# Acs applied materials and interfaces template
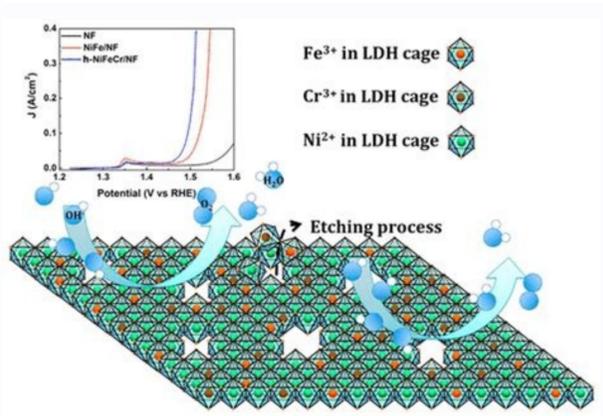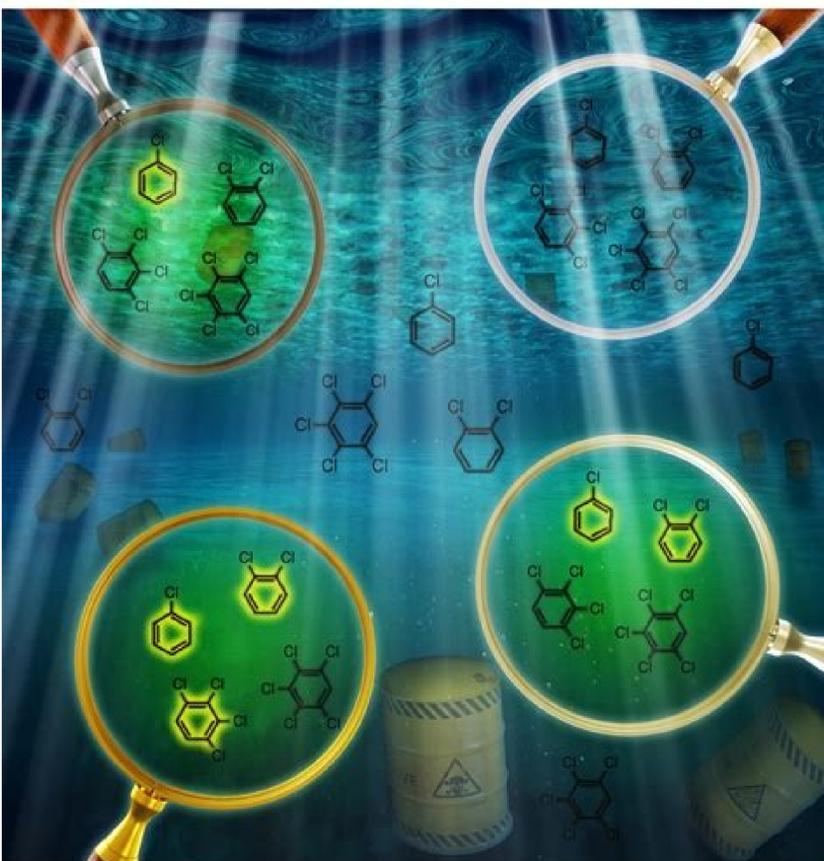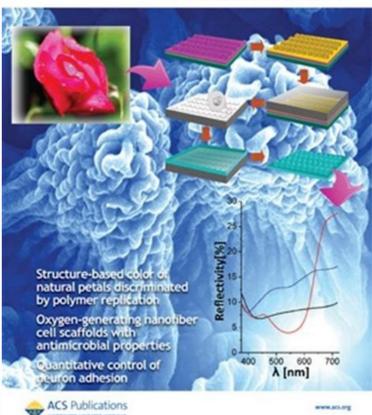
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**I'm not robot!**

to solve the cardiac dynamics of transmembrane potential equation. In this work, the forward Euler finite difference method is extended to 3D space, as in Eq. (8):

$$
\begin{cases}
V_m(x,y,z,t+\Delta t) = V_m(x,y,z,t) + \Delta t \cdot \left( -\dfrac{I_{stim} + I_{ion}}{C_m} + \right. \\
D_{global} \left( \dfrac{V_m(x-1,y,z,t) - 2V_m(x,y,z,t) + V_m(x+1,y,z,t)}{\Delta x^2} \right. \\
+ \dfrac{V_m(x,y-1,z,t) - 2V_m(x,y,z,t) + V_m(x,y+1,z,t)}{\Delta y^2} \\
\left. \left. + \dfrac{V_m(x,y,z-1,t) - 2V_m(x,y,z,t) + V_m(x,y,z+1,t)}{\Delta z^2} \right) \right)
\end{cases}
$$

The stability condition of the forward Euler difference equation is as follows:

$$
R_{\mu\nu} - \tfrac{1}{2} R g_{\mu\nu} + \Lambda g_{\mu\nu} = \frac{8\pi G}{c^4} T_{\mu\nu}
$$

$$
\Delta t \le \frac{\Delta \varsigma^2}{2d \cdot D_{global}}
$$

where $\Delta\varsigma$ is the spatial minimum resolution of 0.33 mm, $D_{global}$ demonstrates the maximum electrical conductivity of 0.154 mm$^2$/ms and $d$ is the spatial dimension. Here, $d = 3$ for 3D ventricle models. Thus we have iterative time step $\Delta t \le 0.118$ ms, i.e. the maximum stable time step $\Delta t_{max}$ is 0.118 ms. In this paper, we use $\Delta t = 0.02$ ms.

For the complex geometries of human ventricular tissue, precise treatment of boundary conditions would be very complicated. Many researchers have used Neumann no-flux boundary conditions for this problem, [46] which is shown as Eq. (10):

$$
n \cdot D_{global} \nabla V_m = 0
$$

where $n$ is the vector normal to the surface, means that there is zero current flow normal to the ventricular tissue boundaries.

In this paper, to solve the no-flux boundary conditions, a phase-field method is employed.

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be obtained. The linear mapping which is achieved by each LLUP is as follows:

$$\Theta_e(V_e(x)) : (\alpha_e(x), L_e(x))$$

$$\Theta_a(V_a(x')) : (\alpha_a(x'), L_a(x'))$$

where $V_e(x)$ is the action potential value and $V_a(x')$ is the tissue gray value. $\alpha_e(x)$ and $L_e(x)$ are opacity and light intensity of $s$ in the volume of cardiac excitation conduction model respectively. $\alpha_a(x')$ and $L_a(x')$ are the analogous variables in anatomy structure volume. Thus the merging attenuation function (MAF) for the cardiac electrophysiology-morphous merging visualization is defined by:

$$\tau'(x) = \alpha_a(x') \cdot \alpha_e(x) \cdot mef$$

$$mef = \begin{cases} ||N(x)|| & V_e(x) = -1 \\ 1 & other \end{cases}$$

Where $mef$ is the merging factor, $N(x)$ is the normalized normal vector and $||N(x)||$ is the norm of $N(x)$. $V_e(x) = -1$ means that only cardiac tissue data exists at current $x$ position on the ray. Otherwise both cardiac conduction data and tissue data are valid at the current position.

The resultant light intensity for cardiac biophysical merging visualization is computed with the novel MAF:

$$L = \int_0^t L_{merging}(x) e^{\tau'(0,x)} dx$$

$$L_{merging}(x) = \frac{\alpha_e(x)}{\alpha_e(x) + \alpha_a(x)}(L_e(x) \cdot \alpha_e(x)) + \frac{\alpha_a(x')}{\alpha_a(x') + \alpha_e(x')}(L_a(x') \cdot \alpha_a(x'))$$

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