Nonequilibrium Arrhythmic States and Transitions in a Mathematical Model for Diffuse Fibrosis in Human Cardiac Tissue

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Abstract

We present a comprehensive numerical study of spiral-and scroll-wave dynamics in a state-of-the-art mathematical model for human ventricular tissue with fiber rotation, transmural heterogeneity, myocytes, and fibroblasts. Our mathematical model introduces fibroblasts randomly, to mimic diffuse fibrosis, in the ten Tusscher-Noble-Noble-Panfilov (TNNP) model for human ventricular tissue; the passive fibroblasts in our model do not exhibit an action potential in the absence of coupling with myocytes; and we allow for a coupling between nearby myocytes and fibroblasts. Our study of a single myocyte-fibroblast (MF) composite, with a single myocyte coupled to Nf fibroblasts via a gap-junctional conductance Gj,m, reveals five qualitatively different responses for this composite. Our investigations of two-dimensional domains with a random distribution of fibroblasts in a myocyte background reveal that, as the percentage Pf of fibroblasts increases, the conduction velocity of a plane wave decreases until there is conduction failure. If we consider spiral-wave dynamics in such a medium we find, in two dimensions, a variety of nonequilibrium states, temporally periodic, quasiperiodic, chaotic, and quiescent, and an intricate sequence of transitions between them; we also study the analogous sequence of transitions for three-dimensional scroll waves in a three-dimensional version of our mathematical model that includes both fiber rotation and transmural heterogeneity. We thus elucidate random-fibrosis-induced nonequilibrium transitions, which lead to conduction block for spiral waves in two dimensions and scroll waves in three dimensions. We explore possible experimental implications of our mathematical and numerical studies for plane-, spiral-, and scroll-wave dynamics in cardiac tissue with fibrosis.

Introduction

Extra-cellular-matrix (ECM) materials constitute about 6% of the volume of human cardiac tissue in an average, healthy heart [1]. These include fibroblasts, non-excitable collagen, and elastin fibrils, which fill the subepicardial space between the epicardium and myocardium [2] and bridge the gaps between myocardial tissue layers. The major component of the ECM are fibroblast cells that produce interstitial collagen, of types I, III, IV, and VI [3]. These contribute to myocardial structure, cardiac development, cell-signaling, and electro-mechanical functions in myocardial tissue. In mammalian cardiac tissue, fibroblast cells show an intimate spatial interrelation with every myocyte that borders one or more fibroblasts [4]. In tissue containing both myocytes and fibroblasts, it has been assumed traditionally that gap-junctional couplings exist exclusively between myocytes; but recent experimental studies have shown that there is a functional, heterogeneous, myocyte-fibroblast coupling [3,4].

Computer simulations of electrical-wave propagation in mathematical models for cardiac tissue have been used to investigate the interplay of spiral and scroll waves with conduction and other inhomogeneities [1,5–21]. Some of these studies [5,6,14–18] concentrate on the interaction of a spiral or scroll wave with a localized inhomogeneity; others are devoted to investigations of the effects of a large number of randomly-distributed, point-type inexorable obstacles on such waves [1,7–13]. We concentrate on the latter types of studies here because they have been designed to mimic arrays of fibrotic strands or diffuse fibrosis in cardiac tissue. By using a simple model for cardiac tissue with many inexcitable obstacles, Pertsov [7] has shown that such obstacles can influence the rotation of spiral waves and lead to anisotropies in propagation. Turner, et al. [8] have studied the effects of fibrosis in the Priebe-Beukelmann model. Spach, et al. [9] have used the Nygren model for human atrial tissue and mimicked the effects of diffuse fibrosis by removing lateral gap junctions; they find that with such heterogeneity in intercellular couplings, there is a tendency for partial wave block and re-entry. Kuipers, et al. [10] have used the Courtemanche model for human atrial tissue and heterogeneous uncoupling to model diffuse fibrosis. These studies indicate that fibrosis can increase vulnerability to re-entry; however, they have not explored in detail the effects of fibrosis on the dynamics of spiral and scroll waves in these models. Such
an exploration has been initiated by Panfilov [11] and ten Tusscher, et al. [1,12,13], who investigate the effects of diffuse fibrosis on the propagation of electrical waves of activation and arrhythmogenesis in both two-variable and detailed ionic mathematical models for human ventricular tissue; they model fibrosis as non-conducting inhomogeneities that are distributed randomly in their simulation domain. They show that, as the concentration of such inhomogeneities increases, CV decreases for plane-wave propagation, the wave fronts become jagged, and there is an increase in the tendency for the formation and break up of spiral waves; at sufficiently large densities of these inhomogeneities, they find that complete conduction blockage occurs.

McDowell, et al. [22] have used a three-dimensional computational model, based on MRI data, of chronically infarcted rabbit ventricles to characterize arrhythmogenesis because of myofibroblast infiltration as a function of myofibroblast density; this study includes periinfarct zones (PZ), ionic-current remodeling therein, and different degrees of myofibroblast infiltration. Their work shows that, at low densities, myofibroblasts do not alter the propensity for arrhythmia; at intermediate densities, myofibroblasts cause AP shortening and thus increase this propensity; at high densities, these myofibroblasts protect against arrhythmia by causing resting depolarization and blocking propagation.

We present a major extension of the work of ten Tusscher, et al. [1,12,13] on diffuse fibrosis in mathematical models for cardiac tissue by introducing fibroblasts randomly in the state-of-the-art TNNP model [16,23] for human ventricular tissue; the fibroblasts are passive, insofar as they do not exhibit an action potential in the absence of coupling with myocytes. Our model for the fibroblasts is much more realistic than the one used by ten Tusscher, et al. [1,12,13]; in particular, we use the fibroblast model of MacCannell, et al. [24,25]; we also allow coupling between nearby myocytes and fibroblasts. The parameters in this model cannot be determined precisely from experiments [4,25] so it is important to explore a wide, but biophysically relevant, range of parameters. Our in silico study is well suited for such an exploration so it is very effective in complementing experimental studies of electrical-wave propagation in fibrotic cardiac tissue.

We begin with an overview of our principal results before we present the details of our study. We first study a single myocyte-fibroblast (MF) composite in which a single myocyte is coupled to $N_f$ fibroblasts via a gap-junctional conductance $G_{gap}$. We study two cases, namely, moderate and strong coupling between fibroblasts and myocytes; for each one of these cases, we consider three parameter sets [23] for the myocytes that are suitable for epicardial, mid-myocardial, and endocardial layers of the heart wall; experiments suggest that $0.1 \text{nS} \leq G_{gap} \leq 8 \text{nS}$ [26,27]; thus, we consider $G_{gap} = 0.1 \text{nS}$, $G_{gap} = 4 \text{nS}$, and $G_{gap} = 8 \text{nS}$ to be the weak-, moderate- and strong-coupling cases, respectively. We excite each such MF composite via an electrical stimulus and then record its responses for different values of $N_f$ and $E_f$. In the moderate-coupling case ($G_{gap} = 4 \text{nS}$), the electrical load of the fibroblasts on the myocyte is significant, except in a very narrow range of parameters. However, in the strong-coupling case ($G_{gap} = 8 \text{nS}$), for different ranges of the parameters $N_f$ and $E_f$, we observe five qualitatively different responses for the MF composite; we call them R1–R5. In R1 the MF composite responds essentially like an uncoupled myocyte. In régime R2, the MF composite produces a secondary AP, after the first one that is generated by the external stimulus. In R3, this composite displays autorhythmicity, i.e., it fires a train of APs, after the first external stimulus, and without the application of subsequent stimuli; each AP in this autorhythmic train differs from the normal AP of an uncoupled myocyte. In régime R4, the MF composite displays an oscillatory state in which the initial AP response to the external stimulus is followed by oscillations of the membrane potential about a mean value without the application of any other external stimulus. In régime R5, the MF composite produces a single AP under the influence of the external stimulus; after that it does not return to the resting state but to another time-independent state in which it is non-excitable.

We then study propagation of plane waves in a 2D simulation domain with TNNP-type [16,23] myocytes (M) or fibroblasts (F) of the type described in Ref. [24]; M and F are distributed randomly through the simulation domain; and there are diffusive couplings between nearest-neighbor cells. We investigate plane-wave propagation through both mural slices, with epicardial parameters, and transmural slices, consisting of epicardial, mid-myocardial, and endocardial regions, in moderate- and strong-coupling cases, i.e., with myocyte-to-fibroblast diffusion constants of $0.0000218 \text{cm}^2/\text{ms}$ and $0.000048 \text{cm}^2/\text{ms}$, respectively. We obtain stability diagrams for both these cases in the $P_f - E_f$ parameter space. In the moderate-coupling case, this stability diagram is simple: for low values of $P_f$, the plane wave leaves the system, which returns to an excitable state; for large values of $P_f$ the plane wave is annihilated by target waves and the medium is left in a state that is weakly excitable or not excitable at all. In the strong-coupling case the stability diagram is very rich; it contains the spatiotemporal analogs of the régimes R1–R5 mentioned above for an isolated MF composite.

The last part of our study examines the effect of diffuse fibrosis on spiral-wave dynamics in 2D and scroll-wave dynamics in 3D with myocytes and fibroblasts distributed randomly as above; we concentrate on the strong-coupling case here. At low values of $P_f$, we find that single, rotating spiral and scroll waves have slightly corrugated wave fronts, but they propagate much as they do in the absence of fibroblasts. For large values of $P_f$, we find that such spiral and scroll waves do not propagate through the simulation domain and are either (a) annihilated by spontaneously generated target waves or (b) absorbed at the boundaries. This crossover from a state with propagating waves and electrical activity to a state with no electrical activity occurs via a sequence of nonequilibrium transitions; the precise sequence depends on the initial conditions and the realization of the disordered array of M and F cells.

Given the spatial and temporal resolution we have been able to achieve in 2D, we find the following rough sequence of states: at low $P_f$ we begin with a state with a single spiral rotating periodically (SRS); as $P_f$ increases this gives way to a state with a single spiral rotating quasi-periodically (SRSQ); as $P_f$ increases we obtain a state with multiple spirals that rotate periodically (MRSP); this then gives way to a state with a multiple spirals that rotate quasi-periodically (MRSQ), which is followed by a spiral-turbulence (ST) state and, eventually, by the absorption state (SA). In 3D, the analogous sequence of states, which we have been able to resolve, is the following: at low $P_f$ we begin with a state with a single rotating scroll wave (SRS); as $P_f$ increases this gives way to a state with multiple rotating scroll waves (MRS); this then gives way to the absorption state (SA).

Materials and Methods

The first system we study is a single myocyte-fibroblast (MF) composite in which a single myocyte is coupled to $N_f$ fibroblasts via a gap-junctional conductance $G_{gap}$; we consider the range $1 \leq N_f \leq 10$. We then carry out studies in 2D and 3D simulation domains in which myocytes M or fibroblasts F are distributed randomly through the simulation domain; we include diffusive
inward Na current, 
slow, delayed-rectifier current, IKr

Here, INa, the total ionic current, is expressed as a sum of the following six major and six minor ionic currents:

\[ I_{\text{ion}} = I_{\text{Na}} + I_{\text{CaL}} + I_{\text{Kt}} + I_{\text{Ke}} + I_{\text{Kr}} + I_{\text{K1}} \]

\[ I_{\text{Na}} = I_{\text{Na,Ca}} + I_{\text{Na,K}} + I_{\text{Na,Ca}} + I_{\text{Na,K}} + I_{\text{Na,Ca}} + I_{\text{Na,Ca}} \]

\[ I_{\text{CaL}} = \text{fast inward Na}^+ \text{ current}, I_{\text{CaL}} \text{ the L-type slow-inward Ca}^{2+} \text{ current}, I_{\text{Kt}} \text{ the inward rectifier} \]
\[ I_{\text{Kt}} \text{ the inward rectifier} \]
\[ I_{\text{Kt}} \text{ the slow, delayed-rectifier current, } I_{\text{Kt}} \text{ the rapid, delayed-rectifier current, } I_{\text{Kd}} \]

\[ I_{\text{Kd}} \text{ the background Na}^+ \text{ current, } I_{\text{Ca}} \text{ the background Ca}^{2+} \text{ current.} \]

Here, INa is the fast inward Na+ current, ICaL the L-type slow-inward Ca2+ current, IK the inward rectifier K+ current, INaCa the Na+ / Ca2+ exchanger current, INaK the Na+ / K+ pump current, IPCa the plateau Ca2+ current, IPk the plateau K+ currents, INa the background Na+ current, and ICaL the background Ca2+ current. The time \( t \) is measured in milliseconds, voltage \( V \) in millivolts, conductances \( (G_k) \) in nanoSiemens per picofarad (nS/ pF), the intracellular and extracellular ionic concentrations \( (X_i, X_e) \) in millimoles per liter (mM/L) and current densities, per unit capacitance, \( I_C \) in picoamperes per picofarad (pA/pF), as used in second-generation models (see, e.g., Refs. [23,29–31]). For a detailed list of the parameters of this model and the equations that govern the spatiotemporal behaviors of the transmembrane potential and currents, we refer the reader to Refs. [16,23].

For the fibroblasts, we use the model of MacCannell, et al. [24]; i.e., we treat the fibroblasts as passive circuit elements that couple with the myocyte in the MF composite; and the fibroblast ionic current \( I_{\text{ion,fb}} \) is

\[ I_{\text{ion,fb}} = G_f(V_f - E_f) \]

where \( G_f, V_f \) and \( E_f \) are, respectively, the conductance, transmembrane potential, and the resting membrane potential for the fibroblast.

We incorporate muscle-fiber anisotropy in both 2D and 3D simulations, as in Refs. [32,33]; we account for diffusive couplings between nearest-neighbor myocytes, nearest-neighbor fibroblasts, next-nearest-neighbor myocytes, next-nearest-neighbor fibroblasts and nearest-nearest myocyte-fibroblast pairs. We use two diffusion tensors, namely, \( D^{\text{mm}} \) and \( D^{\text{ff}} \), for myocyte-fibrocyte (mm) and fibroblast-fibroblast (ff) diffusive couplings. The diffusion tensors \( D^{\text{mm}} \) and \( D^{\text{ff}} \) have the form used in Refs. [32,34]; we give this form below for a diffusion tensor that is denoted generically by \( D \) and has, in three dimensions, the components shown hereunder:

\[ D = \begin{bmatrix} D_{11} & D_{12} & 0 \\ D_{21} & D_{22} & 0 \\ 0 & 0 & D_{zz} \end{bmatrix} \]

where

\[ D_{11} = D_1 \cos^2 \theta(z) + D_{\perp 1} \sin^2 \theta(z), \]
\[ D_{22} = D_1 \sin^2 \theta(z) + D_{\perp 2} \cos^2 \theta(z), \]
\[ D_{12} = D_2 (D_1 - D_{11}) \cos \theta(z) \sin \theta(z). \]
components $D_{11}, D_{12}, D_{21},$ and $D_{22}$ for a mural slice in the $x$–$y$ plane; for a transmural slice in the $x$–$z$ plane we retain only the components $D_{11}$ and $D_{12}.$

We turn now to $mf$ and $fm$ diffusive couplings; the magnitudes of these are not known well experimentally, nor has the role of fiber orientation been investigated in this context. In the following paragraphs, in the interests of a parsimonious description, we neglect fiber orientation in the $mf$ and $fm$ diffusive couplings $D_{mf}$ and $D_{fm},$ respectively, and treat them as scalars. With this idea the interaction between a myocyte and a fibroblast should be weaker than that between two myocytes, but stronger than that between two fibroblasts, we use the following illustrative values: (a) for the strong-coupling case $D_{mf} = 0.00141 \text{cm}^2/\text{ms}$ and $D_{fm} = 0.000048 \text{cm}^2/\text{ms}$; and (b) for the moderate-coupling case $D_{mf} = 0.000642 \text{cm}^2/\text{ms}$ and $D_{fm} = 0.0000218 \text{cm}^2/\text{ms};$ in both these cases $D_{mn} = D_{nm} (C_m/C_f),$ where the total cellular capacitances for myocytes and fibroblasts are $C_m = 185 \text{ pF}$ and $C_f = 6.3 \text{ pF},$ respectively [24]. Note that, in our 2D and 3D models, there is no on-site coupling between myocytes and fibroblasts; this has been translated into diffusive couplings between such cells if they are at nearest-neighbor sites in our simulation domains.

We generate the initial condition for our studies by using the following protocols: We begin with only myocytes on all sites of our 2D simulation domain. For plane-wave-propagation studies, we apply a stimulus, of amplitude $150 \text{pA}/\text{pF}$ for $2 \text{ ms},$ along one edge of the simulation domain. For our studies of spiral-wave dynamics, we obtain a spiral wave in the 2D domain by using the method proposed by Shajahan, et al. [16]. In our 3D scroll-wave studies, we begin with an initial scroll wave that consists of our initial, 2D spiral waves stacked one on top of the other; thus, we begin with a simple scroll wave with a straight filament as in Ref. [17]. Pseudocolor plots of $V_m$ for these spiral and scroll waves, which we use as initial conditions in our subsequent studies, are given in Figs. 1 (a) and (b), respectively.

For every value of $P_f,$ we generate a random array of myocytes and fibroblasts in our 2D and 3D simulation domains as follows by using a random-number generator to assign F or M to a site such that the percentage of F sites is $P_f.$ Illustrative arrays of F and M are given in Fig. 2 (a)–(c). This distribution of F and M cells is held fixed throughout the subsequent spatiotemporal evolution of the initial spiral and scroll waves described above; in the language of condensed-matter physics, a static configuration of F and M is an example of quenched disorder [35–37]. At the initial time, the fibroblast transmembrane potential $V_f$ is set equal to its resting value $E_f = -49.6 \text{ mV}$ [24].

The temporal evolution of the transmembrane potential $V_A$ of the cell at site $A$ in the lattice is governed by

$$\frac{\partial V_A}{\partial t} = -\frac{I_{ion,A}}{C_d} + \mathcal{D}_A,$$

where $\mathcal{D}_A$ indicates the diffusion term. This can be written most easily in discrete form and it depends on (a) whether the cell at site $A$ is M or F and (b) whether the cell on the neighboring site is of type M or F. We illustrate the form of the diffusion term $\mathcal{D}$ for a two-dimensional mural slice for three representative sites $A,$ $B,$ and $C$ in Fig. 2(b), which have M, M, and F cells, respectively:

$$\mathcal{D}_A = D_{mf} (\frac{V_A(t+1,j) - V_A(t,j) - 2V_f)}{\Delta x^2} + D_{mf} (\frac{V_A(t+1,j) - V_A(t,j+1) - 2V_f)}{\Delta y^2}$$

$$+ D_{mf} (\frac{V_f(t) - V_f(t,j-1)}{\Delta y^2}) + D_{mf} (\frac{V_f(t+1,j) - V_f(t,j+1) - 2V_f)}{\Delta y^2} + D_{mf} (\frac{V_f(t,j) - V_f(t,j-1)}{\Delta x^2})$$

$$+ D_{mf} (\frac{V_f(t) - V_f(t,j-1)}{\Delta y^2}) + D_{mf} (\frac{V_f(t+1,j) - V_f(t,j+1) - 2V_f)}{\Delta y^2} + D_{mf} (\frac{V_f(t,j) - V_f(t,j-1)}{\Delta x^2}).$$

In our studies with the MF composite, we apply a stimulus current of $52 \text{pA}/\text{pF}$ for $1 \text{ ms}$ to the composite and allow the system to evolve in time. We record the membrane potential of the central myocyte in the MF composite and plot it as a function of time for different values of $N_f$ and $E_f.$

For our measurements of $CV$ and $\lambda,$ we prepare the 2D simulation domain as discussed above and initiate a plane wave by applying a stimulus of amplitude $150 \text{pA}/\text{pF}$ for $2 \text{ ms}$ along the left edge ($j$ axis) of the domain. We record the time series of $V_m$ at four representative points of the domain. For studies on the mural slice, these points are $(3.375 \text{cm}, 3.375 \text{cm}),$ $(10.125 \text{cm}, 3.375 \text{cm}),$ $(3.375 \text{cm}, 10.125 \text{cm}),$ and $(6.75 \text{cm}, 6.75 \text{cm});$ for studies on the transmural slice, these points are $(3.375 \text{cm}, 0.3375 \text{cm}),$ $(3.375 \text{cm}, 1.0125 \text{cm}),$ $(10.125 \text{cm}, 0.3375 \text{cm}),$ $(10.125 \text{cm}, 1.0125 \text{cm})$ and $(6.75 \text{cm}, 0.675 \text{cm}).$ From these time-series data, we obtain the times $t_1$ and $t_2$ at which the upstroke of the action potential ($AP$) is initiated at pairs of sites that are separated by $\Delta x$ along the axis parallel to the direction of propagation of the wave; $CV = \Delta x/\Delta t,$ where $\Delta t = t_2 - t_1; the$ wavelength $\lambda = CV \cdot APD_{90\%};$ where $APD_{90\%}$ is the action-potential duration at $90\%$ repolarization; we obtain average values for $CV$ and $\lambda$ over the four points mentioned above.

Results

In this Section we present the results of our computational studies. We begin with our investigation of MF composites and discuss how their action potential is influenced by the number of fibroblasts $N_f,$ their resting membrane potential $E_f,$ the gap-junctional coupling $G_{gap},$ and the myocyte parameters, which distinguish myocytes from the epicardium, the mid-myocardium, and the endocardium. We then explore the propagation of plane waves of electrical activation through 2D simulation domains with randomly distributed myocytes and fibroblasts such that the percentage of fibroblasts is $P_f.$ We consider propagation through

Figure 1. The initial configurations for the spiral and scroll waves in our 2D and 3D simulations (see text).
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both mural and transmural slices. Next we consider spiral-wave dynamics in 2D and 3D simulation domains with \( P_f \) fibroblasts. The action potential durations (APDs) are different, for uncoupled myocytes from the endocardium, the mid-myocardium and the epicardium. The APD of the myocyte-fibroblast composite (MF) is also different for the three types of myocytes. However, the APD of an MF depends not only on the type of myocyte but also on the values of the gap-junctional conductance \( G_{\text{gap}} \), the resting membrane potential of fibroblasts \( E_f \), and the number of fibroblasts coupled to a myocyte \( N_f \). Fig. 3 shows pseudocolor plots of the APD for MFs, with epicardial, mid-myocardial, and endocardial myocytes, as functions of \( G_{\text{gap}} \) and \( E_f \), for \( N_f = 1, 2, 3, \) and 4. In our studies, \( G_{\text{gap}} \) is moderate (4 nS) or strong (8 nS), so the influence of the gap-junctional coupling is quite significant.

When \( N_f = 1 \), the differences between the APDs, for the three types of MFs, is considerable, for all values of \( G_{\text{gap}} \) and \( E_f \); as \( N_f \) increases, this difference between the APDs is significant only at low values of \( G_{\text{gap}} \); in particular, for \( N_f = 4 \) and \( G_{\text{gap}} = 0 \) nS, the distinction between these APDs is almost negligible. However, if \( N_f = 4 \) and \( G_{\text{gap}} < 2 \), this distinction is quite prominent at all the values of \( E_f \) that we have considered. For studies on transmural heterogeneity in mouse tissue, please refer to [38,39].

**MF Composite**

The response of the myocyte-fibroblast (MF) composite depends on \( N_f, E_f \), and \( G_{\text{gap}} \) and the properties of the myocyte. In both moderate- and strong-coupling cases, with \( G_{\text{gap}} = 4 \) nS and \( G_{\text{gap}} = 8 \) nS, respectively, if \( N_f = 1 \) the MF composite produces a single action potential on the application of an external stimulus and then returns to the normal resting membrane potential for myocytes \( (\approx -86 \text{mV}) \); we designate this as a response of type \( \text{R1} \); and we illustrate this, for the case \( G_{\text{gap}} = 8 \) nS, by plots of \( V_m \) versus time \( t \) in Figs. 4 (a.1), (a.2), and (a.3) for epicardial, mid-myocardial, and endocardial myocytes, respectively. Four other types of responses are possible and are listed below and portrayed in Fig. 4, for the case \( G_{\text{gap}} = 8 \) nS: \( \text{R2} \): In this case there is a secondary AP, after the first one generated by an external stimulus; the MF composite then returns to the resting state as in \( \text{R1} \) (Figs. 4 (b.1), (b.2), and (b.3) for epicardial, mid-myocardial, and endocardial myocytes, respectively). \( \text{R3} \): Here the MF composite is autorhythmic, i.e., it produces a sequence of APs, after the first external stimulus; each AP in this autorhythmic sequence differs from the normal AP of an uncoupled myocyte (Figs. 4 (c.1), (c.2), and (c.3) for epicardial, mid-myocardial, and endocardial myocytes, respectively). \( \text{R4} \): The MF composite can display an oscillatory response in which the initial, stimulus-induced AP is

![Figure 2. Spatial distributions of myocytes and fibroblasts in our simulation domain.](image-url)
followed by oscillations of $V_m$, about a mean value, without the application of any other external stimulus (Figs. 4 (d.1), (d.2), and (d.3) for epicardial, mid-myocardial, and endocardial myocytes, respectively). \textbf{R5}: The MF composite produces a single AP because of an external stimulus; after that it does not return to the normal resting state but to another time-independent state in which it is non-excitable (Figs. 4 (e.1), (e.2), and (e.3) for epicardial, mid-myocardial, and endocardial myocytes, respectively).

The regions in which our MF composite displays responses of types R1–R5 are shown, for illustrative parameter values, in the $E_f - N_f$ plane in Figs. 5 (a.1), (a.2), (b.1), (b.2), (c.1) and (c.2). For the moderate-coupling case, $G_{gap} = 4$ nS, Figs. 5 (a.1), (b.1), and (c.1) show the stability diagrams for, respectively, endocardial, mid-myocardial, and epicardial myocytes in the MF composite; their analogs for the strong-coupling case, $G_{gap} = 8$ nS, are given in Figs. 5 (a.2), (b.2), and (c.2); here régimes R1, R2, R3, R4, and R5 are denoted, respectively, by yellow hexagrams, red squares, green circles, pink diamonds, and blue pentagrams. All these régimes appear in stability diagrams for the moderate- and strong-coupling cases; but régimes R3 and R4 occupy very small areas especially in the moderate-coupling case; and régime R5, which occupies a significant fraction of the stability diagram in the strong-coupling cases, occurs in a narrow parameter range in the case of moderate coupling, but only when we consider MF composites with mid-myocardial myocytes.
Figure 4. Representative time series of the transmembrane potential recorded from an MF composite. Plots of the transmembrane potential $V_m$ of the myocyte in the myocyte-fibroblast (MF) composite versus time $t$ for the strong-coupling case $G_{gap} \approx 8$ nS showing the following: responses of type R1 for $N_f = 1$, $E_f = -49$ mV, and (a.1) epicardial myocytes, (a.2) mid-myocardial myocytes, and (a.3) endocardial myocytes; responses of type R2 for (b.1) epicardial myocytes and $N_f = 4$, and $E_f = -21$ mV, (b.2) mid-myocardial myocytes and $N_f = 5$, and $E_f = -32$ mV, and (b.3) endocardial myocytes and $N_f = 4$, and $E_f = -22$ mV; autorhythmic responses of type R3 for (c.1) epicardial myocytes and $N_f = 5$, and $E_f = -34$ mV, (c.2) mid-myocardial myocytes and $N_f = 2$, and $E_f = -20$ mV, and (c.3) endocardial myocytes and $N_f = 4$, and $E_f = -29$ mV; oscillatory responses of type R4 for (d.1) epicardial myocytes and $N_f = 3$, and $E_f = -2$ mV, (d.2) mid-myocardial myocytes and $N_f = 4$, and $E_f = -19$ mV, and (d.3) endocardial myocytes and $N_f = 3$, and $E_f = -9$ mV; responses of type R5 for (e.1) epicardial myocytes and $N_f = 6$, and $E_f = -30$ mV, (e.2) mid-myocardial myocytes and $N_f = 5$, and $E_f = -13$ mV, and (e.3) endocardial myocytes and $N_f = 8$, and $E_f = -16$ mV.

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Figure 5. Stability diagrams in the $E_f - N_f$ parameter space for the responses of an MF composite. The regions in which the MF composite shows responses of types R1–R5 are shown in (a.1), (a.2), (b.1), (b.2), (c.1), and (c.2). For the moderate-coupling case, $G_{gap} = 4$ nS, (a.1), (b.1), and (c.1) show the stability diagrams for, respectively, epicardial, endocardial, mid-myocardial, and epicardial myocytes in the MF composite; their analogs for the strong-coupling case, $G_{gap} = 8$ nS, are given in (a.2), (b.2), and (c.2); the régimes R1, R2, R3, R4, and R5 are denoted, respectively, by yellow hexagrams, red squares, green circles, pink diamonds, and blue pentagrams.

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Plane-wave propagation in a 2D domain

We now investigate the propagation of plane waves of electrical activation through a 2D simulation domain of the type we have described above. In this domain, we distribute myocytes and fibroblasts randomly such that the percentage of fibroblasts is $P_f$; we consider propagation through both mural and transmural slices. In addition to $P_f$, the other important parameters in this part of our study are $E_f$ and the components of the diffusion tensors, i.e., $D_{xx}^{mf}$, $D_{yy}^{mf}$, and $D_{ij}^{mf}$ and $D_{ij}^{00}$ (see Eqs. 4). Recall that, in our 2D and 3D models, there is no on-site coupling between myocytes and fibroblasts; but we have diffusive couplings between such cells if they are at nearest-neighbor sites in our simulation domains; here $P_f$ plays a role similar to that of $N_f$, in our studies of MF composites. We show below that the temporal responses, of types $R_1$–$R_5$, for MF-composites, have spatiotemporal analogs when we consider plane-wave propagation through our 2D simulation domain; we denote these analogs by the same symbols, namely, $R_1$–$R_5$, because the spatiotemporal evolution of the plane waves in these stability regimes can be rationalized, qualitatively, in terms of the responses of MF composites that we have discussed above.

We first consider plane-wave propagation through a mural slice. We find the five qualitatively different spatiotemporal behaviors $R_1$–$R_5$. In the régime $R_1$, which occurs both in moderate- and strong-coupling cases, the plane wave propagates smoothly through the simulation domain but with a slightly corrugated wave front. In the régime $R_2$, small clusters of fibroblasts can form around some sites with myocytes; these clusters have the capacity to generate one subsidiary action potential (cf. the response $R_2$ of an MF composite), before returning to a resting potential that is above the resting potential of the myocytes; because of this subsidiary action potential, target waves are generated by the fibroblast clusters and a plane wave, which tries to propagate through the simulation domain, is annihilated by these target waves, so the whole domain returns to a potential that is above the normal resting membrane potential of myocytes, but below their threshold potential; $R_2$ is absent in the moderate-coupling case, in the parameter regimes that we have explored. The parameter régime $R_3$ is characterized by autorhythmicity; the fibroblast clusters about some myocyte now develop the ability to sustain rhythms of their own (cf. the response $R_3$ of the MF composite); here too the initial plane wave is annihilated by the target waves that are generated by the autorhythmic fibroblast clusters; however, unlike in the case $R_2$, the activity of our medium does not stop here; after a considerable length of time, the autorhythmic fibroblast clusters generate sustained beats of their own; beats from fibroblast clusters of different sizes, which are in different parts of the medium, are incoherent; $R_3$ is absent in the moderate-coupling case, in the parameter regimes that we have explored. In the oscillatory régime $R_4$ (cf. the response $R_4$ of the MF composite), the fibroblast clusters produce an initial target wave that annihilates the plane wave; this is followed by temporal oscillations, about some mean potential, of the local membrane potential; $R_4$ is absent in the moderate-coupling case, in the parameter régimes that we have explored. In régime $R_5$ the initial plane wave is terminated by collisions with numerous target waves, which are generated by the fibroblast clusters that are distributed randomly throughout the medium; once the plane wave is removed, the medium moves into a quiescent state with a membrane potential that lies above the excitation-threshold potential for an uncoupled myocyte; no further excitation is possible; $R_5$ is absent in the moderate-coupling case, in the parameter régimes that we have explored. The stability diagram, which shows the regions with spatiotemporal behaviors $R_1$–$R_5$ in the strong-coupling case, is illustrated in Fig. 6; regions $R_1$, $R_2$, $R_3$, $R_4$, and $R_5$ are denoted, respectively, by blue diamonds, green triangles, pink pentagrams, black squares, and red circles; the spatiotemporal evolution of plane waves in these regions is described in the text.

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Figure 6. Stability diagram in the $E_f-P_f$ plane for plane-wave propagation through a mural slice of our 2D simulation domain with a random distribution of myocytes and fibroblasts. The stability diagram shows the regions with spatiotemporal behaviors $R_1$–$R_5$ in the strong-coupling case ($D_{ii}^{mf} = 0.000048$ cm$^2$/ms); the regions $R_1$, $R_2$, $R_3$, $R_4$, and $R_5$ are denoted, respectively, by blue diamonds, green triangles, pink pentagrams, black squares, and red circles; the spatiotemporal evolution of plane waves in these regions is described in the text.

Representative pseudocolor plots of the local membrane potential $V(x,y,t)$ are given in Fig. 7 for several values of the time $t$ to illustrate plane-wave propagation, through a 2D mural slice, in the moderate-coupling case, for different values of $P_f$. $V(x,y,t) = V_m(x,y,t)$, if the site $(x,y)$ is occupied by a myocyte, and $V(x,y,t) = V_f(x,y,t)$, if the site $(x,y)$ is occupied by a fibroblast.) We do not see behaviors of types $R_2$–$R_5$ here; the plane wave propagates through the medium with a slightly corrugated wave front (region $R_1$). Video S1 shows the spatiotemporal evolution of the plane waves in Figs. 7(a.1), (b.1), (c.1), (d.1), (e.1), and (f.1).

Analogous plots, for the strong-coupling case, of plane-wave propagation, through a 2D mural slice, are shown in Fig. 8; plane-wave propagation for régime $R_1$ is illustrated in Figs. 8(a.1)–(a.4), for régime $R_2$ in Figs. 8(b.1)–(b.4), for régime $R_3$ in Figs. 8(c.1)–(c.4), for régime $R_4$ in Figs. 8(d.1)–(d.4), and for régime $R_5$ in Figs. 8(e.1)–(e.5). The spatiotemporal evolution of these plane waves is given in Video S2.

We turn now to illustrative studies of plane-wave propagation through a 2D transmural slice. Here too, we find the five qualitatively different spatiotemporal behaviors $R_1$–$R_5$. In the régime $R_1$, which occurs both in moderate- and strong-coupling cases, the plane wave propagates smoothly through the simulation domain but with remarkable distortion. In the moderate-coupling case, at low values of $P_f$, the wavefront acquires a smoother appearance than in the 2D mural slice; the smoothness begins to disappear as $P_f$ increases. Furthermore, these waves propagate differently within the three layers of the heart wall, inside the simulation domain. For sufficiently large values of $P_f$, electrical conduction is partially blocked in the mid-myocardium and completely blocked in the endocardium; the excitation then travels...
only along the epicardium as illustrated in Fig. 9. In the strong-coupling case, régime R1 occurs only at low values of $P_f$, as in the moderate-coupling case; and here the wave has a smooth wave front. Régimes R2, R3, R4 and R5, analogous to those in the strong-coupling case of the 2D mural slice, are also observed in the strong-coupling case of the 2D transmural slice. However, these are absent in the moderate-coupling case, in the parameter régimes that we have explored. The stability diagram, which shows the regions with spatiotemporal behaviors R1–R5 in the strong-coupling case, is given in Fig. 10; regions R1, R2, R3, R4, and R5 are denoted, respectively, by blue diamonds, green triangles, pink pentagrams, black squares, and red circles.

Representative pseudocolor plots of the local membrane potential $V(x,y,t)$ are given in Fig. 9 for several values of the time $t$ to illustrate plane-wave propagation, through a 2D transmural slice, in the moderate-coupling case, for different values of $P_f$. We do not see behaviors of types R2–R5 here; the plane wave propagates, through the medium, with a distorted wave front (region R1). Video S3 shows the spatiotemporal evolution of the plane waves in Figs. 9(a.1)–(a.5) for $P_f = 5\%$, in (b.1)–(b.5) $P_f = 10\%$, in (c.1)–(c.5) $P_f = 15\%$, in (d.1)–(d.5) $P_f = 20\%$, in (e.1)–(e.5) $P_f = 25\%$, and in (f.1)–(f.5) $P_f = 30\%$. (For full spatiotemporal evolutions see Video S1.)

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Figure 7. Pseudocolor plots of the local membrane potential $V$ illustrating plane-wave propagation through a mural slice of our 2D simulation domain with a random distribution of myocytes and fibroblasts. Here we consider the moderate-coupling case $D_{mf} = 0.0000228 \, \text{cm}^2/\text{ms}$, and $E_f = -30 \, \text{mV}$ and in (a.1)–(a.5) the percentage of fibroblasts $P_f = 5\%$, in (b.1)–(b.5) $P_f = 10\%$, in (c.1)–(c.5) $P_f = 15\%$, in (d.1)–(d.5) $P_f = 20\%$, in (e.1)–(e.5) $P_f = 25\%$, and in (f.1)–(f.5) $P_f = 30\%$. (For full spatiotemporal evolutions see Video S1.)

Dependence of the conduction velocity and the wavelength on the percentage of fibrosis

We characterize the influence of fibroblasts on plane-wave propagation through our mathematical model for myocardial tissue with fibroblasts by studying the dependence of the plane-wave conduction velocity $CV$ and the wavelength $\lambda$ on the percentage of fibrosis $P_f$; we present illustrative studies at a fixed value of the resting membrane potential of fibroblasts, namely,
we increase then may or may not show conduction blockage, depending on $V$.

We call this state SRSP (Single-Rotating-Spiral-Periodic); a representative pseudocolor plot of the local membrane potential $V$ is given in Fig. 13(a) for $t = 10$ s; the time series of $E_t$ is shown alongside in Fig. 13(b). For full spatiotemporal evolutions see Video S3.

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$E_t = -40 \text{mV}$; we choose this value because, from our single-MF composite studies, it is evident that, at such a moderately low value of $E_t$, the MF composite responds to electrical stimuli as in the régime R1, so it is convenient to measure $CV$. When $P_f = 0\%$ we find that $CV \approx 70\%$; the typical value for plane-wave propagation through the human myocardium; and $\lambda \approx 19\text{cm}$. As we increase $P_f$, in the moderate-coupling case, $CV$ decreases gradually, as does $\lambda$. When the MF diffusive coupling is strong, $CV$ decreases gradually at first, but then, once the fibroblast clusters become large enough to generate target waves that can annihilate the plane wave, $CV$ falls rapidly to zero. The medium then may or may not show conduction blockage, depending on whether it has passed into the régime R3, or is still in R2, R3, or R4. Plots of $CV$ and $\lambda$ versus $P_f$ are given, respectively, in Figs. 12 (a) and (b), for both moderate-coupling (open blue circles) and strong-coupling (filled black circles) cases.

**Influence of diffuse fibrosis on spiral waves in 2D**

e now explore the dynamics of spiral waves of electrical activation in our mathematical model in the presence of fiber anisotropy and diffuse fibrosis. We start with a monolayer of myocytes ($P_f = 0\%$) and the initial condition of Fig. 1 (a); we observe that, even after $t = 20$ s, the medium supports only one, temporally periodic, rotating spiral wave, which shows no breaks. We call this state SRSP (Single-Rotating-Spiral-Periodic); a representative pseudocolor plot of the local membrane potential $V(x,y,t)$ is given in Fig. 13(a) for $t = 10$ s; the time series of $E_t$, recorded from a point near the corner of the simulation domain, i.e., from $(x = 2.25\text{cm}, y = 2.25\text{cm})$, is shown alongside in Fig. 13(b); Fig. 13(c) shows the power spectrum $E(\omega)$ of this time series; and the corresponding plot of the inter-beat interval (IBI) versus the beat number $n$ is depicted in Fig. 13(d). The simple periodicity of this time series, the appearance of a single, major peak in $E(\omega)$ at the fundamental frequency $\omega_f \approx 4\text{Hz}$, and the constancy of the IBI confirm that the spiral wave in SRSP evolves completely periodically in time.

Next we increase $P_f$ in steps of 1%. For $P_f < 14\%$, the system continues in the state SRSP; but, as $P_f$ approaches 14%, the single, completely periodic, spiral-wave develops a granular texture that increases with $P_f$; the distance from the wave-front to the wave-back also decreases. In Fig. 14 we show, for representative values of $P_f$, pseudocolor plots of the local transmembrane potential $V(x,y,t)$; these plots illustrate the time evolution of a spiral wave in six qualitatively different states, namely, SRSP, SRSQ, MRSP, MRSQ, ST, and SA, which we have defined above; the spatiotemporal evolution of $V(x,y,t)$ for these states is shown in Video S5. The states SRSP and SRSQ have single spirals that rotate periodically and quasiperiodically, respectively; MRSP and MRSQ have multiple spirals whose temporal evolution is periodic and quasiperiodic, respectively; the state ST displays spiral-wave turbulence; and in SA the spiral wave is absorbed at the boundaries of our simulation domain.

To examine the temporal evolution of spiral waves in these states, it is useful to look at time series of $V(x,y,t)$, from representative points in the simulation domain, and the resulting plots of the IBI and the power spectra $E(\omega)$. These are shown for illustrative values of $P_f$ in Figs. 15 and 16.

In Figs. 15 (a.1)–(d.3) we have chosen the values of $P_f$ so that we can show examples of temporal 2-cycles (Figs. 15 (a.1)–(a.3) for $P_f = 21\%$), 5-cycles (Figs. 15 (b.1)–(b.3) for $P_f = 21.3\%$), 4-cycles (Figs. 15 (c.1)–(c.3) for $P_f = 16.8\%$), and 5-cycles (Figs. 15 (d.1)–
We have chosen the values of $P_f$ so that we can show examples of temporal 6-cycles (Figs. 16 (a.1)–(a.3) for $P_f = 21.1\%$), 7-cycles (Figs. 16 (b.1)–(b.3) for $P_f = 20.7\%$), 9-cycles (Figs. 16 (c.1)–(c.3) for $P_f = 16.6\%$), and 10-cycles (Figs. 16 (d.1)–(d.3) for $P_f = 16.1\%$); these cycles show up most clearly in the IBI plots (Figs. 15 (a.2), (b.2), (c.2), and (d.2)) but their presence can also be surmised from the time series of $V$ (Figs. 15 (a.1), (b.1), (c.1), and (d.1)) and the sharp peaks in the power spectra (Figs. 15 (a.3), (b.3), (c.3), and (d.3)).

In Figs. 16 (a.1)–(d.3) we have chosen the values of $P_f$ so that we can show examples of temporal 6-cycles (Figs. 16 (a.1)–(a.3) for $P_f = 21.1\%$), 7-cycles (Figs. 16 (b.1)–(b.3) for $P_f = 20.7\%$), 9-cycles (Figs. 16 (c.1)–(c.3) for $P_f = 16.6\%$), and 10-cycles (Figs. 16 (d.1)–(d.3) for $P_f = 16.1\%$); these cycles show up most clearly in the IBI plots (Figs. 16 (a.2), (b.2), (c.2), and (d.2)) but their presence can also be surmised from the time series of $V$ (Figs. 16 (a.1), (b.1), (c.1), and (d.1)) and the sharp, fundamental frequencies in the power spectra (Figs. 16 (a.3), (b.3), (c.3), and (d.3)).

Long time series are required to ascertain the temporal periodicity of these states. Here we obtain local time series for $V(x,y,t)$, from the representative point $(x=6.75\text{cm}, y=6.75\text{cm})$, for $0 \leq t \leq 20\text{ s}$, which corresponds to $10^6$ time steps; to remove the effects of initial transients, it is best to disregard data from the first 300000 iterations or so. Given plots such as those of Fig. 15 and 16, we can systematize the sequence of transitions that leads from the state SRSP to SA. For the initial conditions and the distributions of fibroblasts that we use, the sequence of transitions is shown in Fig. 17(a). The exact sequence in which these transitions occur depends sensitively on the initial conditions, boundary effects, and the realizations of fibroblast distributions within the domain, as in other nonequilibrium transitions (see, e.g., Refs. [40–42]).
We have found both oscillatory and autorhythmic states. Although the target waves in both these cases are similar, those in the autorhythmic case have a larger amplitude than in the oscillatory case. Note, furthermore, that the states SRSP, ST, and SA can be identified merely from the time series of $V_m$, with data recorded from any representative point in the simulation domain: The time series for $V_m$ in the state SRSP, is completely periodic, so the plot of IBI versus the number $n$ of the beat is a flat line; in the state ST this time series is obviously chaotic; in the state SA the time series is a flat line, which indicates that there is no trace of activity. In contrast, the states SRSQ, MRSP, and MRSQ cannot be identified unambiguously from a quick inspection of the time series of $V_m$, i.e., a plot versus the frequency $\omega$ of the power spectrum $E(\omega)$ of this time series; (D) a plot of the inter-beat interval (IBI) versus the beat number $n$ for this time series (here we have discarded the first 10 beats to remove the initial transients.

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Figure 11. Pseudocolor plots of the local membrane potential $V$ illustrating plane-wave propagation through a transmural slice of our 2D simulation domain with a random distribution of myocytes and fibroblasts. Here we consider the strong-coupling case and different regimes in the stability diagram of Fig. 10: (a.1)–(a.4) propagation in régime R1 ($P_f=0\%$, $E_f=-30mV$); (b.1)–(b.4) propagation in régime R2 ($P_f=5\%$, $E_f=-30mV$); (c.1)–(c.4) propagation in régime R3 ($P_f=10\%$, $E_f=-30mV$); (d.1)–(d.4) propagation in régime R4 ($P_f=12\%$, $E_f=-30mV$); (e.1)–(e.4) propagation in régime R5 ($P_f=20\%$, $E_f=-30mV$). (For full spatiotemporal evolutions see Video S4.).

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Figure 12. The dependence of the plane-wave conduction velocity $CV$ and wavelength $\lambda$ on the percentage of fibrosis $P_f$. Plots of (a) CV and (b) $\lambda$ versus $P_f$ for the moderate-coupling case (solid black line with filled black circles), i.e., $D_{mf}=0.0000218$ cm$^2$/ms, and the strong-coupling case (solid blue line with unfilled blue circles), i.e., $D_{mf}=0.000048$ cm$^2$/ms.

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earlier studies of spiral waves in mathematical models of cardiac tissue [43–45] without fibroblasts).

Influence of diffuse fibrosis on scroll waves in 3D

We consider now the dynamics of scroll waves of electrical activation in our mathematical model in the presence of fiber anisotropy and diffuse fibrosis. We start with a rectangular parallelepiped of myocytes ($P_f \sim 0\%$) and the initial condition of Fig. 1 (b). We find that, even after $t \sim 4$ s, the medium supports only one, temporally periodic, rotating scroll wave, which does not break up further into smaller scrolls. We call this state SRS (Single-Rotating-Scroll). We now increase $P_f$ in steps of 1% and find that, as $P_f$ increases, this periodic, scroll-wave develops a granular texture, whose granularity increases with $P_f$; the distance from the wave-front to the wave-back also decreases. In Fig. 18 we show, for representative values of $P_f$, isosurface plots of the local transmembrane potential $V(x,y,t)$ that illustrate the time evolution of a scroll wave in three qualitatively different states, namely, SRS, MRS, and SA, which we have defined above; the spatiotemporal evolution of $V(x,y,t)$ for these states is shown in Video S6. The states SRS and MRS have single and multiple scrolls, respectively; their temporal evolution may be periodic, quasiperiodic, or chaotic; to determine this unambiguously, we need far longer time series than we have been able to get with our computational resources. However, we can distinguish clearly between the states SRS, MRS, and SA. Given our initial conditions and the distributions of fibroblasts, the sequence of transitions in our 3D model is shown in Fig. 17(b). As we have

Figure 14. Pseudocolor plots of the local membrane potential $V$ illustrating spiral-wave dynamics in a mural slice of our 2D simulation domain with a random distribution of myocytes and fibroblasts. We obtain six qualitatively different behaviors, namely, SRSP (Single Rotating Spiral Periodic), SRSQ (Single Rotating Spiral Quasiperiodic), MRSP (Multiple Rotating Spirals Periodic), MRSQ (Multiple Rotating Spirals Quasiperiodic), ST (Spiral Turbulence), and SA (Spiral Absorption). Illustrative pseudocolor plots of $V$ show the time evolution of a spiral wave for (a.1)–(a.5) SRSP with $P_f \sim 5\%$, (b.1)–(b.5) SRSQ with $P_f \sim 21\%$, (c.1)–(c.5) MRSP with $P_f \sim 17.8\%$, (d.1)–(d.5) MRSQ with $P_f \sim 18.2\%$, (e.1)–(e.5) ST with $P_f \sim 20.5\%$, and (f.1)–(f.5) SA with $P_f \sim 25\%$. (For full spatiotemporal evolutions see Video S5.).
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noted in the 2D case, the exact sequence in which these transitions occur depends sensitively on the initial conditions, boundary effects, and the realizations of fibroblast distributions.

Discussion

We have presented a comprehensive numerical study of spiral- and scroll-wave dynamics in a state-of-the-art mathematical model for human ventricular tissue with fiber rotation, transmural heterogeneity, myocytes and fibroblasts. Our mathematical model introduces fibroblasts randomly, to mimic diffuse fibrosis, in the TNNP model [16,23] for human ventricular tissue; the passive fibroblasts in our model do not exhibit an action potential in the absence of coupling with myocytes; and we allow for a coupling between nearby myocytes and fibroblasts.

Our in silico study is designed to explore effectively biophysically relevant ranges of the parameters that characterize myocytes,

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{Figure15}
\caption{Time series of \( V \) illustrating high-order temporal cycles during spiral-wave propagation in a mural slice of our 2D simulation domain with a random distribution of myocytes and fibroblasts. Plots of the time series of \( V(x,y,t) \), from representative points in the simulation domain, and the resulting plots of the interbeat interval IBI versus the beat number \( n \) and the power spectrum \( E(\omega) \) versus the frequency \( \omega \). Illustrating temporal 2-cycles (a.1)–(a.3) for \( P_f \approx 21\% \), 3-cycles (b.1)–(b.3) for \( P_f \approx 21.3\% \), 4-cycles (c.1)–(c.3) for \( P_f \approx 16.8\% \), and 5-cycles (d.1)–(d.3) for \( P_f \approx 20.3\% \); these cycles show up most clearly in the IBI plots (a.2), (b.2), (c.2), and (d.2); but their presence can also be surmised from the time series of \( V \) (a.1), (b.1), (c.1), and (d.1) and the sharp peaks in the power spectra (a.3), (b.3), (c.3), and (d.3)).
\doi{10.1371/journal.pone.0045040.g015}
\end{figure}
fibroblasts, and their interactions. Thus, our work complements, in an important way, experimental studies of electrical-wave propagation in fibrotic cardiac tissue [4, 25]; and, as we have mentioned above, it extends significantly the numerical studies initiated by Panfilov [11] and ten Tusscher, et al. [1, 12, 13].

Simulations by Maleckar, et al. [46] on a rabbit ventricular model suggest that the myocyte resting potential and AP waveform, in the case of atrial arrhythmias, are modulated strongly by the properties and number of coupled fibroblasts, the degree of coupling, and the pacing frequency.

Xie, et al. [47] have shown that a fibroblast, coupled with a myocyte, generates a gap-junction current, which flows from the myocyte to the fibroblasts and vice versa, with two main components: an early pulse of transient outward current and a

Figure 16. Time series of $V$ illustrating high-order temporal cycles during spiral-wave propagation in a mural slice of our 2D simulation domain with a random distribution of myocytes and fibroblasts. Plots of the time series of $V(x,y,t)$, from the representative point $(x = 67.5\text{mm}, y = 67.5\text{mm})$, and the resulting plots of the interbeat interval IBI versus the beat number $n$ and the power spectrum $E(\omega)$ versus the frequency $\omega$ illustrating temporal 6-cycles (a.1)–(a.3) for $P_f = 21.1\%$, 7-cycles (b.1)–(b.3) for $P_f = 20.7\%$, 9-cycles (c.1)–(c.3) for $P_f = 16.6\%$, and 10-cycles (d.1)–(d.3) for $P_f = 16.1\%$; these cycles show up most clearly in the IBI plots (a.2), (b.2), (c.2), and (d.2); but their presence can also be surmised from the time series of $V$ (a.1), (b.1), (c.1), and (d.1) and the sharp peaks in the power spectra (a.3), (b.3), (c.3), and (d.3)).

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Figure 17. Representative band diagrams of states, in our 2D and in 3D studies, illustrating transitions between different spiral-wave (for 2D) and scroll-wave (for 3D) states as a function of $\frac{P_f}{P_i}$. Top panel: This band diagram shows the rich sequence of transitions, from one nonequilibrium state to another, that takes us from the state SRSP, which occurs predominantly at low values of $P_f$, to the state SA, which occurs at large values of $P_i$; the values of $\frac{P_f}{P_i}$ are given below the band and the six states SRSP-SA are shown by a gray scale. Bottom panel: This band diagram shows the sequence of transitions, from one nonequilibrium state to another, that takes us from the state SRSQ, which occurs predominantly at low values of $P_f$, to the state SA, which occurs at predominantly large values of $P_i$; the values of $\frac{P_f}{P_i}$ are given below the band and the three states SRSQ-SA are shown by a gray scale. The fine resolution of the transitions in 2D (top panel) cannot be achieved easily in 3D (bottom panel) without a prohibitive increase in computational costs.

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Our investigation of a single MF composite, with a single myocyte coupled to $N_f$ fibroblasts via a gap-junctional conductance $G_{gap}$, reveals five qualitatively different responses for this composite, namely, R1–R5. In R1 the response of the MF composite to an external electrical stimulus is like that of an uncoupled myocyte; in R2 this response has an additional action potential; responses R3 and R4 are autorythmic and oscillatory, respectively; in R5 the MF composite produces a single AP after which it reaches a time-independent, non-excitable state.

Our studies of 2D domains with a random distribution of fibroblasts in a myocyte background reveal that, as the percentage $P_f$ of fibroblasts increases, the $CV$ of a plane wave decreases, slowly at first and rapidly thereafter, until it reaches zero and there is conduction failure. If we consider spiral-wave dynamics in such a medium we find, in 2D, a variety of nonequilibrium states, temporally periodic (SRSP and MRSP), quasiperiodic (SRSQ, MRSQ), chaotic (ST), and quiescent (SA), and an intricate sequence of transitions between them (see Fig. 17 (a)). The analogous sequence of transitions for 3D scroll waves is given in Fig. 17 (b). As we have noted above, such transitions between nonequilibrium states in extended dynamical systems are known in a variety of problems including the onset of turbulence in pipe flow [40], dynamo transitions in magnetohydrodynamics [41], and the turbulence-induced melting of vortex crystals in two-dimensional soap films [42]. The precise sequence of such transitions often depends on initial conditions, boundary conditions, and, in the case we consider, on the random distribution of fibroblasts. However, the important qualitative points to note in our study are that (a) there is a variety of nonequilibrium states and (b) a rich sequence of transitions between them. These states can have important physical consequences. In particular, we speculate that the autorhythmic and oscillatory behaviors in the states R3 and R4 offer a possible model for ectopic foci. Thus, our studies of plane-, spiral-, and scroll-wave dynamics in our simulation domains with myocytes and fibroblasts can provide important qualitative insights into the possible effects of fibrosis on the propagation of electrical waves of activation in human ventricular tissue. In this sense, our work also builds upon the following studies: The in-vitro investigations of Miragoli, et al. [49] also suggest that fibroblasts, introduced into myocardial tissue by pressure overload or infarction, might lead to arrhythmogenesis via ectopic activity; the numerical studies of Jacquemet [50] also suggest that pacemaker-type activity can result from the coupling of cardiomyocytes with non excitable cells like fibroblasts; and Kryukov, et al. [51] have concluded, via in vitro and numerical studies of heterogeneous cardiac cell cultures and mathematical models thereof, that mixtures of excitable cells, which are initially silent, and passive cells can show transitions to states with oscillatory behavior. Interesting nonequilibrium transitions between different dynamical regimes have also been seen studied recently in a two-dimensional model for uterine tissue [52].

Our results are qualitatively in consonance with those of McDowell, et al. [22], who have used the Mahajan model [53] of the rabbit ventricular myocyte in a monodomain model in an anatomically realistic rabbit ventricular domain. In particular, they find that low densities of fibroblasts do not have a significant influence on the susceptibility to arrhythmias, moderate levels of fibroblasts increase the propensity for arrhythmias because of APD dispersion, and high fibroblast densities lead to conduction blockage. Their simulation domain is anatomically realistic whereas ours is not; however, we use the TNBP model for human cardiac tissue in contrast to the rabbit-ventricular model employed by them; furthermore, we carry out simulations at many more values of the fibroblast concentration than they do and, therefore, our simulations can uncover the details of the nonequilibrium transitions from single rotating spiral or scroll waves to the absorption state with no waves.

Tanaka, et al. [54] have studied how the random distribution of fibroblasts affects the dynamics of atrial fibrillation (AF) in sheep cardiac tissue in which heart failure (HF) has been induced artificially; they have found that the number of fibrous patches is significantly larger after HF than in a control sample. They have also carried out simulation studies by using a two-dimensional human atrial model with structural and ionic remodeling that produce HF; in these simulations they demonstrate that changes in AF activation frequency and dynamics are controlled by the interaction of electrical waves with clusters of fibrotic patches.

Muñoz, et al. [55] have carried out optical-mapping experiments in hetero-cellular monolayers of rat cardiac cells. Their study is designed to test whether fibroblast infiltration modifies the dynamics of spiral waves of electrical activation in such monolayers. One half of the monolayer has a randomly distributed myocyte-fibroblast mixture; the other half has a much larger concentration of myocytes ($\geq 95\%$) than of fibroblasts. In the former case, they find that slow (2.75 Hz), sustained re-entry is stabilized; and the waveform propagates preferentially in the region with a high concentration of myocytes, at twice the conduction velocity ($CV$) than in the region with 50% fibroblasts.

Clinically, the distribution of fibroblasts, in cardiac tissue from a normal, healthy, human heart, has been found to be of the
following two types: (i) long, string-type deposits of collagen or (ii) diffuse and randomly distributed patches \[56,57\]. With advancing age, structural remodeling occurs in the heart; this involves the proliferation of fibroblasts and the formation of interstitial collagen \[56,58\]. It has also been established that there is a significant correlation between increased amounts of fibrotic tissue in the heart and increased incidences of atrial and ventricular tachyarrhythmias and sudden cardiac death \[59–67\]. Furthermore, the partial decoupling of muscle fibers, a decrease in \( CV \), and conduction blocks have been attributed to an increase in fibrosis \[57\]; and there is growing consensus that impaired electrical conduction, which can lead to the formation and breakage of spiral- and scroll- waves of electrical activity, plays an important, though perhaps not exclusive, role in arrhythmogenesis.

Nguyen, et al. \[68\] have used a dynamic voltage-patch-clamp technique on adult rabbit ventricular myocytes, to reveal that the coupling of myocytes to myofibroblasts promotes the formation of early-after-depolarizations (EAD) as a result of a mismatch in early- versus late-repolarization reserve caused by a component of the gap-junction current. These cellular and ionic mechanisms may contribute to the risk of arrhythmia in fibrotic hearts.

The principal limitations of our study are that we use a monodomain description for cardiac tissue and we do not use an anatomically realistic simulation domain. These lie beyond the scope of this study. However, studies by Potse, et al. \[69\] have compared potentials resulting from normal depolarization and repolarization in a bidomain model with those of a monodomain model; these studies show that the differences between results obtained from a monodomain model and those obtained from a bidomain model are extremely small. We intend to study our MF-composite models in anatomically realistic domains and with their bidomain generalizations presently. A detailed study of diffuse fibrosis in an anatomically realistic rabbit ventricle is contained in Ref. \[22\]. In a separate study, we have also investigated \[25\] spiral-wave dynamics in a variant of our mathematical model that is motivated by the experiments of Refs. \[3,70\].

Lastly, the difference between the sizes of the myocytes and fibroblasts is accounted for, in one way, in our model, namely, by virtue of the dependence of the total cellular capacitances of these

**Figure 18. Pseudocolor isosurface plots of the local membrane potential \( V \) illustrating scroll-wave dynamics in a mural slice of our 3D simulation domain with a random distribution of myocytes and fibroblasts.** We obtain three qualitatively different behaviors, namely, SRS (Single Rotating Scroll), MRSP (Multiple Rotating Scrolls), and SA (Scroll Absorption). Illustrative pseudocolor plots with isosurface slicing of \( V \) show the time evolution of a scroll wave for (a.1)–(a.3) SRS with \( P_f = 1\% \), (b.1)–(b.3) MRSP with \( P_f = 17\% \), and (c.1)–(c.5) SA with \( P_f = 15\% \). (For full spatiotemporal evolutions see Video S6).

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two types of cells, because they depend on the surface areas of these cells. Aside from this, our model does not account explicitly for the differences in sizes between myocytes and fibroblasts. However, at large values of $P_f$, it is essential to account for fibroblast size in a more realistic way than we have. One possible way of doing this is to follow the study of Kryukov, et al. [51] in which $N_f$ fibroblasts are allowed to couple to one myocyte; we have studied this for $N_f > 1$ at the level of a single MF composite. The extension of this to two- and three-dimensional domains lies beyond the scope of our paper and will be taken up in a future study.

**Supporting Information**

**Video S1** Plane-wave propagation in the 2D TNNP model with fiber anisotropy, randomly distributed fibroblasts, a mural section, and moderate coupling between the myocytes and the fibroblasts; panels (A), (B), (C), (D), (E), and (F), with $P_f = 5\%, 10\%, 15\%, 20\%, 25\%$, and $30\%$ respectively, show the spatiotemporal evolution of the plane waves in Figs. 7 (a.1), (b.1), (c.1), (d.1), (e.1), and (f.1), via pseudocolor plots of the local transmembrane potential $V(x,y,t)$ for the time interval $0s \leq t \leq 0.3s$, at 25 frames per second. (MPEG)

**Video S2** Plane-wave propagation, shown via pseudocolor plots of the local transmembrane potential $V(x,y,t)$, in the 2D TNNP model with fiber anisotropy, randomly distributed fibroblasts, a mural section, and strong coupling between the myocytes and the fibroblasts for (A) régime R1 (parameters as in Figs. 8 (a.1)–(a.4)), (B) régime R2 (parameters as in Figs. 8 (b.1)–(b.4)), (C) régime R3 (parameters as in Figs. 8 (c.1)–(c.4)), (D) régime R4 (parameters as in Figs. 8 (d.1)–(d.4)), and (E) régime R5 (parameters as in Figs. 8 (e.1)–(e.5)) for the time interval $0s \leq t \leq 1s$, at 25 frames per second. (MPEG)

**Video S3** Plane-wave propagation in the 2D TNNP model in the presence of fiber anisotropy, transmural heterogeneity, randomly distributed fibroblasts, and moderate coupling between the myocytes and the fibroblasts. We show the spatiotemporal evolution of the plane waves, via pseudocolor plots of the local transmembrane potential $V(x,y,t)$, for (A) $P_f = 5\%$ (parameters as in Figs. 9 (b.1)), (B) $P_f = 15\%$ (parameters as in Figs. 9 (c.1)), (C) $P_f = 25\%$ (parameters as in Figs. 9 (d.1)), (D) $P_f = 35\%$ (parameters as in Figs. 9 (e.1)), and (E) $P_f = 40\%$ (parameters as in Figs. 9 (f.1)). The time interval covered is $0s \leq t \leq 0.3s$, and number of frames per second is 25. (MPEG)

**Video S4** Plane-wave propagation in the 2D TNNP model in the presence of fiber anisotropy, transmural heterogeneity, randomly distributed fibroblast and strong coupling between the myocytes and the fibroblasts: We show the spatiotemporal evolution of the plane waves, via pseudocolor plots of the local transmembrane potential $V(x,y,t)$, for (A) régime R1 (parameters as in Figs. 11 (a.1)–(a.4)), (B) régime R2 (parameters as in Figs. 11 (b.1)–(b.4)), (C) régime R3 (parameters as in Figs. 11 (c.1)–(c.4)), (D) régime R4 (parameters as in Figs. 11 (d.1)–(d.4)), and (E) régime R5 (parameters as in Figs. 11 (e.1)–(e.4)). The time interval covered is $0s \leq t \leq 1s$, and number of frames per second is 25. (MPEG)

**Video S5** Spiral-wave dynamics in the 2D TNNP model with diffuse fibrosis. Here we show the spatiotemporal evolution of the spiral waves in Fig. 14, for the representative values of $P_f$ considered there, via pseudocolor plots of the local transmembrane potential $V(x,y,t)$ in the following six states: (A) a single spiral that rotates periodically SRSP, (B) a single spiral that rotates quasiperiodically SRSQ, (C) multiple spirals whose temporal evolution is periodic MRSP, (D) multiple spirals whose temporal evolution is quasiperiodic MRSQ, (E) spiral-wave turbulence ST, and (F) a state SA in which the spiral wave is absorbed at the boundaries of our simulation domain. The time interval covered is $0s \leq t \leq 20s$, and number of frames per second is 10. (MPEG)

**Video S6** Scroll-wave dynamics in the 3D TNNP model with diffuse fibrosis: We show, via isosurface plots of the local transmembrane potential $V(x,y,t)$, the time evolution of a scroll wave in the following three states (for the representative values of $P_f$ in Fig. 18): (A) single rotating scroll SRS, (B) multiple rotating scrolls MRS, and (C) SA, which is characterized by scroll-wave absorption at the boundaries. The time interval covered is $0s \leq t \leq 0.648s$, and number of frames per second is 10. (MPEG)

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**Author Contributions**

Conceived and designed the experiments: RM ARN RP. Performed the experiments: RM. Analyzed the data: RM ARN RP. Contributed reagents/materials/analysis tools: RM RP. Wrote the paper: RM RP.

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