta_{i+1}(s, t)) R_d(\Delta \phi) M_i(s) \right] \\
\times \left[ R_z(\omega t) B(0) \right] = n_z \cdot A_{\text{cross}} \left[ R_d(\theta_{i}(s, t)) M_i(s) \right] \times \left[ R_z(\omega t + \Delta \phi) B(0) \right]
\]

(15)

The phase shift \( \Delta \phi \) in \( M(s) \) can be equivalently represented as \( \Delta \phi \) in \( B(t) \) due to the mathematical properties of the cross product in 2D as shown in Eq. 15. Therefore, the phase shifts \( \Delta \phi = \frac{\Delta \phi}{2\pi} T \) are equivalently produced in the oscillating motions of neighboring cilia.

CFD simulations in comparison with experimental data

To model the fluid flow created by the artificial cilia arrays, we use a 3D bidirectional CFD model (15), which can accurately describe the magnetic torque on the cilia, the deformation of the cilia, and the deformation-induced fluid flow at low \( Re \) [see fig. S6 (A and B)]. The model accounts for the fully coupled bidirectional solid-fluid interaction between cilia and fluid, including the cilia deformation induced by the fluid. The computational framework is briefly summarized here. In the CFD model, the cilia are represented by an assemblage of shell elements, which act as an internal boundary to the fluid. The FSI is considered by implicitly coupling the fluid dynamics and solid mechanics equations, where the Stokeslet method is used to account for the viscous drag and implemented using a boundary-element approach. To model the ciliary deformation, we use shell elements with the bending and membrane behavior accounted for by using triangular Kirchhoff elements (50) and constant strain triangles with drilling degrees of freedom (51), respectively. The large and geometrically nonlinear deformation of the cilia is modeled by adopting an updated Lagrangian approach. We focus on flows at low \( Re \) and use Green’s functions (52) to simulate fluid motion in the Stokes regime. The solid and fluid are implicitly coupled through a no-slip boundary condition.

We compared the effect of the phase shift \( \Delta \psi \), (fig. S6C) and intercilium spacing \( d_i \) (fig. S6D) on inducing fluid flow in the simulated and experimental scenarios. The simulations also show that the antiplectic waves produce a maximal directional fluid flow when \( \Delta \psi = \left\{ \frac{\pi}{2}, \frac{3\pi}{2} \right\} \) and artificial cilia arrays with a denser spacing generate a larger \( Q \). Although a discrepancy exists in the absolute quantitative values due to the simplified assumption and approximated parameters of the CFD model, as well as experimental uncertainties, the simulation results can support our findings in experiments in terms of the overall trends.

Encoding metachronal waves in cilia arrays on curved surfaces

The metachronal coordination can be encoded into cilia arrays planted on a curved boundary wall using the design rule of \( \Delta \phi = \Delta \phi + \Delta \alpha \). Here, \( \Delta \alpha \) is the variation of the slope angles along the curved surface by defining the rotational difference between the cilium-attached coordinates \( [L_{s(i+1)}] \) and \( [L_i] \) of two neighboring cilia. The distributed magnetic torques induced on these cilia have a relationship of \( \tau_{\text{mag}}^{\text{st}}(s, t) = \tau_{\text{mag}}^{\text{st}}(s, t + \Delta \phi) \). Therefore, the antiplectic phase profiles of two neighboring cilia in the global frame are given by

\[
\phi_i(s) = \phi_i(s) + (i - 1) \Delta \phi - \alpha_i
\]

(16)

\[
\phi_{i+1}(s) = \phi_{i+1}(s) + i \Delta \phi - \alpha_{i+1}
\]

(17)

To create metachronal waves on arbitrary curved surfaces, \( \phi_i(s) \) of each cilium is inversely designed together with \( [L_i] \) as

\[
\phi_i(s) = \phi_i(s) + (i - 1) \Delta \phi - \alpha_i
\]

(18)

\[
\alpha_i = \alpha_0 + \tan(dy_{ci}/dx_{ci})
\]

(19)

where \( \alpha_0 = \frac{\pi}{8} \) is the angle between the +X axis of \( [L_i] \) relative to the surface tangent vector direction along the boundary wall surface. Specifically, in the bioinspired cilia array in Fig. 4, \( \Delta \phi = \frac{\pi}{2}, \phi_0(s) = \frac{-\pi}{4} + 1.75 \pi(s - \frac{1}{5}) \), and \( y_{ci} = L \sin(\frac{2\pi c_i}{5L}) \), where \( x_i (i = 1, 2, \ldots, 8) \) was chosen with an equal spacing from \(-4L \) to \( 4L \).

Scaling analysis of single-cilium dynamics

We first define \( VE \) to quantify the relative scaling of viscous drag and elastic stress on an infinitesimal element, which is given by

\[
VE = \frac{\text{Viscous drag}}{\text{Elastic stress}} = \frac{C_{D1} \frac{1}{2} \rho |\vec{v}|^2 w_{ci} ds}{E Ids \frac{\partial^2 \theta_i(s, t)}{\partial z^2}} \approx \frac{6C_{D1} \rho \alpha^2}{EL} \frac{\partial^2 \theta_i(s, t)}{\partial z^2}
\]

(20)

where the derivative of cilium curvature \( \frac{\partial^2 \theta_i(s, t)}{\partial z^2} \) is scaling to \( L^{-2} \). This equation shows that the viscosity of the fluid needs to be decreased for reducing the damping effect, \( C_{D1} \), when \( L \) scales down.

We then define \( ME \) to quantify the relative scaling of magnetic torque and elastic stress on an infinitesimal element, which is given by

\[
ME = \frac{\text{Magnetic torque}}{\text{Elastic stress}} = \frac{MBw_{ci} ds}{E Ids \frac{\partial^2 \theta_i(s, t)}{\partial z^2}} \approx \frac{12MB}{EL} \frac{\partial^2 \theta_i(s, t)}{\partial z^2}
\]

(21)

This equation shows that the same \( M(s) \) and \( B(t) \) can induce similar elastic stress within the cilium when \( L \) scales down.

One practical limitation of our system is that the intercilium spacing \( d_i \) cannot be less than a body length. As \( d_i \) further decreases, the dynamics of artificial cilia may be affected by magnetic local interactions. The critical distance between neighboring cilia is a complex
function of the magnetization profile $M(s)$, the material stiffness, and the magnitude of external magnetic fields $B(t)$. To alleviate such an effect, it is found that we could reduce the magnitude of $M(s)$, increase the material Young's modulus $E$, or increase the magnitude of $B(t)$. More research on this critical distance will be carried out in the future.

**Fluid flow measured by PIV**

A PIV system (Dantec Dynamics, Dantec Inc.) was integrated with the magnetic actuation setup for visualizing and quantifying the induced fluid flows (fig. S1A). The fluorescent tracking particles (Cospheric Inc.) used in the PIV experiments had an average diameter of 2 μm. A high-speed camera (Phantom MicroLab 140, Vision Research, Ametek Inc.) with a framerate up to 5000 frames per second (fps) was used for recording the dynamics of cilia and the motion of tracing particles. The acquired image sequences were further analyzed using the software DynamicStudio 6.1 (Dantec Dynamics, Dantec Inc.) and customized codes implemented in MATLAB 2017a (MathWorks Inc.). The average speeds from the video sequences using color-based filtering algorithms implemented in MATLAB R2017a (MathWorks Inc.) were used for recording the trajectories of particles at about 30 fps.

**Particle transportation using cilia arrays**

Fluorescent polyethylene microspheres (1.20 g·cm⁻³, 300 to 355 μm in diameter) were used as neutrally buoyant tracer particles because their densities are close to glycerol (1.257 g·cm⁻³), so that the particle’s motion reflects the fluid velocity by minimizing the effect from the gravity and buoyancy. Two cameras (Flea 3 USB3, FLIR Systems Inc.) were used for recording the trajectories of particles at about 30 fps. Particles trajectories were extracted by segmenting particle contours from the video sequences using color-based filtering algorithms implemented in MATLAB R2017a (MathWorks Inc.). The average speeds of particles were calculated along a trajectory with a displacement of 7 mm in the $+x$ direction (Fig. 2G). The SDs were for $n = 5$ trials using each artificial cilia array. For each trial of the experiment, the particle was released from the same position.

**Preparation and characterization of fluids**

The fluids used for pumping demonstrations included deionized water, mouse blood, synthetic mucus, and syrup. Fluids of low viscosity include fresh mouse blood (Innovative Grade US Origin Mouse Blood; Fig. 7A) and deionized water (Fig. 7B). The mouse blood is shear-thinning with a dynamic viscosity of ~10 mPa·s at a shear rate of 2.4 s⁻¹ (53). Viscous fluids in addition to glycerol include synthetic mucus (synthetic nasal mucus, Kryolan Professional Make-up GmbH; Fig. 5C) and syrup (rice syrup, Serapis Culinary; Fig. 5D). Figure S10 shows the rheology properties of the synthetic mucus and syrup. The rheology test is carried out in a Discovery HR-2 rheometer from TA Instruments Inc.

**Supplementary materials**

Supplementary material for this article is available at http://advances.sciencemag.org/cgi/content/full/6/45/eabc9323/DC1

**Statistical analysis**

The SDs are indicated in the figure captions.

**REFERENCES AND NOTES**


Acknowledgments: We thank all members of the Physical Intelligence Department at the Max Planck Institute for Intelligent Systems for their comments. Funding: This work is funded by the Max Planck Society and European Research Council (ERC) Advanced Grant SoMMoR project with grant no. 834531. Author contributions: X.D., G.Z.L., W.H., and M.S. proposed the research. X.D. designed the experiments. X.D., G.Z.L., W.H., and M.S. performed the CFD simulations. M.S. planned and supervised the research. X.D., W.H., and M.S. wrote the manuscript with input from all authors. All authors commented on or edited the manuscript. Competing interests: The authors declare that they have no competing interests. Data and material availability: The modeling and design of the single-cilium dynamics was performed using customized codes implemented in MATLAB R2017a (MathWorks Inc.). The CFD analysis was performed using customized codes using FORTRAN (IFORT 13.1.2). These codes are available upon reasonable request to the corresponding author. All data needed to evaluate the conclusions in the paper are present in the paper and/or the Supplementary Materials. Additional data related to this paper may be requested from the authors.

Submitted 21 May 2020
Accepted 21 September 2020
Published 6 November 2020
10.1126/sciadv.abc9323

Citation: X. Dong, G. Z. Lum, W. Hu, R. Zhang, Z. Ren, P. R. Onck, M. Sitti, Bioinspired cilia arrays with programmable nonreciprocal motion and metachronal coordination. Sci. Adv. 6, eabc9323 (2020).
Bioinspired cilia arrays with programmable nonreciprocal motion and metachronal coordination
Xiaoguang Dong, Guo Zhan Lum, Wenqi Hu, Rongjing Zhang, Ziyu Ren, Patrick R. Onck and Metin Sitti

Sci Adv 6 (45), eabc9323.
DOI: 10.1126/sciadv.eabc9323

ARTICLE TOOLS
http://advances.sciencemag.org/content/6/45/eabc9323

SUPPLEMENTARY MATERIALS
http://advances.sciencemag.org/content/suppl/2020/11/02/6.45.eabc9323.DC1

REFERENCES
This article cites 52 articles, 13 of which you can access for free
http://advances.sciencemag.org/content/6/45/eabc9323#BIBL

PERMISSIONS
http://www.sciencemag.org/help/reprints-and-permissions

Use of this article is subject to the Terms of Service

Science Advances (ISSN 2375-2548) is published by the American Association for the Advancement of Science, 1200 New York Avenue NW, Washington, DC 20005. The title Science Advances is a registered trademark of AAAS.

Copyright © 2020 The Authors, some rights reserved; exclusive licensee American Association for the Advancement of Science. No claim to original U.S. Government Works. Distributed under a Creative Commons Attribution NonCommercial License 4.0 (CC BY-NC).