

Impact of electric spatially discordant alternans on cardiac magnetic field

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Spatially discordant alternans (SDA) play a crucial role in cardiac arrhythmogenesis by creating steep repolarization gradients facilitating conduction block and reentry. While traditionally studied using electrical indicators, this work provides an alternative perspective by characterizing SDA through their magnetic field signatures. Using a one-dimensional idealized cardiac fiber model, we demonstrate that magnetic-field measurements effectively detect SDA and temperature-dependent changes in cardiac action potentials, offering a noninvasive alternative to conventional electrophysiological metrics. Our results reveal that the spatial organization of SDA is mirrored in the distribution of the magnetic field, with SDA nodes clearly identifiable via spatial mapping. Notably, magnetic restitution curves exhibit a distinct pattern from action potential duration (APD)-based indicators, closely following the dynamics of the action potential upstroke. These findings establish the cardiac magnetic field as a powerful diagnostic tool for detecting SDA, opening alternative avenues for biomagnetic monitoring of arrhythmic risk.

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I. INTRODUCTION

Cardiac arrhythmias are among the leading causes of sudden cardiac death, significantly contributing to global mortality [1–3]. Despite advancements in diagnosis and treatment, further investigation is needed, particularly to understand cardiac alternans, which serve as precursors to arrhythmias [4–9]. Spatiotemporal alternans dynamics have been traditionally studied as beat-to-beat variations in action potential duration (APD) and classified as spatially concordant (SCA), when APD alternates in-phase across the tissue, or spatially discordant (SDA), when different regions alternate out-of-phase. The latter creates nodes where APD remains unchanged [8,10–12]. SDA are of particular interest due to their strong association with arrhythmogenesis [13,14].

Several factors contribute to SDA onset, including ischemic damage, intrinsic tissue heterogeneities, and emergent patterns in cells and tissues [4,15–19]. Among external modulators, temperature plays a crucial role, as lower temperatures prolong APD, increasing dispersion of repolarization and arrhythmic susceptibility [11,20–25]. While hypothermia is

widely used in clinical settings to reduce cellular damage, it also presents proarrhythmic risks [26,27].

Beyond traditional electrical measurements, the magnetic field generated by cardiac electrical activity provides valuable complementary insights [28–37]. However, the main challenge in cardiac magnetic field investigations lies in the extremely low intensity of these fields, typically in the pT–nT range [32,38–43], making them challenging to detect with conventional methods and in a noninvasive way. The magnetic-field measurement of cardiac activity are now technologically feasible thanks to three quantum sensor classes: cryogenic superconducting quantum interference devices (SQUIDs) [39,40,44,45], optically pumped magnetometers (OPMs) [42,46,47], and nitrogen–vacancy (NV) diamond probes [45]. The three classes of sensors are designed to detect magnetic field in the nT–pT range produced by excitable cells such as neurons, muscles, and cardiac myocytes [39,41–51]. State-of-the-art for SQUID indicates a noise floor of 10–30 fT/ $\sqrt{\text{Hz}}$ with a sensitivity of ~ 1 fT/ $\sqrt{\text{Hz}}$ and pickup coils ≥ 3 mm, yielding millimetric spatial resolution (5 to 10 mm), but requiring 4.2 K cooling and magnetically shielded rooms. OPMs operate at ambient temperature with a sensitivity of ~ 10 fT/ $\sqrt{\text{Hz}}$ and smaller spatial resolution (3 to 5 mm) but require shielding rooms. NV-center magnetometers bridge these regimes: single-crystal sensors provide a ~ 10 pT/ $\sqrt{\text{Hz}}$ sensitivity with a nanoscale spatial resolution under physiological conditions.

Mathematical models have significantly contributed to expand our understanding of cardiac dynamics in health and

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disease, as they can offer deep insights into the biophysical mechanisms at the basis of cardiac arrhythmias onset and development [23,24,52,53]. In this work, we aim to expand the domain of cardiac model applications. Mathematical models of cardiac magnetic field [29,54] highlighted its potential investigation, demonstrating that the magnetic field produced by cardiac activity could convey physical information that is not included in the sole electric field [29,55].

In the present work, we analyze the initiation and progression of cardiac alternans in a one-dimensional homogeneous cable under varying thermal conditions, tissue heterogeneity, based on a thermoelectro-magnetic modeling framework recently developed [54]. In particular, we here apply this approach to the investigation of alternans and the development of SDA regimes. Using standard pacing-down restitution protocols, we derive the magnetic equivalent of restitution curves and introduce an approach to characterizing SDA alternans based on magnetic field dynamics. This work provides a methodological framework for biomagnetic analysis of cardiac alternans, which can be extended to more complex spatial and temporal electrical patterns in the heart, ultimately contributing to a deeper understanding of arrhythmogenesis.

In conclusion, despite the fact that the present work is purely theoretical, the field signatures we predict are, in principle, measurable with today's instrumentation [39,41,44–49], supporting the translational relevance of the model.

II. METHODS

A. Cardiac electrophysiology modeling

We model cardiac action potential using a four-variable phenomenological model [4,54], which is suitable for describing temperature-related effects on cardiac dynamics [53] as well as for simulating full pacing down restitution protocols. Moreover, this model is particularly suitable for future extensions of the work to more complex and realistic geometries than the simple one-dimensional (1D) cable here studied. The model presents a different parameters set with respect to [54], opportunely fine-tuned to reproduce SDA in one-dimensional cable geometries. The membrane potential is described with the reaction-diffusion partial differential equation

$$\frac{\partial u}{\partial t} = D_0 \nabla^2 u - (J_{fi} + J_{si} + J_{so}), \quad (1)$$

where u is the nodimensional membrane potential, D_0 is the diffusion coefficient, while J_{fi} , J_{so} , and J_{si} represent the fast-inward, slow-outward, and the slow-inward current densities, respectively. The three currents are voltage (u) and temperature (T) dependent. Moreover, they are also regulated by three gate variables v , w , and s through the following equations:

$$J_{fi} = \eta_{fi}(T) \left[H(u - \theta_v)(u - \theta_v)(u - u_u) \frac{v}{\tau_{fi}} \right], \quad (2)$$

$$J_{si} = \eta_{si}(T) \left[H(u - \theta_w) \frac{ws}{\tau_{si}} \right], \quad (3)$$

$$J_{so} = \eta_{so}(T) \left[[1 - H(u - \theta_w)] \frac{(u_o - u)}{\tau_o} + \frac{H(u - \theta_w)}{\tau_{so}} \right], \quad (4)$$

where H indicates the Heaviside step function. The gating variables are obtained by solving the following coupled system of ordinary differential equations:

$$\frac{\partial v}{\partial t} = \Phi_v(T) \left[[1 - H(u - \theta_v)] \frac{v_\infty - v}{\tau_v^-} - \frac{H(u - \theta_v)}{\tau_v^+} \right], \quad (5)$$

$$\frac{\partial w}{\partial t} = \Phi_w(T) \left[[1 - H(u - \theta_w)] \frac{w_\infty - w}{\tau_w^-} - \frac{H(u - \theta_w)}{\tau_w^+} \right], \quad (6)$$

$$\frac{\partial s}{\partial t} = \Phi_s(T) \left[\frac{\{1 + \tanh[k_s(u - u_s)]\}/2 - s}{\tau_s} \right], \quad (7)$$

where w_∞ and v_∞ are defined as

$$v_\infty = H(\theta_v^- - u), \quad (8)$$

$$w_\infty = [1 - H(u - \theta_o)] \left(1 - \frac{u}{\tau_{w_\infty}} \right) + H(u - \theta_o) w_\infty^*. \quad (9)$$

The time constants (τ_{fi} , τ_{si} , τ_{so} , τ_o , τ_v^+ , τ_v^- , τ_w^+ , τ_w^- , τ_s) in Eqs. (2) to (7) are temperature and voltage dependent as follows:

$$\tau_v^- = [1 - H(u - \theta_v^-)] \tau_{v1}^- + H(u - \theta_v^-) \tau_{v2}^-, \quad (10)$$

$$\tau_w^+ = \tau_{w1}^+ + (\tau_{w2}^+ - \tau_{w1}^+) \frac{\tanh[k_w^+(u - u_w^+)] + 1}{2}, \quad (11)$$

$$\tau_w^- = \tau_{w1}^- + (\tau_{w2}^- - \tau_{w1}^-) \frac{\tanh[k_w^-(u - u_w^-)] + 1}{2}, \quad (12)$$

$$\tau_{so}^- = \tau_{so1} + (\tau_{so2} - \tau_{so1}) \frac{\tanh[k_{so}(u - u_{so})] + 1}{2}, \quad (13)$$

$$\tau_s = [1 - H(u - \theta_w)] \tau_{s1} + H(u - \theta_w) \tau_{s2}, \quad (14)$$

$$\tau_o = [1 - H(u - \theta_o)] \tau_{o1} + H(u - \theta_o) \tau_{o2}. \quad (15)$$

Temperature effects on the membrane potential are modeled by modulating the current densities and the three gating variables v , w , s with temperature-dependent Moore, ($\eta(T)$) and Arrhenius ($\Phi(T)$) factors, defined as follows [53]:

$$\Phi(T) = Q_{10}^{(T-T_A)/10}, \quad (16)$$

$$\eta(T) = A[1 + B(T - T_A)]. \quad (17)$$

Model parameters are listed in Table I. The models is implemented in a straight 3-cm-long cable with a spatial discretization of $dz = 100 \mu\text{m}$, so the final mesh consisted of 300 nodes. The spatial discretization has been optimized with a convergence analysis to ensure the stability of conduction velocity. The temporal discretization ($\Delta t = 0.025 \text{ ms}$) is selected to ensure an accurate description of the AP upstroke, and therefore, of the temporal characteristics of the magnetic field [54]. Temporal and spatial derivatives are approximated with the finite-difference method and solved with the Euler

TABLE I. Model parameters for endocardial action potential minimal-model formulation for a selected thermal range (29–40°C). Units are given in ms, cm, mV, mS, μF , g, °C. We use as initial conditions $u = v = 1$, $w = 0$, $s = 1$ [56]. The diffusion coefficient is $D_0 = 0.001 \text{ cm}^2/\text{ms}$.

Parameter	Value	Parameter	Value	Parameter	Value
θ_v	0.3	τ_{v1}^-	55	τ_{w1}^+	175
θ_w	0.13	τ_{v2}^-	10	τ_{w2}^+	230
θ_v^-	0.2	τ_v^+	1.4506	τ_{s01}	40
θ_o	0.006	τ_{w1}^-	6	τ_{s02}	1.2
τ_{fi}	0.10	τ_{o1}	470	k_{so}	2
τ_{o2}	6	u_w^+	0.0005	u_{so}	0.65
k_w^-	200	τ_{s1}	2.7342	τ_{s2}	2
k_w^+	8	k_s	2.0994	u_s	0.9087
u_w^-	0.00615	τ_{si}	2.9013	$\tau_{w\infty}$	0.0273
w_∞^*	0.78	$Q_{10,v}$	1.5	$Q_{10,w}$	2.45
$Q_{10,s}$	1.5	A_{fi}	1	B_{fi}	0.065
A_{so}	1	B_{so}	0.008	A_{si}	1
B_{si}	0.008				

method using zero-flux boundary conditions to ensure the confinement of the electrical activity on the wire. The model is implemented and solved in matlab R2022b (MathWorks, Natick, Massachusetts).

This simple, yet realistic, geometry is selected to keep the complexity of magnetic field calculation as low as possible while preserving a link with physiological case studies [54,57].

Case studies and stimulation protocol

In this work we study the onset and progression of spatially discordant alternans in the simplified context of 1D cables. In particular, we characterize the onset and progression of alternans depending on the thermal state of the tissue. We select four temperatures representative of severe hypothermia, 29°C, mild hypothermia, 33°C, physiological conditions, 37°C, and hyperthermia, 40°C. As tissue heterogeneity, due to structural changes or scars, is a promoting factor for SDA onset, we explore the influence of tissue heterogeneity on the magnetic field. To phenomenologically reproduce the effects of structural remodeling and scarring typically observed in postinfarct tissue, we introduce spatial heterogeneity in the model by modifying the conductivity term [58,59], making it position-dependent and specifically reduced in selected regions [60,61]. To this purpose we introduce in our system a region with a reduced diffusivity. The region extends for 0.1-cm above and below the center of the wire for a total length of 0.2 cm. The transition between the reduced and normal diffusivity regions is smoothed with a sigmoidal function. Therefore, the spatial profile of the diffusion coefficient is represented by the following equation:

$$D = \phi(z)D_0, \quad (18)$$

$$\phi(z) = \left[1 - \frac{f}{1 + \exp[-(z-z_1)/0.5]}\right] + \frac{f}{1 + \exp[-(z-z_2)/0.5]}, \quad (19)$$

where $z_1 = 1.4 \text{ cm}$ and $z_2 = 1.6 \text{ cm}$ are the initial and final coordinates of reduced diffusivity regions. The factor f

represents the amount of reduction of the diffusion coefficient, so that the final value of the diffusion coefficient is $D = (1 - f)D_0$ in the region between z_1 and z_2 . The simulations with heterogeneous diffusion coefficient are conducted at 37°C.

The onset and progression of alternans is characterized by implementing a full pacing-down restitution protocol for each case study. During the restitution protocol, the wire is stimulated with current pulses while decreasing the period, i.e., the pacing cycle length (PCL). The protocol starts with sinus rhythm gradually decreasing until conduction block arises.

B. Current density estimation

We compute the current originating the magnetic field from the simulated electrical activity applying the classical electro-magnetism laws

$$J(t, z') = -\sigma \nabla V_m(t, z'), \rightarrow J(t, z') = -\sigma \frac{\partial V_m}{\partial z'}, \quad (20)$$

where $V_m(t, z')$ represents the transmembrane potential obtained from its dimensionless equivalent u by using the map $V_m(t, z) = 83.3u(t, z') - V_{\text{rest}}$, $V_{\text{rest}} = -84 \text{ mV}$ is the typical resting potential of cardiac cells, and 83.3 is a scaling factor used to adjust the amplitude of the action potential. Tissue conductivity, $\sigma = DS_0C_m$, is estimated as in [52], assuming that our idealized fiber is homogeneous and isotropic with $C_m = 1 \mu\text{F}/\text{cm}^2$ the membrane capacitance per unit area, $D \text{ cm}^2/\text{ms}$ the diffusion coefficient, and S_0 the surface to volume ratio estimated through a standard dimension of a myocyte in cardiac fibres, i.e., $R' = 40 \mu\text{m}$, $L = 100 \mu\text{m}$.

As usual in the literature, in the above equations and in the following, primed quantities are referred to positions within the current source, i.e., the cylinder on which we simulate the electrical activity.

C. Magnetic-field modeling

The current density $J(t, z')$ obtained as described in Sec. II B is used to compute the magnetic field due to the electrical activity in the wire. Given the slow variations of the current compared to signals traveling at the speed of light, we can safely neglect electromagnetic wave phenomena and use standard magnetostatic tools to compute the magnetic field. Taking into consideration the particular geometry of the system, i.e., a straight cylinder in the z direction with $z' \in [z'_{\min}, z'_{\max}] = [0, 3] \text{ cm}$, we compute the magnetic field generated by the source applying the Biot-Savart's law

$$\mathbf{B} = \frac{\mu_0}{4\pi} \int_{\tau'} \frac{\mathbf{J} \wedge \Delta \vec{r}}{\Delta r^3} d\tau', \quad (21)$$

where τ is a cylindrical volume with a radius $r = 300 \mu\text{m}$ surrounding the wire for its entire length, $\Delta \vec{r} = \vec{r} - \vec{r}'$, $\vec{r} = (x, y, z)$, $\vec{r}' = (0, 0, z')$. We assume the current density propagating in the z direction, uniformly distributed on a circular cross section (in the xy plane) of a wire with radius $R' = 40 \mu\text{m}$, which is approximately the radius of cardiac myocyte. This assumption allow us to compute the current intensity as

$$I(t, z') = \int_{S'} \vec{J}(t, \vec{r}') \cdot d\vec{S}' = J(t, z')\pi R'^2. \quad (22)$$

Therefore, we can write Eq. (21) substituting the current density with the corresponding current intensity flowing in a 1D wire along the z axis

$$\vec{B} = \frac{\mu_0}{4\pi} \int_{z_{\min}}^{z_{\max}} \frac{I(t, z') d\vec{l}' \wedge \Delta\vec{r}}{|\Delta\vec{r}|^3}. \quad (23)$$

Due to the assumption we made on the current density, the magnetic field produced by the propagation of the cardiac action potential along the wire only preserves x and y components, and can be obtained by computing the following 1D integrals:

$$\begin{aligned} \vec{B} = & -\frac{\mu_0}{4\pi} \int_{z_{\min}}^{z_{\max}} \frac{I(t, z') y dz'}{[x^2 + y^2 + (z - z')^2]^{\frac{3}{2}}} \hat{i} \\ & + \frac{\mu_0}{4\pi} \int_{z_{\min}}^{z_{\max}} \frac{I(t, z') x dz'}{[x^2 + y^2 + (z - z')^2]^{\frac{3}{2}}} \hat{j}, \end{aligned} \quad (24)$$

with $\hat{i}$ and $\hat{j}$ indicating the x and y versors, respectively. For further details on the calculation of the magnetic field we refer the reader to [54].

The magnetic field, Eq. (24), is computed on a cylindrical domain ($R = 300 \mu\text{m}$, $L = 3 \text{ cm}$) using the trapezoidal approximation method. The domain is discretized with the following spatial grid: $\Delta x = \Delta y = 10 \mu\text{m}$ and $\Delta z = 100 \mu\text{m}$. The spatial grid is selected to ensure the stability of the integration routine and to accurately reproduce the characteristics spatial variations of the magnetic field.

The cylindrical domain on which Eq. (24) is evaluated is assumed to have the same electromagnetic properties of the vacuum. In this regard, it is worth mentioning that the tissue is immersed in saline conductive solutions, in general. The electrical activity inside the tissue could induce currents in the extracellular medium, which could partially cancel the magnetic field produced by intracellular currents [62]. In the present work we compute the magnetic field within a distance of $50 \mu\text{m}$ from the source. In this approximation, the effects of extracellular return currents can be safely neglected since the spatial extension of the activation wavefront is significantly larger than the distance at which the field is evaluated [62].

D. Cardiac alternans indicators

We analyze the proarrhythmic phenomenon of cardiac alternans both from the electric and magnetic point of view, comparing the traditional indices, as the action potential duration (APD), with the alternative magnetic ones.

For each PCL, we compute the APD_{80} of the last two beats, thresholding the simulated AP at 80% of repolarization.

The magnetic activity associated with AP propagation is investigated by computing the maximal value of the magnetic field norm at a fixed distance from the source ($50 \mu\text{m}$). In this regard, it is essential to note that the norm of the magnetic field is not the direct equivalent of the APD; rather, it is related to the time derivative of the action potential. Indeed, for electrical signals with a constant conduction velocity (v_c), the following equality between the spatial and temporal derivatives holds

$$\frac{\partial V_m(t, z')}{\partial t} = -v_c \frac{\partial V_m(t, z')}{\partial z'}. \quad (25)$$

In particular, the maximal value of the magnetic field norm is directly proportional to the maximal value of the time derivative of the action potential, which occurs around the upstroke, as detailed in [54] and discussed in the results.

Cardiac alternans have been analyzed using restitution curves, which depict the local values of APD_{80} or $|\mathbf{B}|_{\max}$ as a function of PCL. Unlike our previous work [54], we chose to plot pointwise restitution curves to preserve the intrinsic variability of the data, as spatial averaging, particularly in the presence of SDA along the wire, could artificially attenuate the true extent of alternans dynamics.

Spatially discordant alternans were further investigated by analyzing the spatial distribution of APD_{80} and $|\mathbf{B}|_{\max}$ along the wire for the last two beats of each PCL ($n-1$ and n). APD_{80} -spatial maps were constructed by computing APD_{80} at each position along the wire. Similarly, $|\mathbf{B}|_{\max}$ -spatial maps were obtained by averaging the maximal value of the magnetic field norm for each selected beat over the points in the xy plane located at a distance of $50 \pm 10 \mu\text{m}$ from the current source.

III. RESULTS

In this section, we present the main findings of the work. In Sec. III A we analyze the model here refined to reproduce SDA in terms of its ability in reproducing the relation between cardiac action potential and the corresponding magnetic field. Specifically, we describe the spatial and temporal distribution of membrane potential, current density and magnetic field.

In Sec. III B we explore the influence of the thermal state on the action potential amplitude and the magnetic field presenting both the classical APD-PCL restitution curves, and the novel $|\mathbf{B}|$ -PCL curves. Finally, in Sec. III C we demonstrate the magnetic field analysis as a reliable tool for the investigation of spatially discordant alternans.

A. Electric and magnetic description of cardiac APs

We investigate the temperature dependence of the electrophysiological properties of cardiac action potentials and their relationship with the magnetic field. To this end, we focus on the temporal and spatial dynamics of electrical and magnetic activity under normal sinus rhythm conditions (Fig. 1). Specifically, we stimulate the tissue with a pacing cycle length (PCL) of 800 ms, delivering 15 stimuli, and we analyze the last beat of the sequence.

As shown in Fig. 1(a), temperature has a significant influence on the shape of the action potential, particularly affecting its duration, quantified by APD_{80} , and its upstroke, characterized by the maximum rate of voltage change ($\partial V / \partial t_{\max}$). In agreement with experimental data [23,53], at physiological temperature ($T = 37^\circ\text{C}$), the action potential duration is approximately 250 ms, with an amplitude of 100 mV and an upstroke velocity of 228 V/s. As the temperature decreases, the action potential duration progressively increases, nearly doubling at 29°C , while the upstroke velocity slows down to 96 V/s. Conversely, under hyperthermic conditions ($T = 40^\circ\text{C}$), APD decreases below 200 ms whereas the upstroke velocity increases. In contrast to APD and $\partial V / \partial t_{\max}$, the action potential amplitude (APA) remains largely unaffected by temperature variations. Notably, the thermal state of the

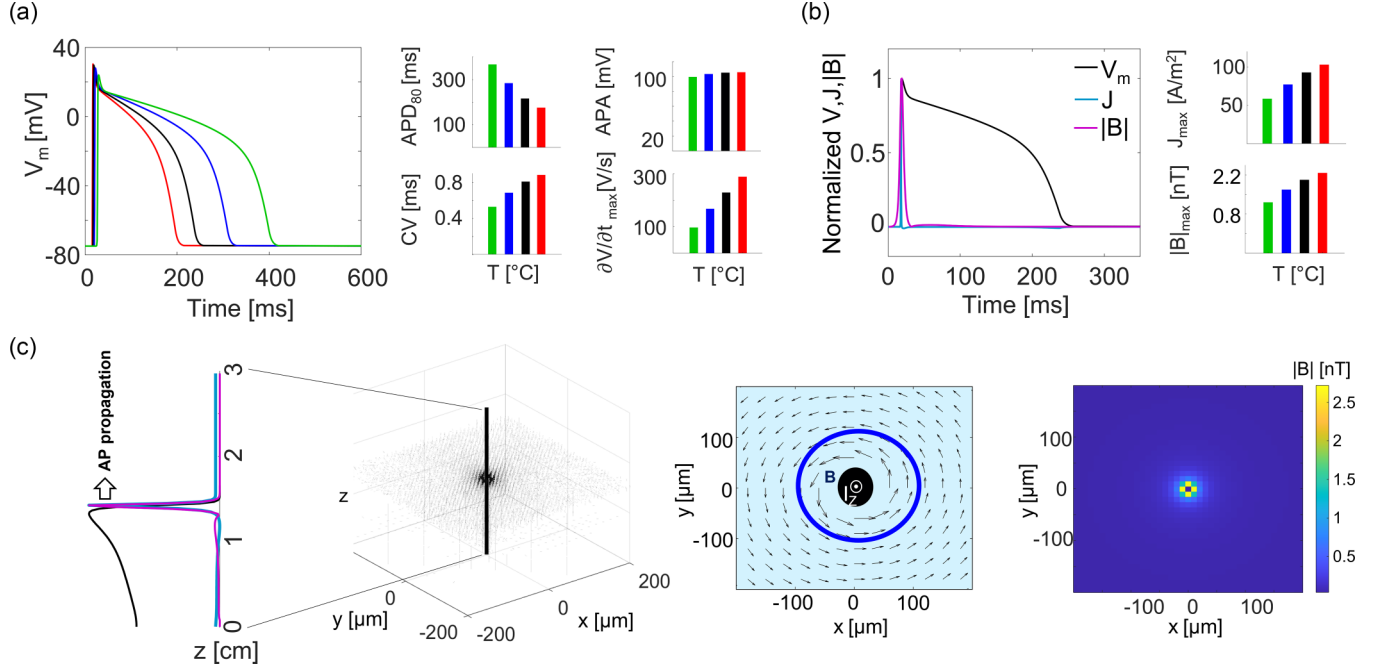


FIG. 1. Simulation configuration. (a) Temperature-induced changes in the cardiac action potential. The panel reports the time course of the cardiac action potential for a fixed location in the center of the wire at the three selected temperatures: 29°C in green, 33°C in black, 37°C in blue, and 40°C in red. The bar plots on the right report the action potential duration, the conduction velocity, the action potential amplitude (APA), and the maximum of the AP time derivative $\partial V / \partial t_{\max}$ for the action potentials shown on the left. (b) Relation between action potential, current density, and magnetic field. In the panel, we show the normalized temporal behavior of the action potential (in black), current density (in violet), and magnetic field (in cyan) at 37°C for a fixed location in the middle of the wire. The current density is computed from the spatial derivative of the AP using Eq. (20) and the magnetic field is obtained from Eq. (24). The reference action potential is already described for panel (a). (c) Spatial configuration of the magnetic field. In this panel, we resume the main spatial features of the magnetic field produced by the cardiac APs. On the left, we show the normalized spatial distribution along the wire of the membrane potential, current density, and of the magnetic field at a fixed time frame. In the middle, we show the vectorial representation of the magnetic field along the wire at a fixed time point. The insert on the right shows the 2D representation of the magnetic field in the xy plane, with the blue ring indicating the distance from the wire at which we computed the norm of the magnetic field. Finally, on the right we report a colormap showing the spatial distribution of the magnetic field intensity in the xy plane. All the plots in panel (c) are obtained at 37°C in the same simulation conditions of panel (a).

tissue strongly impacts conduction velocity, with higher CVs at higher temperatures [Fig. 1(a)].

Among the classical parameters used to characterize the action potential, the upstroke velocity plays a critical role in explaining the relationship between cardiac electrophysiology and the magnetic field generated by electrical activity. For electrical signals propagating at a constant velocity, the magnetic field is directly linked to the temporal derivative of the action potential, as described by Eqs. (20) to (25). Specifically, the current density, J , depends on the spatial derivative of the action potential, Eq. (20), which, in turn, is related to its temporal derivative, as outlined in Eq. (25). As illustrated in Fig. 1(b), the current density exhibits a sharp peak during the upstroke, where the temporal derivative of the action potential reaches its maximum. In contrast, during the repolarization phase, the current density rapidly decreases to zero, displaying a small negative peak toward the end of the action potential. As expected, the current density follows the same temperature dependence as $\partial V / \partial t_{\max}$, i.e., increasing at higher temperatures.

According to these observations, the total current density responsible for generating the magnetic field increases with temperature increment. As a direct result, the magnetic field

intensity also increases, reaching a peak value of approximately 1.8 nT at 37°C at a distance of 50 μm from the source. Similar to the current density, the magnetic field reaches its maximum during the upstroke and subsequently declines to zero throughout the plateau and repolarization phases.

As the final step of this preliminary analysis, we examined the spatial distribution of the membrane potential, current density, and magnetic field [Fig. 1(c)]. The current density and the corresponding magnetic field reach their maximum values on the wavefront (depolarization phase). Due to the intrinsic symmetry of the problem, the magnetic field is confined to a narrow region along the z -axis, where its peak is localized. As expected, the magnetic field streamlines form concentric circles centered in the wire and oriented according to the direction of the current. Moreover, in agreement with the laws of magnetostatics, the magnetic field intensity decreases with increasing distance from the wire [Fig. 1(c)].

B. Electric and magnetic alternans

After investigating the relation between the action potential and the magnetic field under normal rhythm conditions, we analyze the electrical and magnetic activity

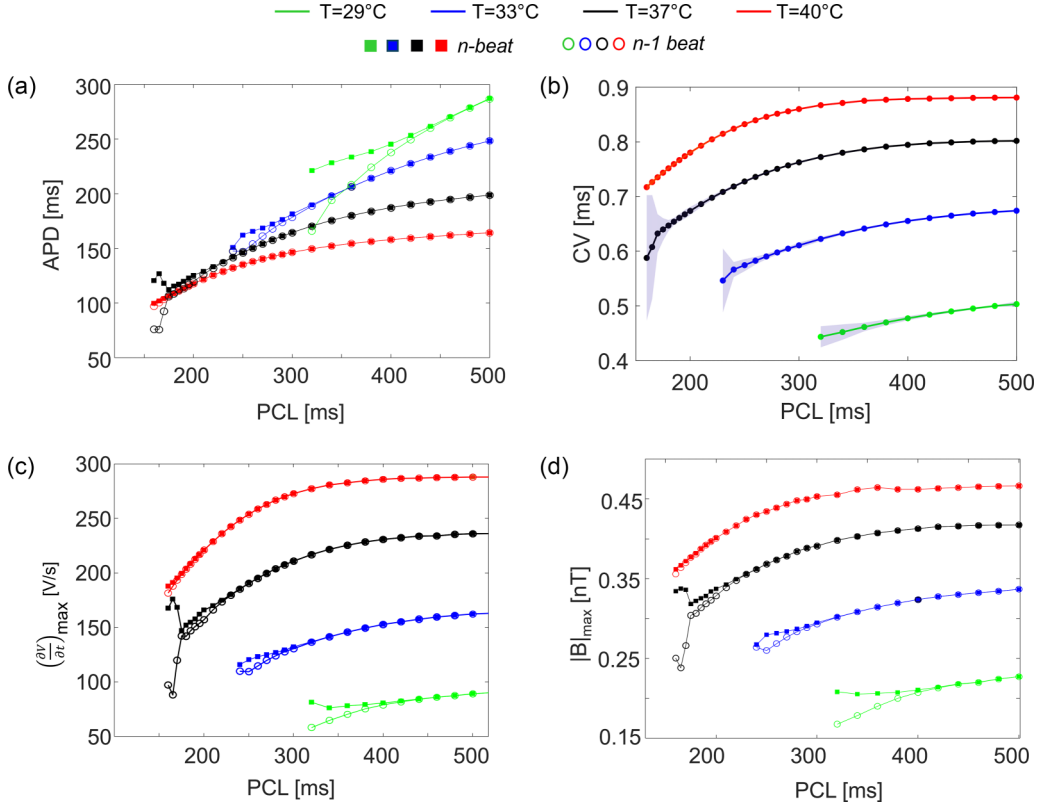


FIG. 2. Action potential and magnetic field. (a) APD_{80} restitution curves. This panel presents classical restitution curves, illustrating the variations in action potential duration as a function of pacing cycle length (PCL) and tissue temperature. The curves are color-coded as reported in the legend. (b) Conduction velocity restitution curves. This panel displays the average conduction velocity (CV) computed for each temperature and PCL. The shaded regions represent the variation in CV between beats n and $n-1$. The average CV is obtained by averaging the conduction velocities of the n and $n-1$ beats. (c) Upstroke velocity restitution curves. This panel reports the maximum value of the time derivative of the action potential ($\partial V / \partial t_{\max}$) for each PCL in the restitution protocol, evaluated for beats n and $n-1$. (d) $|B|_{\max}$ restitution curves. The magnetic-field norm restitution curves follow the same color scheme as panel (a) to indicate different temperatures.

of the tissue in the alternans regime. To characterize alternans dynamics, we report four key indicators: APD_{80} , maximum upstroke velocity ($\partial V / \partial t_{\max}$), conduction velocity (CV), and the peak value of the magnetic-field norm.

Action potential alternans were investigated by stimulating the sample using a full pacing-down restitution protocol, as described in Sec. II. In agreement with previous studies [11,53], lower temperatures promote the onset and progression of cardiac alternans [Fig. 2(a)]. APD restitution curves indicate that at 29°C, alternans emerge for PCLs below 340 ms, whereas at physiological temperature (37°C), the onset of alternans is significantly shifted to shorter PCLs ($PCL_{\text{onset}} = 175$ ms). At the highest temperature analyzed (40°C), no alternans are detectable in the APD , even at low PCLs.

The other electrophysiological parameters considered, namely, the maximum upstroke velocity ($\partial V / \partial t_{\max}$) and the conduction velocity, exhibit the same behavior as APD during the pacing-down restitution protocol, as shown in Figs. 2(b) to 2(c). At this stage, we do not report the action potential amplitude as a function of PCL since, as discussed, this parameter is not significantly affected by temperature variations. Also, APA alternans are rarely arising in the absence of APD alternans, and can be partially isolated from APD alternans, only

under particular experimental conditions masking variations in the APD [63].

We characterize the alternans regime from the perspective of the magnetic field by constructing the magnetic restitution curves shown in Fig. 2(d). These curves illustrate the relationship between the maximal value of the magnetic field at a distance of 50 μm from the source and the corresponding PCL. Our results suggest that AP alternans can be detected from the magnetic restitution curves, with the onset of alternans being more easily identifiable in these curves than in those of APD . Notably, the magnetic restitution curves appear in reverse order compared to the corresponding APD curves, while they closely follow the behavior of $\partial V / \partial t_{\max}$ [Figs. 2(b), 2(c), and 2(d)], as expected from Eqs. (20) to (24). Regarding the relation between the magnetic fields alternans and changes in the conduction velocity, we observe that magnetic restitution curves generally reflect the behavior of the conduction velocity. However, it is worth noting that a direct relation cannot be established between the conduction velocity and the magnetic field, as in the case of the $\partial V / \partial t_{\max}$, especially at low temperatures. At 29°C the restitution of the CV is nearly flat while action potential alternans are arising for $PCL = 550$ ms. This behavior was also observed at 27°, where APD alternans are arising with

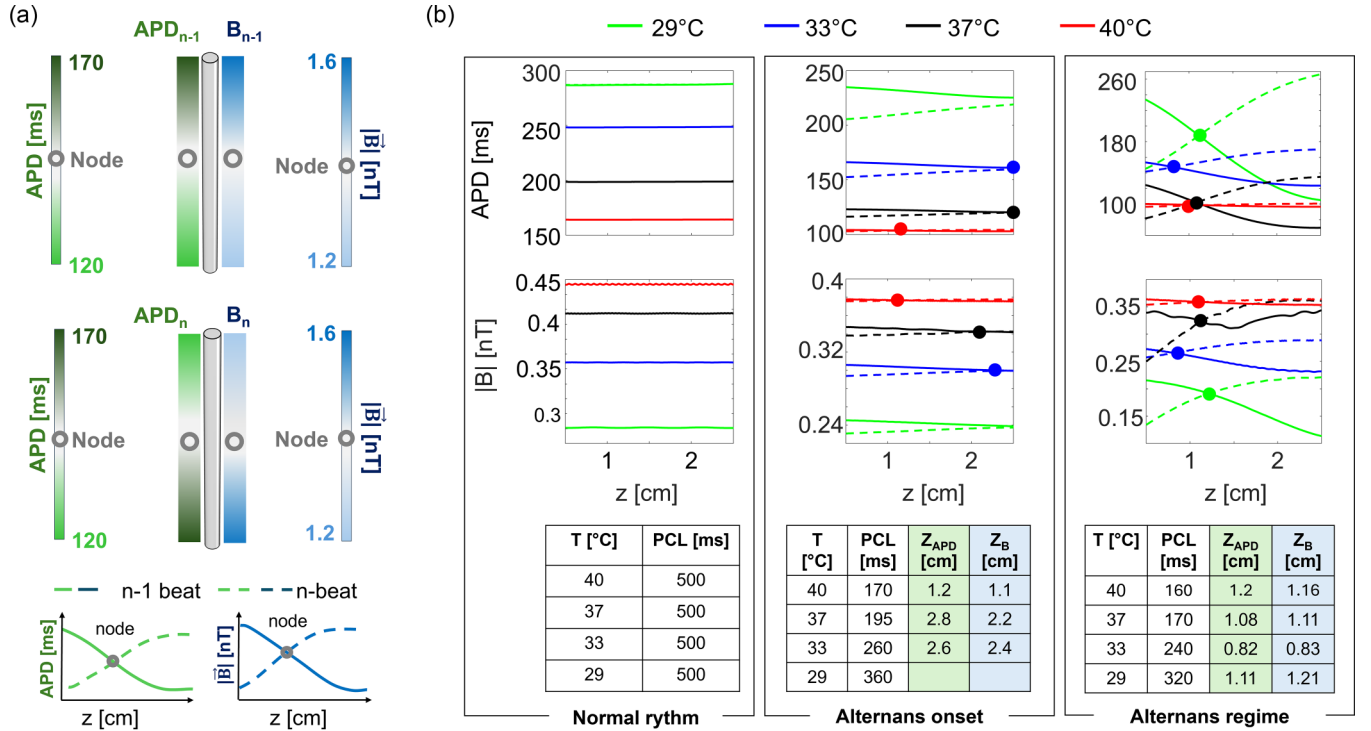


FIG. 3. Spatially discordant alternans. (a) APD and $|B|_{\max}$ SDA. Representation of the spatial alternans map for Action Potential Duration (APD) and the magnetic field norm $|B|_{\max}$ along the wire for two consecutive beats ($n-1$ and n). (b) Temperature dependence of SDA. The spatial distributions of APD and $|B|_{\max}$ are illustrated for each temperature across three distinct electrophysiological regimes: sinus rhythm, alternans onset, and fully developed alternans. Pacing cycle lengths (PCLs) and the corresponding spatial locations of the node for both measured parameters (Z_{APD} and Z_B) are summarized in the tables at the bottom. Circles denote spatial nodes where the electrical and magnetic properties assume equal values between the selected beats. Empty cells denote the absence of the node.

almost flat CV. The simulations at 27°C are not shown in the paper because such a low temperature severely compromises the functionality and vitality of the tissue, and therefore, it is not representative of common physiological and experimental configurations.

C. Electric and magnetic SDA

Spatially discordant alternans (SDA) are characterized by different regions of the tissue exhibiting out-of-phase action potentials, leading to dispersion of repolarization and enhanced tissue heterogeneity [4,54]. As the final step of our work, we analyze the spatial behavior of alternans (Fig. 3) to investigate whether temperature-dependent SDA arise in a simple fiber geometry and whether they can be detected via magnetic-field measurements. To this end, we compute the APD and the magnetic-field norm along the length of the wire at three selected PCLs, representative of normal rhythm, alternans onset, and the fully developed alternans regime. Given the temperature-dependent variability in alternans onset, the PCLs corresponding to alternans onset and the alternans regime are specifically selected for each temperature, see Fig. 3.

As expected, under normal rhythm conditions, the APD and the magnetic-field norm assume the same value for beats n and $n-1$ at each position along the wire. At alternans onset, SDA emerge at 33°C , 37°C , and 29°C , while for 40°C ,

all regions of the domain do not show measurable alternans [Fig. 3(b)].

Notably, SDA are also clearly detectable by analyzing the magnetic-field norm as a function of space. The magnetic-field curves closely mirror the APD ones, exhibiting spatially discordant alternans at 33°C and 37°C [Fig. 3(b)].

The space-dependent positions of the nodes in spatially discordant alternans, denoted as Z_{APD} and Z_B , are summarized in Fig. 3 for different temperatures and PCLs. At 37°C and 33°C , corresponding to PCLs of 170 ms and 240 ms, respectively, the node positions for APD and the magnetic-field norm value show close agreement, with deviations of only a few millimeters ($Z_{\text{APD}} = 1.08\text{ cm}$, $Z_B = 1.11\text{ cm}$ at 37°C ; $Z_{\text{APD}} = 0.82\text{ cm}$, $Z_B = 0.83\text{ cm}$ at 33°C). This strong spatial correlation between electrical and magnetic indicators highlights the capability of magnetic-field measurements to reliably detect SDA. Similarly, at lower temperatures (29°C) and higher PCL (320 ms), the nodes shift further along the wire ($Z_{\text{APD}} = 1.11\text{ cm}$, $Z_B = 1.21\text{ cm}$). This displacement indicates a temperature-dependent variation in the dynamics of SDA, as lower temperatures are associated with a broader spatial distribution of alternans.

These results demonstrate the sensitivity of both electrical and magnetic indicators to spatial alternans patterns, emphasizing the benefit of magnetic-field measurements for noninvasive detection of SDA in cardiac tissue.

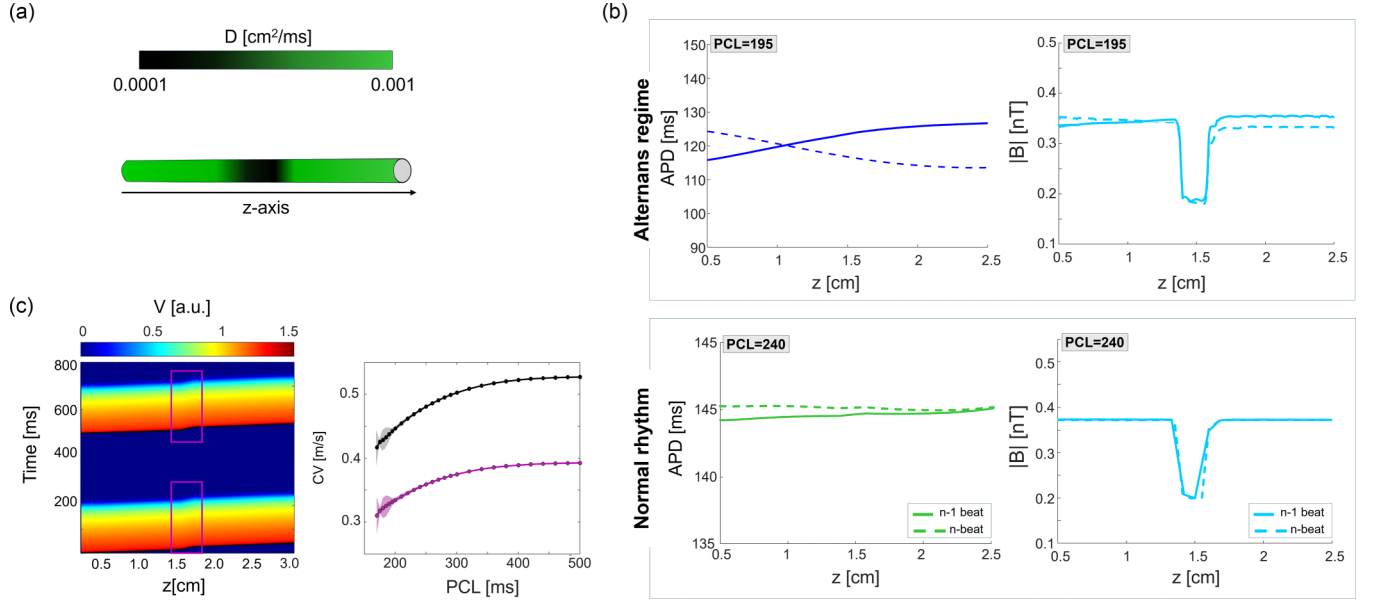


FIG. 4. Spatially discordant alternans in heterogeneous cables. (a) Graphical representation of the heterogeneity in the diffusion coefficient. We introduce at the center of the cable a 0.2-cm-long region where the diffusion coefficient is reduced by 70%. APD and $|B|_{\max}$ alternans in heterogeneous cables. In the panels we report the APD and $|B|_{\max}$ values along the cable in alternans regime (top row) and normal rhythm (bottom row). (c) Spatiotemporal map of AP propagation along the cable. The panel shows the propagation of two beats along the cable. The simulations have been performed at constant temperature ($T = 37^\circ\text{C}$) and the magnetic field is computed at $50\ \mu\text{m}$ from the wire.

D. SDA in heterogeneous cables

As the last step of the work, we study the effect of structural heterogeneity on SDA onset and on the corresponding magnetic field. The structural heterogeneity is introduced reducing the diffusion coefficient by 70% in the central region of the cable [Fig. 4(a)] according to Eqs. (18) and (19). In this way, we mimic regions of postinfarct scarring or fibrosis in tissue [58,59].

The reduction of diffusivity affects the conduction velocity, which is significantly reduced in the region with low diffusivity, while presenting normal values outside [Fig. 4(c)]. Interestingly, in the magnetic-field representation the region with reduced diffusivity is clearly highlighted by the decrease of magnetic-field intensity [Fig. 4(b)], which is not reflected by the corresponding APD analysis. The capability of the magnetic field to reveal structural heterogeneities in the tissue is further confirmed with the analysis in normal rhythm conditions. Indeed, even in normal rhythm condition, the APD analysis does not effectively capture the structural heterogeneity [Fig. 4(b)].

IV. DISCUSSION AND CONCLUSION

The presented work aimed to investigate how spatially discordant alternans, a well-known arrhythmogenic mechanism, modulates the magnetic field generated by cardiac activation. To do this, we developed a controlled numerical framework based on the monodomain formalism and used it to simulate the evolution of both the transmembrane voltage and the associated magnetic field in an idealized one-dimensional fiber. This simplified configuration enabled precise control of key model parameters, e.g., conduction velocity, tissue

conductivity, action potential duration, upstroke rise velocity, and allowed us to isolate the individual contributions of such parameters to the magnetic signature.

A key aspect of the model is the incorporation of temperature dependence, which is well known to modulate ion kinetics and affects both APD and CV-restitution curves entailing an increased risk of arrhythmias at low temperature.

In particular, we observed that SDA and its associated magnetic-field alternans patterns persist even at reduced temperatures (e.g., 29°C), where the CV restitution becomes nearly flat (Fig. 2). This confirms that the insorgence of magnetic alternans does not rely only on restitution steepness, but rather reflects a more general spatiotemporal interplay between wavefront propagation and repolarization heterogeneity.

We also observed a B-intensity marked thermal dependence, that decreases as the temperature lowers and consistent with the temperature-dependence of the upstroke velocity, suggesting that the norm of the magnetic field might be used as a reliable indicator for investigating cardiac action potential features, in particular for a fine characterization of the AP-upstroke phase.

The integrative relation between the magnetic field and the action potential may amplify small alterations of the upstroke which might not be easily detectable with standard electrophysiological measurements, offering new interesting opportunities for gaining deeper insights on the molecular mechanisms of cardiac channelopathies involving mutations of the sodium channels, such as Brugada Syndrome, Isolated Progressive Cardiac Conduction Disease (PCCD), or Long-QT 3 syndrome [64]. Instead, the strict relation between the time derivative of the action potential and the magnetic field suggests that magnetic measurements could be less informative

for the investigation of the plateau and repolarization phases of the action potential because current density, and therefore, the magnetic field, is almost null.

In addition, our results show that the magnetic field is not a direct mirror of APD but encodes its spatiotemporal dynamics, even in the case of different thermal baths. Our results suggest that magnetic-based cardiac alternans can be detected at all temperatures, showing an inverted behavior with respect to the classical APD restitution curve.

Though differences between n and $n - 1$ beats are not significantly amplified in the magnetic-field curves (Fig. 3), the bifurcation point is more easily detectable than in the APD restitution curves, suggesting that magnetic-field measurements could be a valuable alternative for the identification of arrhythmias precursors.

In addition to functional heterogeneity induced by SDA, we explored the effects of structural heterogeneity by introducing a spatially localized reduction in tissue conductivity. This perturbation, intended to mimic regions of postinfarct scarring or fibrosis [58,59], significantly altered the spatial distribution of the magnetic field. Notably, we observed a pronounced reduction in the B-intensity localized at the borders of the low conductivity zone, with no appreciable changes in the APD distribution. This highlights a key distinction: while voltage-based measures primarily capture local electrophysiological dynamics, the magnetic field integrates spatial information and is particularly sensitive to gradients in current density and conduction velocity. Clinical studies on magneto-cardiography support this concept, reporting high sensitivity to coronary-artery occlusion and ischaemic cardiomyopathy [33–37].

A critical aspect regarding the magnetic-field interpretation lies in its nonlocal nature, as it reflects the integrated effect of the transmembrane currents across the tissue. Previous theoretical and computational studies demonstrated that the dominant contribution to the cardiac magnetic field arises from the vicinity of the depolarization wavefront, where transmembrane potential gradients are steepest and current densities are highest [Fig. 1(b)]. In contrast, regions that are fully depolarized or still at rest contribute minimally due to the negligible current density flow. Additionally, contributions from distant regions tend to cancel out because of their opposing orientations and the vector nature of the magnetic field. These effects, combined with the spatial decay of the magnetic field, result in a measured signal that is effectively dominated by local sources near the wavefront, despite the theoretical nonlocality of the B-field.

Another important aspect deserving further discussion regards the intensity of the magnetic field, in particular in relation to the current measuring techniques. Our simulations predict magnetic-field intensities in the nT at a distance of 50 μm from the wire, which could be detected with the state of the art of magnetic field sensors, such as NV centres, SQUIDs or OPM [41–43,45–48,51,55]. While comparing the results with available literature it is important to consider that

the recorded intensities as well as the spatial resolution are strongly influenced by stand-off distance of the sample from the sensor. Overall, our results are in agreement with available experimental data reporting the magnetic fields of excitable cells and tissues [41,43,45,48,51,55].

Before concluding, it is worth highlighting some critical aspects of the present work. The first and the main limiting factor is the simplicity of the selected geometry, a 1D cable [65], which intrinsically restricts the variety of the spatiotemporal phenomena observed. While this configuration does not capture the full complexity of the cardiac anatomy, the qualitative features we observed (as the alternans node and amplitude attenuation at low-conductivity regions) are expected to persist in higher-dimensional settings. Although the transition from one-dimensional to higher-dimensional models will introduce several challenges (as wavefront complex geometries, curvature, and interaction patterns) future work will extend the present framework to two- and three-dimensional geometries. These lead to more intricate current distributions and, consequently, to spatially integrated magnetic fields that are harder to interpret. Additionally, tissue heterogeneities, fiber orientation, and boundary conditions can all affect the current pathways, potentially altering both the magnitude and topology of the resulting magnetic fields [66–69]. We expect that these aspects have to be carefully addressed through computational optimization and advanced signal analysis.

In conclusion, despite the discussed limitations, this work provides a clear understanding of how SDA and structural heterogeneities shape the cardiac magnetic field, offering additional perspectives for future investigation of complex spatiotemporal dynamics occurring in arrhythmogenic regimes. In future work, we aim to incorporate realistic geometries, material tissue properties, and inverse algorithms [70,71] to explore the full diagnostic potential of magnetic-field analysis in cardiac electrophysiology.

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DATA AVAILABILITY

The data are not publicly available. The data are available from the authors upon reasonable request.

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