Biologically-Inspired Microrobotics

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Abstract

The locomotion of microorganisms in low Reynolds number regimes has inspired engineers to design and fabricate robotic systems at micro- and nano-scales. Here, we review the swimming at low Reynolds number and analyze the kinematic reversibility property and the scallop theorem theoretically and experimentally. First, we present dynamical models for the planar flagellar and helical flagellar swimming of microorganisms. Second, we study the realization of microrobotic systems based on the mentioned locomotion mechanisms at micro- and nano-scales using fabrication based on nano-technology. Finally, the locomotion mechanisms of the biologically-inspired microrobotic systems are experimentally investigated using electromagnetic and magnetic systems.

Keywords: Microrobots, microorganisms, locomotion, low Reynolds number

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1. The Journey to Biologically-Inspired Microrobots

In seeking nutrient efficiently, microorganisms undergo different locomotion mechanisms. For example, peritrichously flagellated Escherichia coli swim by wrapping their flagella together in a helical bundle. The continuous rotation of this bundle enables locomotion and swimming back-and-forth. The Escherichia coli tumble whenever its necessary to change the swimming direction. Other
Figure 1: A timeline of the development of biologically-inspired microrobots. The first microrobot is developed by Drefus et al. in 2004 [1]. This microrobot breaks time-reversal symmetry using a propagating wave along its lumped structure. In 2007, Bell et al. have fabricated and controlled the smallest artificial helical bacterial flagella using rotating magnetic fields [2], and in 2009 Ghosh et al. have decreased the size of the artificial bacterial flagella by 30 times in length [3]. Behkam et al. [4] and Magdanz et al. [5] have presented the first hybrid micro-bio-robots through coupling with *Serratia marcescens* bacteria (2007) and sperm cells (2013), respectively. In 2014, Khalil et al. [6] and Williams et al. [7] have developed microrobots based on the propagation of planar waves through a flexible tail.

Monotrichous bacteria cannot tumble with single flagellum. Therefore, they depend on rotational Brownian motion to change direction [8]. In his widely read lecture entitled *Life at Low Reynolds Number* [9], Purcell explained the swimming strategies of microorganisms that are based on helical rotation without symmetry and more than one degree of freedom. Two opposite examples are again the *Escherichia coli* and the scallop, the first rotate their flagellar bundle counterclockwise and the cells rotate clockwise. This mechanism is based on more than one degree of freedom, whereas the scallop opens and closes its shell periodically and possesses one degree of freedom. The propulsion with
helical rotation is not the only mechanism for locomotion. During their journey towards the ovum, sperm cells undergo a wide variety of swimming patterns by a beating tail [10]. The sperm cell propagates planar or three-dimensional travelling wave (that breaks time-reversal symmetry) along the tail. This microorganism consists of a head and a flagellum that contains a mid-piece and an actively beating tail. The travelling waves are generated by local bending moment along the flagellum. This motion does not impart momentum to the fluid due to the absence of inertia, and hence, in the pursuit of locomotion at micro- and nano-scales, researches have mimicked and adapted swimming strategies of microorganisms.

Dreyfus et al. have mimicked the locomotion mechanism of the sperm cells by colloidal magnetic particles that are connected together using DNA and attached to a red blood cell [1]. The external magnetic fields have allowed this biologically-inspired microrobot to be adjusted and driven using a flagellated swim (Fig. 1). Bell et al. have also fabricated artificial bacterial flagella and demonstrated the first locomotion mechanism based on the helical propulsion of the Escherichia coli [2]. The artificial bacterial flagella is propelled using external rotating magnetic fields. In the same year (Fig. 1), Behkam et al. have
Table 1: Summary of biologically-inspired microrobots based on the microorganisms.

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demonstrated a novel hybrid micro-bio-robot [4]. This microrobot is fabricated using polystyrene beads and driven by motile Serratia marcescens bacteria. The bacteria provide propulsive force to the bead that can be steered using external magnetic field. The coupling between the bead and the motile microorganism provides broad possibilities in biomedical application and nano-technology. The microorganisms provide propulsive force to the bead without an onboard power source. In 2006, Martel et al. have suggested that magnetotactic bacteria can be used as microrobots without coupling with another micro- or nano-object [17, 20]. Magnetotactic bacteria develop magnetite nano-crystals inside their cells, and hence their magnetic dipole moment enables directional control using external magnetic field. In addition, a swarm of magnetotactic bacteria have been used to achieve non-trivial tasks such as manipulation of spherical
beads and microassembly of non-magnetic microobjects [21, 22]. We realize that in the pursuit of locomotion at low Reynolds number regime in micro- and nano-scales, scientists have either mimicked the designs of nature or used the micro-organism itself to provide propulsion, as shown in Fig. 2. Behkam et al. [4] and Julius et al. [16] have used the same microorganisms, i.e., *Serratia marcescens*, to actuate beads and microbarge, respectively. In this case, bio-adhesion has to be achieved between the motile cells and the surface of the bead or the microbarge. The bio-adhesion can be enhanced by the chemical treatment of these object to increase the attraction with the microorganisms. Another approach has been demonstrated by Magdanz et al. by achieving mechanical coupling between micro-tubes and motile sperm cells (Fig. 2). The disadvantage of this approach is its low coupling efficiency due to the dependence on random coupling between sperm cells and the magnetic micro-tubes [12, 23]. This coupling is necessary as the elements of the hybrid micro-bio-robots are not useful alone, unlike magnetotactic bacteria that possess all elements required for locomotion and steering, i.e., flagellar propulsion and magnetic dipole moment.

Table 1 provides the contributions of some research groups in the field of biologically-inspired microrobots. We primarily focus only on those designs that are or can be implemented on micro- and nano-scales. The work in this field has started in the last decade. Although other work has been proposed earlier but not on micro- or nano-scale. For example, Honda et al. were the first group to demonstrate helical propulsion on macro-scale in 1996 [24]. There design achieves locomotion by imparting momentum to the fluid. In this chapter, we study the microrobotic systems that cannot impart momentum to the fluid but rather utilize and adopt different strategies to navigate at low Reynolds number. The remainder of this chapter is organized as follows: Section 2 provides descriptions pertaining to locomotion at low Reynolds number through breaking time-reversal symmetry using planar and helical propulsion. A facile fabrication technique (based on electrospinning) to provide robotic sperms is presented in Section 3. An overview of the motion control of biologically-inspired microrobotic system using magnetic-systems with closed- and open-configuration is
Figure 3: Helical flagellated swim is achieved by magnetotactic bacteria (*Magnetospirillum Magnetotacticum* Strain MS-1) and helical microrobot inside a microfluidic chip with maze structure [25] and a catheter segment, respectively. Left: The magnetotactic bacterium is controlled by directing the magnetic fields and swim using its flagella. Right: The helical microrobot is propelled using rotating magnetic fields.

included in Section 4. Finally, Section 5 concludes and provides directions for future studies.

2. Locomotion at Low Reynolds Number

Dynamics of all fluids is governed by Navier-Stokes equations that consists of the following Newtons second law and the continuity equation [26]:

\[
\frac{\partial \mathbf{u}}{\partial t} + \mathbf{u} \cdot \nabla \mathbf{u} = -\frac{1}{\rho} \nabla p + \frac{\mu}{\rho} \nabla^2 \mathbf{u} + \mathbf{b} \quad \text{and} \quad \nabla \cdot \mathbf{u} = 0, \tag{1}
\]

where \( \mathbf{u} \) is the velocity vector of the fluid and \( p \) is the pressure. Further, \( \rho \) and \( \mu \) are the density and the dynamic viscosity of the fluid, respectively, and \( \mathbf{b} \) is a body force. Simplifications can be made to (1) using the concept of the Reynolds number which has been suggested by Gabriel Stokes (1851) and Osborne Reynolds (1883). The Reynolds number is the ratio of the inertial force and viscous force, and is approximated by

\[
Re \equiv \frac{\text{Inertial force}}{\text{Viscous force}} = \frac{\rho UL}{\mu}, \tag{2}
\]

where \( L \) and \( U \) are length and velocity that depend on the flow. Figure 3 shows a magnetotactic bacterium (MTB) inside a microfluidic channel with a maze structure [25]. The length \( L \) of the Bacterium is 5 \( \mu m \) and its speed \( U \) is...
35 $\mu$m/s. The microfluidic channel is filled with growth medium with density of approximately 1000 kg/m$^3$ and dynamic viscosity ($\mu$) of $10^{-3}$ Pa.s. Therefore, Reynolds number is on the order of $1.75 \times 10^{-4}$. Reynolds number has another interpretation, let us assume that we control the flow of the growth medium of the MTB inside the microfluidic channel. In this case, $L$ is the width of the channel and $U$ is the flow of the growth medium, and hence Reynolds number has different interpretations that depend on the flow situation. Another example is the helical microrobots in Fig. 3, it swims using helical propulsion under the influence of external rotating magnetic fields inside a catheter segment with controlled flow. The Reynolds number of this microrobot is on the order of 1.3. Inducing a flow in the catheter segment would enable another interpretation of Reynolds number, where the flow and inner-diameter of the segment are represented using $U$ and $L$, respectively. In contrast to the propulsion of the helical microrobot that relays on imparting momentum to the fluid ($R_e = 1.3$), an MTB depends solely on viscous damping ($R_e = 1.75 \times 10^{-4}$). Therefore, we can ignore the inertial terms in (1) and make some simplifications to obtain the following linear Stokes equation:

$$\nabla p = \mu \nabla^2 \mathbf{u} \quad \text{and} \quad \nabla \cdot \mathbf{u} = 0.$$  \hspace{1cm} (3)

We observe the following properties from Stokes equation:

1. Stokes equations are linear, unlike Navier-Stokes equations (1);
2. In contrast to Navier-Stokes equations, Stokes equations are time-reversible;
3. The instantaneous structure of the flow does not depend on the history of motion, but rather on the present configuration.

The second property is illustrated experimentally in Fig. 4. three paramagnetic microparticles are attached together and oscillate under the influence of weak oscillating magnetic field ($\mathbf{B}$). The magnetic fields are uniform and does not exert magnetic force on the dipole moment ($\mathbf{m}$) of the microparticles. This reciprocal (flapping) motion cannot generate any propulsion due to absence of inertia. The experiment provided in Fig. 4 is nothing but the scallop theorem.
of Purcell which states that \textit{any reciprocal motion cannot generate net propulsion (or fluid transport)} \cite{9}. Therefore, a microorganism with a rigid flapping motion cannot generate any propulsive force for locomotion. Fig. 4 proves that the rigid flapper is stationary, and hence locomotion at low Reynolds number regime necessitates time-reversal asymmetry. Microorganisms break the time-reversal symmetry by generating planar flagellar waves or propagating helical flagellar waves.

2.1. Planar Flagellar Propulsion

Taylor \cite{27} approximated the planar flagellar propulsion using a two-dimensional infinite wave $\Psi(x, y, t)$ in a velocity fields, $\mathbf{v} = u\mathbf{x} + v\mathbf{y}$. A propulsion speed ($U$) develops for a travelling wave along the opposite direction. The elements of the velocity field are given by

$$
  u = \frac{\partial \Psi}{\partial y}(x, y, t) \quad \text{and} \quad v = -\frac{\partial \Psi}{\partial x}(x, y, t).
$$

Figure 4: A cluster of 3 paramagnetic microparticles undergoes a rigid flapping under the influence of oscillating magnetic fields. This rigid reciprocal motion does not provide propulsion at low Reynolds number. Therefore, the cluster is almost stationary in all time instants. The white and blue (solid and dashed) arrows represent the magnetic dipole moment of the cluster and the magnetic fields, respectively. The white lines indicate the initial position of the cluster.
Table 2: Boundary conditions of the biharmonic equation.

<table>
<thead>
<tr>
<th>1st order</th>
<th>BCs</th>
<th>2nd order</th>
<th>BCs</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \frac{\partial \Psi_1}{\partial y}</td>
<td>_{x,y\to \infty} )</td>
<td>( U_1 )</td>
<td>( \frac{\partial \Psi_2}{\partial y}</td>
</tr>
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<td>( \frac{\partial \Psi_1}{\partial x}</td>
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<td>0</td>
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<tr>
<td>( \frac{\partial \Psi_1}{\partial x}</td>
<td>_{x,y=0} )</td>
<td>( \cos(x-t) )</td>
<td>( \frac{\partial \Psi_2}{\partial x}</td>
</tr>
</tbody>
</table>

Pak et al. [26] have proven that the stream function formulation of the Stokes equations for planar flagellar propulsion is given by

\[
\nabla^4 \Psi(x, y, t) = 0. \tag{5}
\]

It has been also shown that (when the wave amplitude is much smaller than the wavelength) solution of (5) with the boundary condition in Table II is approximated by

\[
\psi = \varepsilon \psi_1 + \varepsilon^2 \psi_2 + \cdots \quad \text{and} \quad U = \varepsilon U_1 + \varepsilon^2 U_2 + \cdots, \tag{6}
\]

where \( \varepsilon = ak \) is the dimensionless wave amplitude. Further, \( a \) and \( k \) are the amplitude and wave number, respectively. The first- and second-order solutions (\( \psi_1 \) and \( \psi_2 \)) are given by

\[
\psi_1 = (1 + y) \exp^{-y} \sin(x - t) \quad \text{and} \quad \psi_2 = \frac{y}{2} - \frac{y \exp^{-2y}}{2} \cos 2(x - t). \tag{7}
\]

Therefore, the swimming speed \( U \) is proportional to

\[
U \sim a^2 k^2 c. \tag{8}
\]

Fig. 5 shows a flexible flapping motion of a sperm-shaped microrobot [28] at frequency of 1 Hz [29]. The flexibility of the tail enables wave propagation and breaking of the time-reversal symmetry. We construct a numerical model to analyze the motion of this biologically-inspired microrobot. The numerical model of the magnetic microrobot consists of three components: elastohydrodynamics for dynamic tail deformation based on Timoshenko-Rayleigh beam
Figure 5: Flexibility of the tail of the sperm-shaped microrobot is demonstrated under the influence of oscillating magnetic field at 1 Hz. This flexibility enables a non-reciprocal deformation and flagellated swim. The radius of curvature is provided at each time instant and the tail is almost straight at time, t= 1.0 seconds. The tail deformations are overlayed and schematically represented in the bottom-left corner of the first time instant.

theory [30, 31], Magnetohydrodynamics based on the Biot-Savart law for electromagnetic coils [32, 33], and rigid-body kinematics based on transient Stokes-flow approach with force-free swimming conditions [27, 34]. Each component utilizes resistive force coefficients based on resistive-force-theory [8, 35] to calculate the resultant hydrodynamic forces acting on the elastic tail and magnetic body of the swimmer. The equation of motion of the magnetic microrobot based on the force-free swimming condition [8] is given by

\[
\begin{bmatrix}
V \\
\Omega
\end{bmatrix} = -B_{sw}^{-1} \begin{bmatrix}
F_{mag} + F_{add} \\
T_{mag} + T_{add}
\end{bmatrix},
\]

where \( V \) and \( \Omega \) are the linear and angular rigid-body velocities of the microrobot, respectively. Further, \( F_{mag} \) and \( F_{add} \) are the magnetic force and inertial force due to added mass on the microrobot, respectively. Further, \( T_{mag} \) and \( T_{add} \) are the magnetic torque and the inertial torque due to added mass acting on the microrobot, respectively. In addition, \( B_{sw} \) is the resistance matrix of the
microrobot that consists of the resistance matrices of the body \((B_b)\) and the tail \((B_t)\), and is given by

\[
B_{sw} = B_b + B_t. \tag{10}
\]

In (10), the resistance matrix of the body is represented using

\[
B_b = \begin{bmatrix}
D_{\text{tran}} & -D_{\text{tran}}S_b \\
S_bD_{\text{tran}} & D_{\text{rot}}
\end{bmatrix}, \tag{11}
\]

where \(D_{\text{tran}}\) and \(D_{\text{rot}}\) are the diagonal matrices of translational and rotational resistive force coefficients of the body, and \(S_b\) is a skew-symmetric matrix signifying the cross-products. In (10), \(B_t\) is given by

\[
B_t = \int_0^{l_t} \begin{bmatrix}
R_{\text{CR}}^T & -R_{\text{CR}}^T S_t \\
S_t R_{\text{CR}}^T & -S_t R_{\text{CR}}^T S_t
\end{bmatrix} d\ell \tag{12}
\]

where \(l_t\) is the length of the sperm-shaped microrobot. Further, \(C\) and \(R\) are the diagonal matrix of the local resistive force coefficients, and the rotation matrix from local Frenet-Serret coordinate frames to the inertial frame of the reference of the microrobot along the elastic tail, respectively. Furthermore, \(S_t\) is another skew-symmetric matrix signifying the cross-products. The magnetic force \((F_{\text{mag}})\) and torque \((T_{\text{mag}})\) acting on the magnetic body are given by

\[
\begin{bmatrix}
F_{\text{mag}} \\
T_{\text{mag}}
\end{bmatrix} = \begin{bmatrix}
R_{sw}^T V (M \cdot \nabla) B \\
VM \times B
\end{bmatrix}, \tag{13}
\]

where \(V\) is the volume of the magnetic material and \(M\) is its magnetization in the frame of the laboratory. Further, \(B\) is the magnetic flux density vector. The frame of the electromagnetic system and the frame of the microrobot are related using the rotation matrix \((R_{sw})\). The components of the electromagnetic fields generated using our electromagnetic configuration are calculated using [36]

\[
B_x = \left(\frac{5}{4}\right)^{\frac{3}{2}} \frac{B_0}{4\pi} \int_0^{2\pi} \left(1 - \frac{y}{r} \cos \theta - \frac{z}{r} \sin \theta\right) F_0^2 d\theta, \tag{14}
\]

where \(B_x\) is the magnetic field along \(x\)-axis and \(r_i\) is the inner-radius of the
electromagnetic coil. $B_0$ is given by

$$B_0 = \left(\frac{5}{4}\right)^2 \frac{\mu_0 N I_c}{r}. \quad (15)$$

In (15), $\mu_0$, $N$, and $I_c$ are the permeability of the iron core, number of turns of each coil, and the input current on each of the electromagnetic coils, respectively. Further, $F_0$ is given by

$$F_0 = \left(\frac{x}{r}\right)^2 + \left(\frac{y}{r} - \cos \theta\right)^2 + \left(\frac{z}{r} - \sin \theta\right)^2, \quad (16)$$

The magnetic field along $y$-axis is given by

$$B_y = \left(\frac{5}{4}\right)^2 \frac{B_0}{4\pi} \int_0^{2\pi} \left(\frac{x \cos \theta}{r} \frac{F_0^2}{F_0^2}\right) d\theta, \quad (17)$$

furthermore, the magnetic field along $z$-axis is given by

$$B_z = \left(\frac{5}{4}\right)^2 \frac{B_0}{4\pi} \int_0^{2\pi} \left(\frac{x \sin \theta}{r} \frac{F_0^2}{F_0^2}\right) d\theta. \quad (18)$$

The magnetic field gradient is also be calculated using (15), (17), and (18) to determine the magnetic force (13) exerted on the magnetic dipole of the micro-robot. The structural deformation in the frame of the microrobot is calculated based on the Timoshenko-Rayleigh beam model [30, 31] in order to take the vibration effect and large deformations into account for high actuation frequencies and axial shear forces. Finally, the transient hydrodynamic force due to the added-mass is calculated using [18]

$$\begin{bmatrix} F_{add} \\ T_{add} \end{bmatrix} = \begin{bmatrix} \Phi & \int_{-\infty}^{t} \frac{dV}{dr} \frac{dr}{\sqrt{A-\pi}} - \frac{\pi}{3} R^3 \rho_l \frac{dV}{dt} \\ S \Phi \end{bmatrix} \begin{bmatrix} F_{add} \\ F_{add} \end{bmatrix}, \quad (19)$$

where $\rho_l$ is the density of the liquid medium and $R$ is the radius of the body. In (13), $\Phi$ is calculated as $\Phi = 6 R^2 \sqrt{\pi \mu \rho_l}$, where $\mu$ is the dynamic viscosity of the liquid environment. The driving currents on the coils are modeled as:

$$i_A = i_C = I_{\text{max}} \sin \left(\Omega_z + \frac{\pi}{4} \cos(2\pi ft)\right), \quad (20)$$

where $i_A$ and $i_C$ are the current inputs to electromagnetic coils A and C (Fig. 6), respectively. Further, $I_{\text{max}}$ and $f$ are the maximum input current and the
Figure 6: Experimental and simulation frequency response of sperm-shaped microrobots. The average speed is calculated from 10 trials for each frequency [29]. The maximum swimming speed is observed at frequency of 10 Hz. The simulation results (red line) are calculated using 9. The diameter of the head is 40 µm and the length of the tail is 100 µm. Oscillating magnetic fields are generated by applying the current inputs (20) and (21) to the 4 electromagnetic coils.

frequency, respectively. The current inputs to electromagnetic coils B \( (i_B) \) and D \( (i_D) \) are given by

\[
i_B = i_D = I_{\text{max}} \cos \left( \frac{\Omega_z + \pi}{4} \cos(2\pi ft) \right),
\]

(21)

Our model is verified by comparing its results to the experimental frequency response of the microrobot, as shown in Fig. 6. The swimming speed of the sperm-shaped microrobot increases almost linearly with the frequency of the oscillating fields. This frequency response experiment is achieved using an electromagnetic system with orthogonal configuration and currents (20) and (21) are supplied to the electromagnetic coils.
2.2. Helical Flagellar Propulsion

A helical microrobot is subjected to the following magnetic force ($\mathbf{F}$) and magnetic torque ($\mathbf{T}$) under the influence of external magnetic fields:

$$\begin{bmatrix} \mathbf{F} \\ \mathbf{T} \end{bmatrix} = V \begin{bmatrix} (\mathbf{M} \cdot \nabla) \mathbf{B} \\ (\mathbf{M} \times \mathbf{B}) \end{bmatrix} = V \begin{bmatrix} (\mathbf{M} \cdot \nabla) \mathbf{B} \\ \hat{\mathbf{M}} \mathbf{B} \end{bmatrix}, \tag{22}$$

where $V$ is the volume of the magnetic material of the helical microrobot. In (22), $\hat{\mathbf{M}}$ is the skew-symmetric form of the magnetization vector $\mathbf{M}$, where $\hat{\mathbf{M}} = \text{SK}(\mathbf{M})$, and $\text{SK}(\cdot)$ is the skew-symmetric operator [37]. The helical microrobot navigates in a viscous flow and experiences force ($\mathbf{f}$) and torque ($\mathbf{t}$) that are approximated by

$$\begin{bmatrix} \mathbf{f} \\ \mathbf{t} \end{bmatrix} = - \begin{bmatrix} a & b \\ b & c \end{bmatrix} \begin{bmatrix} \mathbf{v} \\ \omega \end{bmatrix}, \tag{23}$$

where $\mathbf{v}$ and $\omega$ are the linear and angular velocity vectors of the helical microrobot, respectively. Further $a$, $b$, and $c$ are the coefficient of the propulsion matrix and are given by

$$a = 2\pi n R_h \left( \frac{\xi_\parallel \sin^2 \theta + \xi_\perp \cos^2 \theta}{\cos \theta} \right), \tag{24}$$

where $n$ and $R_h$ are the number of turns of the helical flagella and the radius of the helix. Further, $\theta$ is the helix angle, and the drag coefficients ($\xi_\parallel$) and ($\xi_\perp$) are given by

$$\xi_\parallel = \frac{4\pi \eta a}{\ln \left( \frac{2a}{b} \right) - \frac{1}{2}} \quad \text{and} \quad \xi_\perp = \frac{8\pi \eta a}{\ln \left( \frac{2a}{b} \right) + \frac{1}{2}}, \tag{25}$$

where $a$ and $b$ are the major and minor diameter of the elliptical head of the helical microrobot, and $\eta$ is the dynamic viscosity of the medium. In (25), $b$ is given by

$$b = 2\pi n R_h^2 \left( \xi_\parallel - \xi_\perp \right) \sin \theta. \tag{26}$$

Finally, $c$ is calculated using

$$c = 2\pi n R_h^3 \left( \frac{\xi_\parallel \cos^2 \theta + \xi_\perp \sin^2 \theta}{\cos \theta} \right), \tag{27}$$

The motion of the helical microrobot is governed by (22) and (23).
The advancements in nano-technology enable fabrication of biologically-inspired microrobots at the nano- and micro-scales. We focus on the fabrication of sperm-shaped microrobots and readers are referred to the original papers of Bell et al. [2] and Fischer et al. [3] for details on the fabrication of helical microrobots. The sperm-shaped microrobots are fabricated by electrospinning [39, 40] using a solution, i.e., polystyrene and dimethylformamide, that is slowly injected through a needle via a syringe pump. This solution is mixed with iron oxide nano-particles. An electrical potential is applied using a high voltage power supply between the needle and the collector to introduce free charge at the liquid surface (Fig. 7(a)). The free charge generates electric stress that causes the liquid jet to stretch into fine filaments and accelerate away from the needle towards the collector. The liquid meniscus at the needle opening forms a conical shape when the electrical potential is increased to 10 kV. A liquid jet with high charge density is observed at the cone apex where the free charge is...
highly concentrated on the way from the needle to the collector. The liquid jet stretches via bending instability, and hence increases its surface area. Furthermore, the solvent of the polymer solution evaporates until the jet solidifies and the beaded fibers are localized on the collector. Rayleigh instability occurs if the diameter of the jet is relatively small and if the solution is in liquid state. This instability allows the jet to disintegrate into beads. If the jet is very thin and the solvent is evaporated, the Rayleigh instability is suppressed [39].

The transition from formation of beads, beads with fibers, and pure fibers depends on increasing the initial concentration of the polymer [39]. Beaded fibers (Fig. 7(b)) are formed on the collector for solutions with intermediate viscosity. We extract the beaded fibers from the metallic grit collector Θ. Then, the collected structures are cut and extracted using tweezers under high magnification. One cut is made close to the bead and the second is made based on the desired length of the tail ($l_t$). This procedure leads to a geometry that resembles the morphology of a sperm cell, as shown in Fig. 7(e) and Fig. 8. The beads provide the magnetic dipole, whereas the propulsive force is generated
by the ultra-thin fiber using a flagellated swim. Motion of the sperm-shaped microrobot is controlled using an electromagnetic system with orthogonal configuration (Fig. 6).

4. Magnetic Control and Applications

In contrast to microrobotic system that are powered and steered using bubble propulsion [41], electric propulsion [42], ultrasound-propulsion [43], and self-electrophoretic propulsion, majority of biologically-inspired microrobots are propelled and steered using external source of magnetic fields. These fields are generated using electromagnetic and magnetic systems with closed-configurations [44] and open-configurations [45], as shown in Fig. 9. The function of these configurations of electromagnetic coils is twofold: first, to provide magnetic field lines that directs the microrobot (the easy axis of the microrobot [46]) towards a reference position; second, to rotate or oscillate these fields to enable propulsion. Embedded magnetic layer permits the microrobot to align along these fields, and hence rotate or oscillate to provide helical propulsion of planar wave, respectively. Let us analyze the dynamics of a helical microrobot in a controlled magnetic field. A simple motion control strategy is based on directing the magnetic field lines towards a reference position. Therefore, we use the rotational dynamics of the helical microrobot to analyze and design the control input. Using (22) and (23), we obtain the following rotational dynamics of the helical microrobot:

\[ VMB - b\dot{v} - c\omega = 0. \tag{28} \]

The magnetic field (B) is controlled to align the microrobot (let us assume simple rotations) towards a reference position. Therefore, the angular position and velocity errors (e and \( \dot{e} \)) are given by

\[ e = \Phi - \Phi_{\text{ref}} \quad \text{and} \quad \dot{e} = \dot{\Phi} = \omega, \tag{29} \]

where \( \Phi \) and \( \Phi_{\text{ref}} \) are the angular position of the microrobot and the fixed reference orientation that directs the microrobot towards the reference position,
respectively. We rewrite (28) using (29), and devise a proportional control input 
\( (B \mapsto K_p e) \) to obtain the following error dynamics:

\[
\dot{e} - \frac{V}{c} \hat{M}_p e = -\frac{b}{c} v.
\]  

(30)

We select the following storage function \( S(e) \):

\[
S(e) = \frac{1}{2} e^T e.
\]  

(31)

Taking the time-derivative of (31) yields

\[
\dot{S}(e) = \frac{V}{c} e^T K_p \hat{M} e - \frac{b}{c} v^T e \leq -v^T e.
\]  

(32)

Therefore, the system with the input \( (-v) \) and output \( (e) \) is passive with the storage function \( (S(e)) \). This control strategy enables directional control of the microrobot towards reference position. The oscillation or rotation of these fields allows for propulsion along these field lines as shown in Fig. 10 and Fig. 11. The oscillating field enables the sperm-shaped microrobot to orient towards a
Figure 10: A representative closed-loop motion control of the sperm-shaped microrobot. The sperm-shaped microrobot slides on the bottom of the petri-dish towards the reference position (small blue circle) at a sliding speed of 12 µm/s, at an oscillating magnetic field with frequency of 5 Hz. The large blue circle indicates the head of the sperm-shaped microrobot.

reference position (small blue circle) and achieve a flagellated swim under the influence of oscillating field of 5 Hz, as shown in Fig. 10. A helical propulsion is also achieved using the same control strategy in three-dimensional space using a helical microrobot, as shown in Fig. 11. The helical microrobot follows a circle (dashed red circle) along \( xz \)-plane and overcome its weight by providing a propulsive force component along \( z \)-axis. Oscillating magnetic field of 8 Hz enables helical flagellated swim at a speed of 171 µm/s.

The control result shown in this chapter indicate that biologically-inspired microrobots hold promise in medical and diverse nano-technology applications. For instance, the helical microrobots can be coated with a chemotherapeutic agent and controlled towards a diseased region to achieve targeted therapy. Helical microrobots have also the potential to be used in clearing of clogged blood vessels or the arterial plaque [50]. Fig. 12 shows a helical microrobot drilling through a blood clot inside a catheter segment under the influence of rotating magnetic fields. In this experiment, the helical microrobot decreases the size of the blood clot with approximately 50% following 36 minutes of drilling using magnetic field of 20 mT and frequency of 6 Hz [51]. Mahoney et al. [52] have experimentally demonstrated the existence of magnetic torques that can
Figure 11: The helical microrobot is following a circular trajectory in $xz$-plane with a diameter of 3 mm. The force ($mg$) due to gravity is compensated by the helical propulsion of the microrobot ($yz$-plane), where $m$ is the mass of the microrobot and $g$ is the acceleration due to gravity. The speed of the microrobot is calculated to be 171 $\mu$m/s, at frequency of 8 Hz.

simultaneously stabilize and destabilize a helical microrobot in soft-tissue. An empirical model of helical microrobot has been developed from experimental measurements in an agar gel phantom to predict the effect of the tissue material and the role of the microrobot geometry on the resulting trajectory [53]. Despite this progress in fabrication, actuation, and control of helical microrobots, numerous challenges remain for translating these microrobots into in vivo applications such as: first, the flow rates of blood in aorta, arteries, capillaries, veins, and vena cava are higher than the flow rates used in this study and necessitate larger magnetic field gradient to move and hold the micro-particles against the flow of bodily fluids; second, clinical imaging modality must be integrated to the electromagnetic configuration to provide feedback to the control system and its resolution has to be at the micro scale; third, the biocompatibility and physiological conditions of the drug release must be studied and implemented experimentally.

5. Concluding Remarks

In this chapter, we have studied two propulsion mechanisms of microorganisms that are used in the design and implementation of microrobotic systems. We have focused on the modeling, fabrication, and motion control using exter-
Figure 12: Penetration of a blood clot with diameter and width of 3 mm and 5 mm, respectively, is achieved using a helical microrobot [51]. The microrobot swims in the phosphate buffered saline at an average speed of 600 µm/s and contact with the blood clot is observed at time, \( t = 5 \) seconds. At time, \( t = 436 \) seconds, the blood clot becomes mobile and both rotation and translation along the catheter segment are observed. In this representative experiment, the size of the blood clot is decreased by 50% following 36 minutes of drilling using the microrobot.

Biologically-inspired microrobotic systems hold promise in medicine and di-

nal magnetic fields. We have provided a proof-of-concept experiment for a rigid reciprocal swimmer (Fig. 4) in low Reynolds number regime and demonstrated that propulsion cannot be achieved as this swimmer does not impart momentum in the medium. We have also shown a fabrication technique (Fig. 7) to develop microrobots with a similar morphology to sperm cells. The flexibility of the artificial flagellum of the sperm-shaped microrobot enables wave propagation (Fig. 5) and breaking of the time-reversal symmetry. The motion of this microrobot is modeled using elastohydrodynamics approach and Timoshenko-Rayleigh beam theory, and we find agreement with the frequency response experimental results. This chapter also provides general control strategy to orient the microrobots towards a desired direction. This strategy is tested on a helical microrobot and a sperm-shaped microrobot and enables motion in three- and two-dimensional spaces. We have also showed the helical microrobot can be used to clear blood clots \textit{in vitro} inside catheter segment and using rotating dipole fields (Fig. 12).

Biologically-inspired microrobotic systems hold promise in medicine and di-
verse biomedical applications. Very recently, Servant et al. have achieved magnetically controlled navigation of a swarm of artificial bacterial flagella in the peritoneal cavity of a mouse in vivo [54]. However, a few challenges still remain to achieve a comprehensive intervention in animal experimentation, such as the fabrication of these microrobots using biodegradable materials and the ability to deliver therapeutic agents locally in the presence of fluid flow.

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