Designing maneuverable micro-swimmers actuated by responsive gel†

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We use computational modeling to design a self-propelling micro-swimmer that can navigate in a low-Reynolds-number environment. Our simple swimmer consists of a responsive gel body with two propulsive flaps attached to its opposite sides and a stimuli-sensitive steering flap at the swimmer front end. The responsive gel body undergoes periodic expansions and contractions leading to a time-irreversible beating motion of the propulsive flaps, which propels the micro-swimmer. We examine the effects of body elasticity and flap geometry on the locomotion of the swimmer and show how they can be tailored to optimize the swimmer propulsion. We also probe how the swimmer trajectory can be controllably changed using the steering flap that bends when exposed to an external stimulus. We demonstrate that the steering occurs due to two effects: steering flap bending and periodic beating. Furthermore, our simulations reveal that the turning direction can be regulated by changing the stimulus strength.

Introduction

The development of swimming micro-devices that are able to carry payloads and navigate autonomously through highly viscous fluidic environments has the potential to transform drug delivery systems, lab-on-a-chip devices, and open new approaches for performing microsurgery and micro/nano-fabrication.1–3 Advances in MEMS technology have stimulated researchers to design various biomimetic devices and systems with active propulsion that are capable of performing complex functions at the microscale.4–12 However, despite recent notable progress in creating micro-swimmers, the development of miniature self-propelling swimming robots that can effectively navigate through microchannels remains a challenge.2,11

Herein, we use computational modeling to design a synthetic micro-swimmer that controllably navigates through highly viscous (low-Reynolds-number) environments. Our simple swimmer consists of a cubic gel body which periodically changes volume in response to external stimuli, two rigid rectangular flaps bonded to the opposite sides of the gel body, and a flexible steering flap attached to the front end of the swimmer body (see Fig. 1a). The stimuli-sensitive body undergoes periodic expansions (swelling) and contractions (deswelling), which can be triggered, for example, by an oscillatory chemical reaction,14–16 by oscillating magnetic and electric fields,17–21 or by temperature changes.22–27 The time-periodic volumetric changes of the body lead to a time-irreversible beating motion of the propulsive flaps. This periodic flap motion, in turn, propels the micro-swimmer unidirectionally through the inertialless fluid. Thus, the responsive gel body acts as an “engine” actuating the motion of the swimmer. To direct the swimmer motion towards a specific target or location, we use a stimuli-sensitive steering flap that bends in response to changes in the environmental conditions such as light intensity, temperature, or magnetic field.28–34

In our simulations described below, we first probe the unidirectional self-propelling motion of the gel swimmer in a micro-channel. Specifically, we examine how the swimming speed depends on the flap size and on the elasticity of the responsive gel. We then investigate how the swimmer turns in the desired direction when the steering flap changes its shape due to an external stimulus. Finally, we discuss relevant experimental parameters that should be considered in order to design an effective steerable micro-swimmer.

Methodology

To capture the dynamics of the swimmer and the surrounding fluid we employ a hybrid computational approach,35–40 which integrates the lattice Boltzmann model (LBM) for incompressible Newtonian fluids and the lattice spring model (LSM) for elastic solids and gels.

The LBM is a mesoscale lattice-based model that consists of two processes: the propagation of fluid “particles” to neighboring lattice sites and collisions between particles when they reach a site. The fluid is described in terms of a single particle velocity distribution function $f(r,t)$ representing the mass density of fluid particles with velocity $v$ at a lattice node $r$ at time $t$. The distribution function $f_i$ is a continuous variable, whereas $f_v$, $r$, and
Fig. 1  Panel (a) shows a schematic of gel-actuated micro-swimmer. Panel (b) shows snapshots illustrating swimmer motion during one period of the gel body oscillations. Colors in panel (b) represent the material strain $\epsilon$. Green cones represent velocity vectors in the $x-y$ plane through the middle of the swimmer body (also see ESI†). The swimmer parameters are $AR_p = 2$, $AR_s = 1$, and $\epsilon = 3$. 

$t$ are discrete variables. Moments of the distribution function give the hydrodynamic quantities: the fluid density $\rho = \sum_i f_i$, the momentum $j = \rho u = \sum_i c_i f_i$ with $u$ being the local fluid velocity, and the stresses $\Pi = \sum_i c_i e_i f_i$. 

The time evolution of the distribution function is governed by the discretized Boltzmann equation $f_j(r + c_i \Delta t, t + \Delta t) = f_j(r, t) + O[f(r, t)]$.41 The collision operator $O[f(r, t)]$ accounts for the change in $f_j$ due to instantaneous collisions at the lattice nodes, and its action depends on all $f_i$ at a node $r$ denoted by $f(r, t)$. We employ a multi-relaxation-time collision operator42 that conserves mass and momentum and relaxes stresses towards a local equilibrium.

The velocity $e_i$ in the $i$th direction is chosen such that fluid particles propagate from one lattice site to the next in one time step $\Delta t$. In our three-dimensional model, we use 19 velocities of the distribution function that correspond to propagation to the nearest- and next-nearest-neighboring nodes of a simple cubic lattice as well as to the rest particle. In our simulations, we set the distance between the neighbor LBM nodes $\Delta x_{1B}$ equal to unity, fluid density to $\rho = 1$, and viscosity to $\mu = 1/6$, which yield the speed of sound $c_s = 1/\sqrt{3}$.44,45 Unless specified otherwise, all dimensional values are expressed in lattice Boltzmann units.

The LSM is also a lattice-based model in which an elastic solid is modeled as a network of harmonic springs connecting mass points, whose positions are updated by integrating Newton’s equation of motion.44,45 We use a three-dimensional cubic lattice (see Fig. 1a) to model the elastic swimmer body. In this case, the elastic energy associated with a node at position $r_i$ is $E_i(r_i) = \frac{1}{2} \sum_{j=1}^{18} k_s (r_{ij} - r_{ij0})^2$, where the summation is over all the nearest- and next-nearest-neighboring nodes.44,45 The elastic forces acting on each node are then obtained using $F_i = \partial E_i / \partial r_{ij}$. Here, $r_{ij} = r_i - r_j$ is the vector representing the spring between two nodes at positions $r_i$ and $r_j$ with length $r_{ij} = |r_i - r_j|$, $r_{ij0}$ is the equilibrium length of stretching springs, and $k_s$ is their spring constant. This spring arrangement results in a Poisson’s ratio $\nu = 1/4$ and Young’s modulus $E = 5k_s / 2\Delta x_{1S}$, where $\Delta x_{1S}$ is the spacing between the LSM nodes.46 We change the magnitude of the spring constant $k_s$ to model swimmers with different body elasticity $E$.

In order to model the gel body periodic expansion and contraction, we dynamically change the equilibrium length of the lattice springs according to a sinusoidal law $r_{ij0} = r_{ij0}^0 [1 + \alpha \sin(\omega t)]$, where $r_{ij0}^0$ is the initial equilibrium value, $\alpha = 0.25$ is the amplitude of oscillations, and $\omega = 2\pi / T$ is the angular velocity of oscillations with $T = 5 \times 10^3$ being the oscillation period. In this way, we impose internal stresses in the gel leading to periodic volumetric changes of the swimmer body.

To model rigid propulsive flaps with thickness that is much smaller than other dimensions, we use a two dimensional square lattice with fixed relative distances between the nodes so that the flaps follow the dynamics of a rigid body (see Fig. 1a). The propulsive flaps are attached to the gel body via nodes on gel body sides. Thus, these nodes are shared between the flaps and gel body and the gel sides are confined by rigid flaps. This mimics a perfect attachment between rigid flaps and elastic gel body.

We use a two-dimensional triangular lattice to model the flexible steering flap (see Fig. 1a). Each node in the triangular lattice is bonded to six nearest neighbors by stretching springs with constant $k_s$. A three body interaction with bond-bending spring with constant $k_b$ is added among every three consecutive nodes that are initially inline. The stretching and bending springs give rise to, respectively, the in-plane stiffness and bending rigidity.46

The energy associated with each node in the triangular lattice takes the form $E_i(r_i) = \frac{1}{2} \sum_{j=1}^{6} k_s (r_{ij} - r_{ij0})^2 + \sum_{j=1}^{6} k_b \left( \frac{r_{ij} \cdot r_{ij} - r_{ij0}^2}{r_{ij}^2} + 1 \right)$ which leads to a Poisson’s ratio $\nu = 1/3$, Young’s modulus $E = 2k_s / \sqrt{3}$, and the bending modulus $EI = \frac{3\sqrt{3}}{4} k_b w_0 (1 - r^2)$, where $I = w_0 L^2 / 12$ is the moment of inertia with $w_0$ and $L$ being the width and length of the rectangular flaps, respectively (Fig. 1a). The node $f$ in the second term represents the neighbor opposite to the neighbor $j$.46,47

To model stimulus-induced bending of the steering flap, we apply a uniformly distributed internal moment $M_z$ to the steering
flap that mimics internal stresses arising in the flap material in response to an external stimulus. Here, $M$ is the magnitude of internal moment and $\hat{z}$ is the unity vector in the $z$ direction. We impose the moment by applying forces $F_j = -M \times r_j / n_j \hat{r}_j$, $F_f = -M \times r_f / n_f \hat{r}_f$, and $F = -(F_f + F_j)$ to consecutive nodes $j$, $i$, and $f$ initially aligned along the $x$ direction. Here, $n_j = 1 + 2 w / \sqrt{3} \Delta x_{LS}$ is the number of flap nodes in the $z$ direction. Thus, the net force and the net moment applied to the steering flap are zero.

To bond the steering flap to the swimmer body, the flap is extended one lattice row with spacing $\Delta x_{LS}$ into the gel body. In other words, the flap and the body share two rows of nodes: one row on the body surface and another row inside the body. In this way, a clamped boundary condition is imposed at the root of the steering flap.

The coupling between the LBM and LSM is done using two different techniques for, respectively, the swimmer body and the flaps. When the porous gel body of the swimmer changes its volume, the surrounding solvent penetrates inside or is expelled from the gel. To mimic this property and to allow the fluid to pass through the LSM network, we use a coupling via a frictional force between the LBM fluid and the LSM nodes representing the gel body. The frictional force is in the form of the Stokes drag force between the LBM fluid and the LSM nodes representing the gel body. The frictional force is $F_{\text{friction}} = -\xi (u_i - u(r_i))$, where $u_i$ is the velocity of the $i$th node, $u(r_i)$ is the interpolated fluid velocity at position of the node $r_i$, and $\xi = 6 \pi \mu a$ is the drag coefficient with $\sigma$ being the effective hydrodynamic radius of the nodes. Here, we set $\sigma$ that matches 80% porosity of the gel swimmer body. We apply the force $F_{\text{friction}}$ to the corresponding LSM node and the opposing force to the LBM fluid surrounding the node using a local body force.

The interaction between the fluid and solid flaps is modeled by applying the interpolated bounce-back condition to the distribution functions $f_i$ that intersect the swimmer flaps. In this manner we impose no-slip and no-penetration boundary conditions at the solid–fluid interface between the flaps and the surrounding viscous fluid.

The size of our computational domain is $100 \times 75 \times 50$ in the $x$, $y$, and $z$ directions, respectively. We use periodic boundary conditions in the $x$ and $y$ directions, and apply solid wall boundaries in the $z$ direction using the bounce-back rule. The cubic swimmer body with initial side $w_0 = 10$ is formed from $7 \times 7 \times 7$ LSM nodes with internode distance $\Delta x_{LS} = 5/3$ (Fig. 1a).

The flaps have size $7 \times n$ LSM nodes with $\Delta x_{LS} = 5/3$ and $\Delta y_{LS} = 10/\sqrt{27}$ for propulsive and steering flaps, respectively. We vary $n$ to probe the dynamics of swimmers with flaps of different aspect ratio $\text{AR} = L / w_0$ where $w_0 = 10$ is the flap width, which is equal to the side of the swimmer body at $t = 0$. Here, the flap length $L$ is equal to $L_p = 5(n - 1)/3$ and $L_o = 10(n - 1.5)/\sqrt{27}$ for propulsive and steering flaps, respectively.

Our choice of parameters leads to a flap oscillation Reynolds number $\text{Re} = an_o \rho \sigma / \mu$ equal to 0.188 with $\sigma = \mu / \rho$, which indicates that the effect of inertial forces on the fluid dynamics is minor. To evaluate the effect of fluid inertia on the swimmer propulsion, we have lowered the Reynolds number to 0.094 by doubling the period of gel oscillations while keeping other dimensionless parameters constant. Our simulations show less than 3% change in the results indicating that viscous forces are dominant and the effect of fluid inertia is indeed negligible.

**Results and discussion**

### 1. Swimmer propulsion

We first examine the self-propelled motion of our synthetic micro-swimmer. Fig. 1b shows a sequence of snapshots for one period of body oscillations in which the propulsive flaps move inwards and outwards relative to their initially parallel position. At a low-Reynolds-number environment, net propulsion arises when the swimmer undergoes a time-irreversible, non-reciprocal motion.

In the case of our swimmer, such motion is produced due to the coupling between elasticity of the periodically expanding and contracting gel body and fluid stresses on the rigid flaps. Specifically, when the body swells and expands in volume, the propulsive flaps move outwards and rotate towards the swimmer symmetry plane. When the body shrinks, on the other hand, the flaps move inwards and rotate in the opposite direction. These periodic translational and rotational movements of the flaps have a phase lag, which makes the motion non-reciprocal and leads to a unidirectional propulsion.

For a fixed frequency and amplitude of the swimmer body volume changes, the time-irreversible motion of the flaps and, consequently, swimmer velocity are defined by the elasticity of the gel body and the flap length. In Fig. 2, we plot the swimmer velocity versus body stiffness $\epsilon$.
propulsive characteristics versus the dimensionless gel elasticity $\varepsilon = ET/\mu V_\text{c} = T/\tau$ for flaps with different aspect ratio $\text{AR}_p$. Here, $\phi$ is the gel permeability estimated as $\phi^2$ (ref. 53 and 54) and $\tau = \mu E/\text{Ef}$ is the gel relaxation time.\textsuperscript{85-87} The dimensionless swimming velocity is $V = u/V_\text{c}$, where $u$ is the period-averaged velocity of the swimmer center of mass and $V_\text{c} = aw_0\phi$ is the characteristic velocity of the body change. The dimensionless power dissipated by the swimmer is $P = W/T\mu_0 V_\text{c}^2$ and the efficiency is $\eta = \mu_0 w_0^2 T W = V^2 IP$. Here, $W$ is the work done by the fluid on the swimmer during one oscillation period $T$, and $\mu_0 V_\text{c}^2$ estimates the power required to move the swimmer with the velocity $V_\text{c}$.

Fig. 2a shows how the dimensionless swimming velocity $V$ changes when we vary the elasticity of the gel body. When the gel is soft with $\varepsilon < 0.1$, the swimmer cannot generate any propulsive motion and has a nearly zero velocity. When we increase the body stiffness, the swimmer moves forward and its velocity increases with increasing $\varepsilon$ until it reaches the maximum. When the body stiffness is further increased, the propulsion slows down and nearly vanishes when $\varepsilon > 100$. Soft swimmers with $\varepsilon \sim 0.1$ exhibit a slow backward motion when the propulsive flaps are relatively long yielding $\text{AR}_p > 2.5$. The fastest backward swimming however is an order of magnitude slower than the maximum forward velocity.

We find that the maximum swimmer speed depends on the flap aspect ratio. It is the fastest when $\text{AR}_p = 2$ and slightly decreases when the flaps are shorter or longer. To further characterize the swimmer propulsion, we calculated the power required to drive the swimmer (Fig. 2b). It is interesting that the power is not sensitive to the flap aspect ratio when the body is softer than $\varepsilon < 10$. It means that most of the work is done on expanding and contracting the gel body which has the same geometry for all our swimmers, whereas the contribution of the work done on the flaps is minor. On the other hand, for stiffer swimmers with $\varepsilon > 10$, for which the oscillation amplitude is maximized (Fig. 3a), slightly greater power is required to oscillate longer flaps due to their greater viscous resistance.

A higher power required to oscillate the longer flaps results in a reduced efficiency of such swimmers (Fig. 2c). Indeed, we find that the maximum efficiency corresponds to a swimmer with $\text{AR}_p = 2$, which is also the fastest propelling swimmer (Fig. 2a). Thus, for the parameters considered in our simulations, $\text{AR}_p = 2$ is the optimal flap geometry leading to the maximum propulsion speed and efficiency.

To get an insight into the effect of body compliance on the locomotion of the swimmer, we examine the influence of the body stiffness on the dynamics of the rigid propulsive flaps. We use two parameters to characterize the flap dynamics: the change in the size of the swimmer body $w$, which represents the translational flap motion, and the flap deflection angle $\theta$, which represents the magnitude of the flap rotation. In Fig. 3a, we show the variation of maximum body size change $\Delta w$ as a function of body elasticity $\varepsilon$. In Fig. 3b, we present corresponding data for the difference between the maximum and minimum deflection angles of propulsive flaps $\Delta \theta = \theta_{\text{max}} - \theta_{\text{min}}$. The deflection angle $\theta$ is positive when the flaps move outwards and negative when they move inwards.

We find that $\Delta w$ practically does not depend on the flap length and only changes with $\varepsilon$. For softer gels with $\varepsilon = T/\tau < 1$, for which the gel relaxation time $\tau$ is larger than the oscillation period $T$, the change in the body size is small and, as a result, soft swimmers do not propel in the fluid. In this situation, the soft gel network does not have sufficient time to fully expand/contract and to equilibrate the periodically imposed internal stresses that are balanced by a viscous drag of the fluid penetrating the network. When gel stiffness is about unity, $\Delta w$ rapidly increases with increasing $\varepsilon$. In this case, the network relaxation time is comparable with the oscillation period. For $\varepsilon > 10$ (i.e. $\tau < T$), the maximum change in the gel size remains nearly constant. In this case, the gel is close to the equilibrium during the entire volume transition and the amplitude of the body change is about $2aw_0$. We note that for stiff gels, $\Delta w$ slightly exceeds unity. This occurs due to non-affine deformations of the swimmer body, which are caused by the attachment of rigid flaps that locally restrict the gel expansion and contraction.

The magnitude of flaps deflection $\Delta \theta$ also depends on the gel stiffness $\varepsilon$ (Fig. 3b) and, therefore, can affect the swimmer propulsion. This deflection is caused by a combination of the drag-induced moment experienced by the flaps moving in a viscous fluid, and a restoring moment imposed by the attached elastic swimmer body. The moment created as a result of fluid drag is proportional to the flap size and the amplitude of body oscillation. We find that $\Delta \theta$ is maximized for $\varepsilon \sim O(1)$ and is small for relatively soft and stiff swimmers. When the gel body is soft and the restoring elastic moment is weak, the deflection angle is mostly defined by the magnitude of $\Delta w$, which is also small in this case (Fig. 3a). On the other hand, for $\varepsilon > 10$, the deflection angle decreases due to the increased body stiffness and, consequently, a stronger restoring moment opposing flap rotation. This explains why the propulsion velocity decreases for stiff swimmers, although the oscillation amplitude in this situation is maximized (Fig. 3a). We note that for $\varepsilon > 100$, the low gel compliance significantly suppresses the flap rotation and $\Delta \theta$ asymptotically approaches zero.
Furthermore, we find that the phase difference between the body size change $w$ and the deflection angle $\theta$ is not constant and varies with the gel stiffness (Fig. 4). For soft gels (Fig. 4a), there are a nearly 90° phase difference between $\Delta w$ and the imposed internal force $f_{im}$ (shown in Fig. 4 by the dashed line), and a 180° phase difference between $w$ and $\theta$, indicating that the motion is time reversible. As the body stiffness increases, the phase shift between $f_{im}$ and $w$ decreases (see Fig. 4b for $\varepsilon = 3$) and reaches zero for stiff swimmers (see Fig. 4c for $\varepsilon = 30$). Similarly, the phase shift between $w$ and $\theta$ decreases as the gel stiffness is increased. The phase variation between $w$ and $\theta$ is critical for the swimmer propulsion. When this phase difference is different from 180° (or 0°), the motion of the flaps is time-irreversible giving rise to a net propulsive force.

2. Swimmer steering

In order to control the swimmer trajectory, we use a steering flap attached to the front part of our gel swimmer (Fig. 1). Unlike the rigid propulsive flaps, this steering flap is flexible. It is also responsive and it bends in the $x$–$y$ plane due to the action of an external stimulus. The purpose of the responsive steering flap is to introduce controlled left–right asymmetry into the otherwise symmetrical swimmer. When the swimmer self-propels, the flap asymmetry creates a net rotational moment that steers the swimmer in the desired direction.

To characterize the steering flap deformation created by the external stimulus, we define the dimensionless flap curvature $\kappa = w_0 M/EI$, where $M$ is the magnitude of the internal bending moment induced in the flap material by the stimulus. Additionally, we introduce the dimensionless bending stiffness $\Lambda = E/II(L^3/\mu_0)$, which represents the relative strength of steering flap bending rigidity and viscous forces due to periodic oscillations of the flap induced by the pulsating gel body.

To probe the effect of the steering flap deflection on the swimmer trajectory, we conduct simulations in which we temporarily introduce the stimulus and monitor the resulting change in swimmer trajectory. We set the swimmer parameters to $AR_p = 2$ and $\varepsilon = 3$ that yield the most efficient swimmer propulsion. We first let the swimmer, which is initially at rest, swim with a straight steering flap for 10 periods of body oscillations. This allows the swimmer to reach its steady velocity. We then introduce the stimulus by applying a constant internal moment $M$ to the steering flap. The stimulus acts for 50 periods of body oscillations during which the flap is deflected and the swimmer moves along a curved trajectory. After that, we remove the stimulus and the associated internal moment and allow the swimmer to move additional 30 periods during which the flap relaxes to its initial straight shape. To characterize the swimmer turning due to the stimulus, we measure the turning angle $\beta$ that is defined as a difference between the swimmer spatial orientation at the initial and final sections of the swimmer trajectory (see inset in Fig. 5b). The turning angle is positive when the swimmer turns in the direction of steering flap bending.

Fig. 5 shows the dependence of the turning angle $\beta$ on the dimensionless steering flap curvature $\kappa$ for flaps with different aspect ratio $AR_d$ and bending stiffness $\Lambda$. The figure shows that indeed the swimmer changes the direction of motion when the stimulus is introduced and an internal bending moment is applied to the flexible steering flap. Surprisingly, we find that for larger $AR_d$, the turning angle $\beta$ can be either positive or negative depending on the steering flap curvature. In other words, the swimmer can turn not only in the direction of the flap bending, but also in the opposite direction when the flap curvature is sufficiently large.

We can rationalize this counterintuitive swimmer behavior by considering two physical mechanisms that cause swimmer turning: (1) geometrical asymmetry created as a result of steering flap bending (Fig. 6a) and (2) beating motion of the curved elastic flap (Fig. 6b) that generates a sidewise hydrodynamic force misaligned with the direction of swimmer motion. These two simultaneously acting mechanisms may have opposing effects on the swimmer rotation.

To estimate the relative importance of these mechanisms, we conducted simulations in which we isolated the effect of flap asymmetry on swimmer turning. To this end, we simulated the motion of a non-oscillating swimmer. In this case, the swimmer body did not oscillate to create propulsion, but rather the swimmer was dragged through the fluid by a constant external
force uniformly applied to the swimmer body. The magnitude of this force was set to match the swimmer mean propulsion velocity due to gel oscillations. The turning angle $\beta$ was measured following the same routine as for swimmers with an oscillating body.

The results of the simulations of the non-oscillating swimmer are shown in Fig. 5 by the dotted lines. These results indicate that the turning due to the flap asymmetry is always in the clockwise direction (positive $\beta$). The turning angle is maximized when $k = 1.5$ and is greater for longer flaps that create a larger asymmetry of the swimmer, thereby producing a stronger hydrodynamic drag that turns the swimmer in the direction of flap deflection.

By comparing the results for oscillating and non-oscillating swimmers, we find that they significantly deviate from each other. The difference is the most pronounced for the swimmer with a short steering flap (Fig. 5a), which exhibits the fastest turning rate but for which the effect of the flap asymmetry is practically nil. We attribute the difference in turning rate between oscillating and non-oscillating swimmers to the beating motion of the flexible steering flap. This beating of the curved elastic flap arises due to the periodical contractions and expansions of the oscillating gel body.

Fig. 6b shows flap deformation during one period of body oscillations for flaps with curvature $k = 1$ and different bending stiffness $\Lambda$. For clarity, the profiles are shown relative to the averaged position of the curved flap and are normalized by the flap lengths. The steering flap aspect ratios are $AR_s = 0.5, 1,$ and $1.5$ for swimmers in the left, middle, and right columns, respectively. Other swimmer parameters are $k = 1$, $AR_p = 2$, and $E = 3$.

Fig. 5 Turning angle $\beta$ versus dimensionless curvature $k$ for steering flaps with different dimensionless bending stiffness $\Lambda$. Steering flap aspect ratios in panels (a), (b), and (c) are $AR_s = 0.5, 1,$ and $1.5$, respectively. Other swimmer parameters are $AR_p = 2$ and $E = 3$. The dotted lines show the results for a non-oscillating swimmer propelled with a constant velocity equal to the period-averaged velocity of the corresponding oscillating swimmer.
due to the flap beating. For longer steering flaps, the effects of the drag asymmetry and the sidewise steering propulsion are of comparable magnitude. In this case, not only the turning rate, but also the turning direction can be changed by simply varying the magnitude of the internal moment of the responsive steering flap. Since this moment can be regulated by changing the strength of the external stimulus, this behavior may be utilized for directing the motion of artificial self-propelling swimmers in microfluidic channels.

### 3. Experimental parameters

In our computational study, we use dimensionless parameters to describe the properties and dynamics of our synthetic swimmer. These parameters can be readily translated into relevant experimental values. For example, the swimmer with \( \epsilon = 3 \) can be built using a 50 \( \mu \)m cubic hydrogel with a collective diffusion coefficient of \( D_0 \sim 10^{-5} \text{ cm}^2 \text{s}^{-1} \) that oscillates with a frequency 1 Hz and an amplitude \( \Delta v \sim 25 \mu \text{m} \). In this case, our simulations predict that the most efficient swimmer will propel with a speed of about 1 \( \mu \text{m} \text{s}^{-1} \).

In fact, different combinations of elastic modulus, size, and oscillation frequency can be selected to design swimmers with similar performance, which is defined by the dimensionless elasticity \( \epsilon \) that relates the physical properties. Furthermore, non-cubic and/or anisotropic oscillating gels can be used to drive the motion of the swimmer. In this case, however, the body oscillations should result in symmetric beating of propulsive flaps that is required for the straight swimming. The performance of the non-cubic/anisotropic swimmer will depend on the degree of material anisotropy and/or relative magnitude of volume oscillations and may be different from that of swimmers with a cubic body. Since the velocity of the swimmer is proportional to the rate of the body volume change, materials with larger swelling/contraction ratios should provide faster swimmer motion.

Cyclic changes in the volume of the responsive gel body can be induced by a range of stimuli. For instance, Yoshida et al. have synthesized gels capable of undergoing autonomous periodic changes in volume. The self-sustained motion in these gels is driven by the oscillating Belousov–Zhabotinsky (BZ) reaction taking place within the gel polymer network. In this system, the BZ catalyst (ruthenium metal-ion complex) is grafted to the polymer network and the catalyst periodic oxidation and reduction induce cyclic changes in the gel volume. Experiments have shown that BZ gels can maintain autonomous volume oscillations of up to 20% for several hours. The period of the oscillations can be of the order of seconds. Although volume changes of BZ gels are not sinusoidal in time, as those considered in our simulations, they can be still used to drive time-irreversible motion of the flaps and thereby induce swimmer propulsion.

Furthermore, researchers have synthesized thermoresponsive gels that swell and deswell as a result of temperature changes in the surroundings. The temperature-sensitive gels often contain nanoparticles embedded in their polymer network. Upon the application of an external magnetic or optical field, the nanoparticles heat up and trigger a rapid volume change of the responsive material. Thus, an alternating magnetic field or laser light could be used to create continuous volume oscillations in the gels with embedded nanoparticles. In this case, the oscillation frequency is defined by the intensity of the excitation and the thermal properties of the environment that controls the heat dissipation.

Lastly, thin polymeric films that bend in response to external stimuli can be potentially employed as steering flaps. A light-sensitive liquid-crystal polymer plate is an example of such materials. These thin plates deform upon exposure to a light source and can yield a fast response time and large deformation. It was also demonstrated that the amount of bending can be tuned by controlling the light intensity. This property could be harnessed for precise control of the swimmer trajectory.

### Summary

Using computational modeling, we designed a simple maneuverable micro-swimmer that can self-propel and navigate in microfluidic channels. The swimmer is actuated by periodic swelling and deswelling of its responsive gel body that drives the motion of rigid propulsive flaps attached to the body sides. We showed that the swimmer propulsion in the inertialless, low-Reynolds-number environment is due to non-reciprocal oscillations of the rigid flaps that undergo translational and rotational motions with a phase lag defined by the elasticity of the responsive gel body. Thus, swimmer propulsion can be regulated by tuning the physical properties of the gel. We found that the propulsion is the most efficient, when the flap aspect ratio is about two.

To control the swimmer trajectory, we introduced a steering flap that is attached to the front end of the swimmer body. We assumed that the flexible steering flap is responsive and bends when exposed to an appropriate external stimulus. In this situation, we found two physical mechanisms that redirect swimmer motion: an asymmetric drag force due to the flap curvature and sidewise propulsion due to the periodic flap oscillations. The second mechanism is the most pronounced for short flaps, whereas for longer flaps these effects may be of comparable magnitude. In the latter case, the intensity of an external stimulus can change not only how fast the swimmer turns, but also in what direction it turns, which may greatly simplify control of swimmers in microfluidic channels.

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### References


Supplementary Information

**Self-propelling motion of the swimmer**

The animation demonstrates self-propelling motion of the gel micro-swimmer in a highly-viscous low-Reynolds-number fluid. The swimmer motion is due to non-reciprocal beating of rigid propulsive flaps driven by oscillating swimmer body that periodically expands and contracts. The swimmer first moves along a straight line during ten oscillation periods. After that the steering flap, attached to the swimmer front end, bends due to an external stimulus (indicated by a light bulb) and causes swimmer turning. The swimmer turns due to the asymmetry of the curved steering flap and its periodical beating. The swimmer parameters are $\delta = 3$, $AR_\mu = 2$, $AR_\gamma = 0.5$, $\Lambda = 2.12$, and $\kappa = 1$. 
**Fluid flow induced by the swimmer motion**

The figure shows fluid velocities in the $x - y$ and $x - z$ planes through the middle of the swimmer body during one period of the gel body oscillations. Blue arrows represent velocity vectors and black contours represent swimmer body and flaps. The size of the reference arrow in the figure top corresponds to the magnitude of the characteristic velocity $V_c$. The swimmer parameters are $AR_p = 2$, $AR_i = 1$, and $\delta = 3$.