Acceleration of Tissue Plasminogen Activator Mediated Thrombolysis by Magnetically Powered Nanomotors

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Supplementary Information

1. Experimental information

1.1 In-vitro experimental setup

As shown in Figure S1(a), the experimental setup consists of a magnetic field generator, a PDMS plate with test samples inside, a light pad, and a data acquisition system. Particularly, the rotating magnetic field is generated by four solenoids (Air-core Solenoid #14825, Science Source, Waldoboro, ME), which are connected to two independent power supplies and controlled by a computer. The PDMS plate is suspended at the center of the solenoids by a glass slide with printed scales (green-dash lines), and is placed above a light pad (LightPad A920, Artograph, Delano, MN) which can provide a uniform and stable white light illumination for

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quantitative data extraction. The experimental data are obtained by a digital camera (Infinity 1, Lumenera Corp, Ottawa, Canada) monitoring PDMS channel from the top.

Figure S1 In-vitro experimental setup. (a) An image of the experimental setup consisting of the magnetic field generator, light pad, PDMS plate, and a microscope. (b) Numerical simulation of the rotating magnetic field in one cycle/period. (c) Experimental calibration of the magnetic field induced by a single solenoid.

Figure S1(b) shows the top view of the magnetic field generated by the solenoids, calculated by Biot-Savart law using Matlab (The MathWorks, Natick, MA). For the two pairs of solenoids, the input alternating currents have the same magnitude and frequency, but with a phasic lag of π/2.

Figure S1(b) also depicts the magnetic field distribution within a 4R×4R blue-dash frame, where
$R$ is the radius of the solenoid. From the field plots at different phase angle in one revolution, one can see the field is relatively uniform in the central area where the PDMS plate is placed. To verify the validity of the theoretical calculation, the field strength is measured to compare with the numerical field distribution. As shown in Figure S1(c), a Gauss meter (Model 5080, F. W. BELL) probe through a homemade delrin frame embedded in a solenoid is used to measure the magnetic field strength in $x$ direction at different locations, the theoretical prediction is consisted with the experimental data.

![Figure S2](image)

**Figure S2** Fabricating clot inside PDMS channels. (a) PDMS channels plugged with clot, shown as light-yellow segments. (b) A clot pricked from a PDMS channel after fully reaction.

A novel method is developed to fabricate the length-controlled clot sections in PDMS channels. The clotting process is slowed down by mixing all the reagents at about 0°C, and the flow-able “clot” is injected into the PDMS channel to form uniform clot segments as described in the Experimental section in the main text. The light-yellow sections in the PDMS channels, as
shown in Figure S2(a), are the clot segments. Figure S2(b) shows the tangled clot after it is removed from a channel after fully reacted.

1.2 In-vitro experimental data treatment method

The boundary moving is determined by the dye front moving in the PDMS channel. Since Rhodamine B (RDB) is a red colored dye, the grayscale in green color channel of the video image represents the RDB absorbance, which will depend on the RDB concentration and gives the best contrast, as shown in Figure S3(a) and (b).

![Figure S3](image_url)

**Figure S3** Data treatment process of *in-vitro* experiments. (a) Original image of a Dye+Rod channel at the beginning of the experiment with x-y coordinate definition. (b) The individual RGB color channel of the original image. (c) The calibration curve of the grayscale of the green channel $G$ vs. RDB concentration $c$.

It is assumed that the transmitted green light intensity $I$ follows Beer’s Law,
\[ \frac{I}{I_0} = e^{-\alpha c}, \]  
(Eq.S1)

where \( I_0 \) is the incident green light intensity before the PDMS channel filled with RDB solution, \( c \) is the RDB concentration, \( \alpha \) is the absorption coefficient, and \( l \) is the thickness of RDB solution, or the thickness of the channel. The grayscale of the image should be proportional to the transmitted light intensity,

\[ \frac{G}{G_0} = \frac{I}{I_0} = e^{-\alpha c}, \]  
(Eq.S2)

where \( G \) and \( G_0 \) are the corresponding grayscale of the green channel of the video image. The calibration curve is shown in Figure S3(c).

During the thrombolysis process, the grayscale value \( G \) of each channel is first averaged in y-direction,

\[ G(x) = \frac{1}{h} \sum_y G(x,y), \]  
(Eq.S3)

where \( h \) is the height of the image or channel width. Thus, \( G \) is a function of \( x \) coordinate only. The concentration of RDB is able to be estimated by,

\[ c(x) = -\frac{1}{\alpha l} \ln \frac{G(x)}{G_0}. \]  
(Eq.S4)

The boundary of the reaction solution and clot interface \( X_{\text{boundary}} \) is defined at where \( \frac{dc}{dx} \) reaches the maximum value. The clot dissolving speed can be extracted by linear fitting of function \( X_{\text{boundary}}(t) \) vs. time \( t \).
To obtain the diffusion coefficient of RDB, the concentration curve at different time $c(x,t)$ is
smoothened by averaging the five neighboring data points (5-point average) except at two boundaries,

$$c_{\text{smooth}}(x,t) = \frac{1}{5} \sum_{x-2}^{x+2} c(x,t).$$  \hspace{1cm} \text{(Eq.S5)}

Since the solution of 1D diffusion equation $\frac{\partial c(x,t)}{\partial t} = D \frac{\partial^2 c(x,t)}{\partial x^2}$ is an error function:

$$c(x,t) = A \int_{\sqrt{Dt}}^{\infty} e^{-y^2} dy.$$  \hspace{1cm} \text{(Eq.S6)}

It indicates that the first derivative of $c$ with respect to $x$ will follow a Gaussian distribution,

$$\frac{\partial c(x,t)}{\partial x} = Ae^{-\frac{(x-\mu)^2}{\sigma^2}}.$$  \hspace{1cm} \text{(Eq.S7)}

Thus, the derivative $\frac{\partial c_{\text{smooth}}(x,t)}{\partial x}$ can be fitted by a Gaussian function, and the width of the
Gauss function follows,

$$\sigma^2 = Dt.$$  \hspace{1cm} \text{(Eq.S8)}

Therefore, the diffusion constant $D$ is obtained by liner fitting of $\sigma^2$ vs. time $t$.

1.3 In-vivo experimental setup

The rotational magnetic field employed for a rat embolic model is induced by permanent
magnets. As schematically shown in Figure S4(a), the device consists of two identical NdFeB
magnets of rectangular shape (NdFeB, BY0X08DCS, K&J, Pipersville, PA), a homemade
rotating frame and an AC motor (US560-001U2, Oriental Motor, Torrance, CA), of which the
rotating speed can be continuously adjusted and digitally read from a indicator/controller
The permanent magnets are symmetrically mounted on the frame with their poles attractively facing each other. The rotating frame is made of aluminum and the mounting position is adjustable. Thus, the magnetic field distribution between the two magnets can be continuously adjusted for a designated value of local magnetic field strength at the infected hindlimb of rat (red-dash circle on rat paint).

Figure S4 *In-vivo* experimental setup. (a) Schematic illustration of the apparatus. (b) Magnetic field distribution between two permanent magnets at different separations.

Figure S4(b) also plots four representative field distributions induced by the magnets in different locations. From the results, one can see the field strength is pretty uniform at the center of the magnet system. Its value varies from 40 mT to 200 mT, which satisfies the *in-vitro* experimental conditions.
2. Theoretical analysis on thrombolytic enhancement

2.1 Diffusion-controlled reaction of thrombolytic process

The clot remove process is a classic mass transport governing reaction process, and it could be simplified into 3 sub-process: (1) t-PA molecules (T) diffuse to the fibrin surface and bind onto fibrin lysine sites (S) to form t-PA-lysine complex (ST); (2) t-PA-lysine complex (ST) activates plasminogen into plasmin which cleaves fibrin into soluble product (P); (3) Product (P) desorbs from the fibrin surface and exposes new lysine sites, as proposed in Figure 4(b) in the manuscript and expressed in Eq. S9,

\[ S + T \xrightarrow{k_T} ST \xrightarrow{k_{TP}} SP \xrightarrow{k_P} S + P \]  

(Eq.S9)

Here, \( k_T \) is the adsorption rate of t-PA molecules on the fibrin surface and \( k_P \) is the desorption rate of product \( P \) removing from the fibrin surface. The process \( ST \rightarrow SP \) involves multiple steps of bio-chemical reactions, and the intrinsic reaction rate is characterized by \( k_{TP} \).

The thrombolysis rate is determined by how fast sub-processes (1) – (3) happen sequentially and is dominated by the slowest sub-process. For different concentration of t-PA, the overall thrombolysis could either belong to a diffusion limited reaction or a reaction limited process. The rate of diffusion limited reaction would depend closely on the t-PA bulk concentration, while a reaction limited process will be independent on the t-PA concentration, as well known in all chemical reaction systems. To confirm whether our current experiment condition (t-PA concentration = 50 μg/ml) belongs to a diffusion- or reaction- limited process, we have conducted a t-PA concentration dependent clot lysis speed experiment in the eight channels of our fluidic system, and the result is shown in the Figure S5 below. From Figure S5, when \( C_{tpa} \geq 200 \mu g/mL \) we can see the clot lysis speed \( v_T = 55 \mu m/min \) and it is saturated at higher \( C_{tpa} \). At \( C_{tpa} = 50 \mu g/mL \), the clot lysis speed \( v_T \) is about 35 μm/min and depends strongly on \( C_{tpa} \), which
is a clear demonstration of diffusion limited process, \( i.e., \) the transport (diffusion) of t-PA to the surface of clot is the slowest step to determine the overall reaction rate. Therefore, the t-PA molecular mobility or the diffusion coefficient can be enhanced through the rotating nanorods, and thus lead to enhanced reaction rate. Furthermore, the maxima clot lysis speed is limited at about 2 times as the speed at \( C_{\text{tpa}} = 50 \, \mu\text{g/mL} \), which is consistent to the rod enhanced experiment.

![Figure S5](image)  
**Figure S5** The clot boundary moving speed \( v_T \) *versus* the t-PA concentration \( C_{\text{tpa}} \)

2.2 *Theoretical model of thrombolytic enhancement*

As shown in Eq. S9, assuming the coverage of ST on the surface is \( \Theta_1 \) while the coverage of SP is \( \Theta_2 \), and the coverage for the vacant lysine sites should be \( 1 - \Theta_1 - \Theta_2 \). Then the rate equations for \( \Theta_1 \) and \( \Theta_2 \) can be expressed as,
The removal rate of product $P$ with $N_p$ available sites is,

$$\frac{dN_p}{dt} = k_p \Theta_2.$$  \hfill (Eq.S11)

Under equilibrium, Eq.S10 gives,

$$\begin{align*}
\Theta_1 &= \frac{k_T k_p}{k_T k_{TP} + k_{TP} k_p + k_p k_T} \\
\Theta_2 &= \frac{k_T k_{TP}}{k_T k_{TP} + k_{TP} k_p + k_p k_T}.
\end{align*}$$  \hfill (Eq.S12)

Combining Eq.S11 and Eq.S12, we have,

$$\frac{dN_p}{dt} = \frac{k_T k_p k_{TP}}{k_T k_{TP} + k_{TP} k_p + k_p k_T}.$$  \hfill (Eq.S13)

Eq.S13 shows that the thrombolysis speed is determined by the reaction coefficients of all three steps in the thrombolytic model of Eq.S9. According to the experimental observation, however, if we consider a diffusion limited process, i.e, $k_{TP} \gg k_T$ and $k_p \gg k_T$, Eq.S13 reduces to,

$$\frac{dN_p}{dt} = k_T.$$  \hfill (Eq.S14)

If every productive molecule $P$ frees a volume $V_p$ from solid clot by dissolving process, then the observed clot dissolving speed in our experiments can be expressed as,

$$v_{r,T} = \frac{V_p}{A_c (1 - \Phi)} k_T,$$  \hfill (Eq.S15)

where $A_c$ is the cross-section area of the PDMS channel, $\Phi$ is the porosity of clot which can be assumed to be a constant.
According to the von Smoluchowski’s equation, the classic result of conjugation rate $k_T$ between t-PA molecules and lysine sites is given as\(^1\),

$$k_T = 4\pi \left( D_{tPA} + D_{lysine} \right) \left( R_{tPA} + R_{lysine} \right) C_{tPA},$$

(Eq. S16)

where $D_{tPA}$, $D_{lysine}$ represents the molecular diffusivities of t-PA and lysine site. $R_{tPA}$, $R_{lysine}$ is the radius of t-PA molecules and lysine sites. $C_{tPA}$ is the bulk t-PA concentration. Since the vacant lysine sites on clot surface are not moving, **Eq.S16** becomes,

$$k_T = 4\pi D_{tPA} \left( R_{tPA} + R_{lysine} \right) C_{tPA}.$$  

(Eq.S17)

From **Eq.S17**, the reaction rate $k_T$ is proportional to the diffusivity $D_{tPA}$ and the concentration $C_{tPA}$ of t-PA molecules. With rotating rods induced flow field in our experiments, apparently, $D_{tPA}$ contains two parts: thermally induced diffusion $D^T_{tPA}$ and convectional flow enhanced diffusion by rotating rods $D^C_{tPA}$\(^2,3\),

$$D_{tPA} = D^T_{tPA} + D^C_{tPA}.$$  

(Eq.S18)

$D^T_{tPA}$ can be estimated from Einstein’s theory (see the result below), while the expression of $D^C_{tPA}$ depends on the Peclet number $Pe$ according to convection theory,

$$Pe = \frac{\bar{u} d_R}{D^T_{tPA}},$$

(Eq.S19)

where $\bar{u}$ is the average velocity of the flow field and $d_R = C_{NR}^{-1/3}$ is the size of each cellular flow induced by a rotating rod, confined in a cylindrical volume with height and diameter of $d_R$, as shown in **Fig. 3(b)** (for the convenience of derivation, $C_{NR}$ is the number concentration of rods, which is different from the mass concentration of rod $C_R$ defined in the manuscript). Due to the low speed and small size, the rotating rods are modeled as concentrated torques inducing Stokes’ creeping flow at low Reynolds number. For a singular point torque $\vec{M}$ at the origin, the strength
of rotlet $\vec{\gamma}$ in a closed control surface is \(^4\),

$$\vec{\gamma} = \frac{\vec{M}}{8\pi \eta},$$

(Eq.S20)

where $\eta$ is the viscosity of liquid.

While the velocity of a point on the surface positioning of vector $\vec{r}$ equals,

$$\vec{u} = \frac{\vec{\gamma} \times \vec{r}}{|\vec{r}|^3},$$

(Eq.S21)

Thus, the velocity field induced by a rotating nanorod can be expressed as,

$$\vec{u} = \frac{\vec{M} \times \vec{r}}{8\pi \eta |\vec{r}|},$$

(Eq.S22)

Obviously, the hydrodynamic volume by a single rod is cylindrically symmetric with induced velocity decaying as $r^{-2}$. If more than one rod are put in the liquid with a uniform distribution and applied by an equal driven torque $M$, the flow velocity by multi-rotating rods can be averagely estimated as,

$$\bar{u} = \frac{\sum_{i=1}^{n} |\vec{u}| d\Omega}{V_L} = \frac{|\vec{M}| C_{NR}^{2/3}}{4\eta \ln(1+\sqrt{2})},$$

(Eq.S23)

Here, $\Omega_i = \pi d_r^3 / 4$ is the hydrodynamic volume of a single rod, $V_L$ is the total volume of the liquid. At the low Reynolds, the singular point torque $|\vec{M}|$ by a rotating nanorod equals to the hydrodynamic drag torque on it,

$$\vec{M} = 2 \int_0^{L/2} (\vec{D}_l \times \vec{l}) dl = \frac{-2\pi^2 \eta \varepsilon L^3 f}{3},$$

(Eq.S24)

where $\vec{D}_l = 4\pi \eta \varepsilon U(l)$ is the Lamb drag per unit length of rod\(^5\), which is proportional to the distribution of flow velocity along the rod $U(l) = 2\pi f l$ and the shaped determined constant
\[ \varepsilon = \left( \ln \left( \frac{L}{d} \right) - 0.66 \right)^{-6}. \]

For a rod of length \( L = 1 \mu m \) and diameter \( d = 0.5 \mu m \) rotating at frequency \( f = 20 \) Hz, the induced torque \( M = 3.97 \times 10^{-18} N \cdot m \). With all known conditions, for example, when \( C_R = 7 \) mg/mL, the highest concentration used in our experiments, the average hydrodynamic velocity induced by rotating rods \( \bar{u} \) is about \( 11 \mu m/s \); while \( C_R = 1 \) mg/mL, \( \bar{u} = 3 \mu m/s \). Since the diameter of t-PA soluble molecule is about \( 10 \) nm, the thermal diffusivity of t-PA molecules without rods can be estimated using Einstein’s theory,

\[ D_{qpa}^T = \frac{k_B T}{6\pi \eta R_{qpa}} = 4.5 \times 10^{-11} m^2 / s, \quad (\text{Eq. S25}) \]

where \( k_B, T \) and \( \eta \) are the Boltzmann constant, temperature, and viscosity of water, respectively.

According to results of \textbf{Eq.S23}, with \textbf{Eq.S25}, we have Peclet number \( Pe \) in \textbf{Eq.S19} varies from 1.6 to 3.1 when \( C_R \) changes from 1 to 7 mg/ml in our experiments. When the Peclet number \( Pe \) is small \( (Pe<3)^2 \), the flow scaling law is valid for rigid-boundary conditions and the convectional flow enhanced diffusivity \( D_{qpa}^C \) in \textbf{Eq.S18} can be estimated as \( ^3, ^8, \)

\[ D_{qpa}^C = aPe^2D_{qpa}^T, \quad (\text{Eq.S26}) \]

where \( a \) is a constant regarding to the flow field.

Combining \textbf{Eq.S15-Eq.S26}, the theoretical prediction of thrombolysis observed in our experiments shall obey the following relation with respect to nanorod concentration \( C_R \),

\[ v_{R+T} = v_T + \alpha C_R^{2/3}, \quad (\text{Eq.S27}) \]

where, \( v_T = \kappa C_{qpa} D_{qpa}^T \), where \( \kappa = \frac{4\pi R_{qpa} V_p}{A_c (1-\Phi)} \), \( \chi \) is a unit convertor so that \( C_{NR} = \chi C_R \), and

\[ \alpha = \kappa C_{qpa} \frac{0.0486a M^2}{\eta^2 D_{qpa}^T}\chi^{2/3}. \]

When \( C_R = 0 \), \textbf{Eq.S27} degrades to,
\[ v_r = \kappa C_{tpa} D_{tpa}^T. \]  

(Eq.S28)

Here, the thrombolysis speed \( v_T \) (without rods) is proportional to \( C_{tpa} \). This is true in the region of low concentration of t-PA, as shown in Figure S5. Obviously, \( C_{tpa} = 50 \, \mu g/ml \) for the rod enhanced thrombolysis is in the linear region of our experiments. Thus, the coefficient \( \kappa D_{tpa}^T \) in Eq.S28 can be extracted from the slop of the curve in the linear region in Figure. S5, as shown in the insert plot, and we obtain \( \kappa D_{tpa}^T = 3.03 \times 10^{-6} \, m^4 \cdot kg^{-1} \cdot s^{-1} \). Since \( D_{tpa}^T \) is given by Eq.S25, we have \( \kappa = 6.69 \times 10^4 \, m^2 \cdot kg^{-1} \). In addition, the coefficient \( a = 2/3 \pi^2 \) in the spatially periodic hydrodynamic flows\(^8\), therefore,

\[
\alpha = \kappa C_{tpa} \frac{0.0486a|M|^2 \chi^{2/3}}{\eta^2 D_{tpa}^T} = 1.59 \times 10^{-4} \, ml \cdot mg^{-2/3} \cdot min^{-1}. \tag{Eq.S29}
\]

Experimentally we obtain \( \alpha = (9 \pm 2) \times 10^{-4} \, ml \cdot mg^{-2/3} \cdot min^{-1} \), which is in the same order of magnitude estimated by the theory (Eq.S19). The discrepancy between the experimental value and the theoretical value could be due to the hydrodynamic effect of “rod cluster”.

2.3 The effect of nanorods aggregation

The nanorods were treated by PVP to enhance the stability. When no external magnetic field applied, the rods of both low and high concentration suspended in the solution are uniformly distributed. However, it is observed they will form chain-like structures under a static external magnetic field. When the magnetic field is rotating, such chain structure will either maintain to rotate as “big rod” or break down into rods clusters as individual rotates\(^9\). We performed an experiment to test the cluster size of nanorod solution at different concentration under current experiment condition. If the cluster is not circular, the longest diameter will be used. Figure S6 shows the average cluster size versus nanorod concentration for our experiments.
Figure S6 The cluster size distribution (black) and the theoretical prediction of the upper-bound thrombolysis with uncertainties (red) versus different rod concentration $C_R$.

From the Figure S6, the average cluster size $n = C_{NR} / C_{\text{cluster}}$, defined the average number of rods in a cluster, and the uncertainty of the cluster size increases by the rods concentration. (Note that the error bar of cluster size $n$ is up to ±17.28, and of theoretical thrombolysis speed $v_{R+T}$ is up to ±327.51 μm/min at $C_R = 8$ mg/ml.) If each cluster can be treated as a rigid particle, Eq.S27 can be modified as,

$$v_{R+T} - v_T = \Delta v = \alpha_{\text{cluster}} C_{\text{cluster}}^{2/3} = \kappa C_{\text{p}} \frac{0.0486 \alpha |\vec{M}|^2}{\eta^2 D_{\text{p}}} C_{\text{cluster}}^{2/3},$$

(Eq.S30)

where $|\vec{M}|_{\text{cluster}}$ and $C_{\text{cluster}}$ are the torque induced by a single cluster and the concentration of the torques, both of which depend on the cluster size. However, it is very difficult to estimate the magnitude of $|\vec{M}|_{\text{cluster}}$ using the hydrodynamic drag torque because the shape of the cluster varies location by location. On the other hand, even though we assume the driven torque induced by the magnetic field equals to the resistant torque by the hydrodynamic drag at the static rotation of a
cluster, the arrangement of rods in a cluster is unknown. Thus, we cannot simply employ the
relation $|\vec{M}_{\text{cluster}}| = \sum |\vec{M}_{\text{rod}}|$. However, we know that,

$$|\vec{M}_{\text{cluster}}| \leq n |\vec{M}_{\text{rod}}|.$$  \hspace{1cm} (Eq.S31)

Plug Eq.S31 into Eq.S30, we have,

$$\Delta \nu \leq \frac{0.0486 \alpha \kappa C_{\text{npa}} |\vec{M}|^2 C_{\text{NR}}^{2/3}}{\eta^2 D_{\text{npa}}} n^{4/3} = \alpha' C_{\text{NR}}^{2/3},$$  \hspace{1cm} (Eq.S32)

where $\alpha' = \alpha n^{4/3}$ is the new proportionality of enhanced thrombolysis $\Delta \nu$ with respect to $C_{\text{NR}}^{2/3}$ due to the presence of cluster. Then we can estimate the error bar of $v_{R+T}$ using the uncertainty of rod-cluster size,

$$\delta v_{R+T} = \left| \frac{\partial \Delta \nu}{\partial n} \right| \delta n \leq \frac{4}{3n} \alpha' C_{\text{NR}}^{2/3} \delta n.$$  \hspace{1cm} (Eq.S33)

According to Eq.S33 and Figure S6, the error bar of thrombolysis is predicted to increase by the rod concentration $C_{R}$, which is consistent to the trend of the experimental data in Figure 3(d). Moreover, the term of $n^{4/3}$ in Eq.S32 suggests the elevation of thrombolysis due to the hydrodynamic effect of clusters rather than rods.

2.4 The role of mechanical stimulation

The nanorods indeed have mechanical interaction with clot fiber. However, the force due to this interaction is too weak to break the clot fiber. Two evidences support this claim.

First, there is always a control channel only with nanorods in our experiments like channel R in Figure 3(b). In this channel, a lysis process can never been observed, instead, a diffusion process has always been observed as we reported in the manuscript (Page 6 Line 1). Figure S7 shows the first derivative of dye concentration profile and the corresponding Gaussian fittings of $R$ and $D$ channels in Figure 3(b). The parameter $\mu$ describes the central location of the Gaussian curve
and could be treated as the position of clot/dye diffusion boundary.

**Figure S7** Change rate of dye concentration along the clot region

**Figure S8** shows the plot of $\mu$ vs. $t$ and simple liner fittings are given. According to **Figure S8**, the clot/dye boundary is moving in both $D$ and $R$ channels, and has almost the same speed (about 9 $\mu$m/min).

**Figure S8** Plot of $\mu$ vs. $t$ and liner fitting
This speed is coming from the staining process of clot since the clot is light yellow color and the stained clot appears a darker color than the dye solution. So the central location of the Gaussian curve will move accordingly. However, this does not mean that the clot is being dissolved. If the mechanical stimulation could enhance the clot lysis, we should expect a very different boundary moving speed in the $R$ channel from in the $D$ channel (a large slope in Figure S8), which is not the case in our observation.

Second, the mechanical force on a fibrin fiber by a rotating rod can be theoretically estimated. The Ni nanorod is modeled as a point-like magnet, as shown in Figure S9. Inside a uniform magnetic field, the magnet is subjected to a torque $\vec{T}$,

$$\vec{T} = \vec{m} \times \vec{B}, \quad \text{(Eq. S34)}$$

where $\vec{m}$ is the dipole moment of the magnet, and $\vec{B}$ is the external magnetic flux. The torque is of the maximum magnitude when the dipole moment and the external magnetic flux are perpendicular to each other, and equal to zero when they are aligned with each other. The maximum torque, $T_{\text{max}}$ is given by,

$$T_{\text{max}} = mB, \quad \text{(Eq. S35)}$$

where $m$ and $B$ are the magnitudes of the corresponding vectors.
**Figure S9**: Schematic showing a magnet within a magnetic field.

Assuming that the Ni nanorods are magnetized along their longitudinal axes with magnetization $M$, the maximum torque is given by,

$$T_{\text{max}} = MBA,$$  \hspace{1cm} (Eq.S36)

where $V$ is the volume of the nanorod. The saturation magnetization of crystalline Ni is about 58.5 emu/g$^{10}$, and that of amorphous Ni is about one half of the value.$^{11}$ The mass density of Ni is 8900 kg/m$^3$. For a Ni nanorod with length $l = 5 \ \mu m$ and diameter $d = 0.2 \ \mu m$, the volume, $V = 0.157 \ \mu m^3$. Let us assume $B = 0.1$ T, which is an overestimation in comparison to the field used in our experiments ($B = 8$ mT). Substituting all these parameters (note $M$ set to be 32.9 emu/g for amorphous Ni) in Eq.S36, we can estimate $T_{\text{max}} = 4.60$ fN·m. It means that for the characteristic length scale down to the fibers of a fibrin clot, which should be ~ micron, such a nanorod/clot interaction can generate forces on the order of Nano Newton. On the other hand, according to the literature, it takes about 100 pN to unfold a fibril$^{12}$, and several thousands of parallel fibrils can be counted in a fiber cross section$^{13}$. It means that it would take a force of ~ 1 μN to cause any damage to a fibrin fiber. As estimated above, the Ni nanorod can only exert a force about 1 nN. Thus, these nanorods cannot do any damage to individual fibers.

Therefore, from both experimental and theoretical analyses, we could conclude that the mechanical interaction between the magnetic nanorods and clot could hardly play an appreciable role in our present experiments of t-PA mediated thrombolysis.
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