A New Far-Field Cardiac Defibrillation Mechanism

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Abstract

Introduction. Experimental research activity has recently focused on a promising new method for low-energy defibrillation. Called far-field defibrillation, the method imposes electric field pulses that engage the bulk of the heart tissue, in contrast to other methods that deliver electrical energy locally through implanted electrodes. The effectiveness of this method can potentially depend on the timing of the delivery of the pulses. Here we describe a new mechanism by which these electric field pulses might terminate reentrant waves that operates independently of shock timing. Methods. A three-dimensional finite-difference monodomain computer simulation, which includes a full ion channel model and resistive gap junction coupling, is run in rectangular domains of different widths, designed to represent heart walls of varying thicknesses. Once a reentrant action potential scroll wave is established in the system, an electric field stimulus is delivered with varying field vector orientations through the imposition of its effect on the domain boundary conditions. Results. We find that, once the surface perpendicular to the scroll wave filament is depolarized by the electric field, termination of the scroll wave always results. Termination is nearly immediate in the case of thin walls (0.5 cm). In thicker walls (e.g., 2.0 cm), interaction of the induced wave with the scroll wave results in an L-shaped filament, which then shrinks and disappears by the same mechanism by which scroll wave rings terminate. Termination thus occurs independently of wall thickness, timing, and electric field orientation, as long as the latter has a normal component sufficient (about 1 V/cm) to elicit a wave. This new mechanism will likely operate alongside other mechanisms, and thus has the potential to lower the defibrillation threshold.

1. Introduction

Low-energy cardiac defibrillation techniques are desirable for several reasons: less psychological trauma for the patient, less damage to the heart and surrounding tissues, and lower energy utilization when used with implantable defibrillators. Recent efforts towards developing low-energy defibrillation methods have focused on multiple shocks delivered by field electrodes [1–3] rather than single shocks delivered directly to the heart tissue via catheters. Different mechanisms have been proposed for why defibrillation methods of this type should be effective. Shocks delivered in this way may generate action potentials propagating away from structural heterogeneities acting as "virtual electrodes" [4] within cardiac tissue, which may either be effective in disrupting the dynamics of the core of rotating waves [1] or in depolarizing increasingly large fractions of the tissue with each successive shock [3]. Recently, Otani [5] also proposed a mechanism based on wavefront-waveback collisions, and Bragard et al. [6] examined the defibrillating effects present when the rotating wave is confined to a ring. Here we describe a new, fundamentally three-dimensional mechanism that requires a significant component of the applied electric field to be oriented parallel to axis of rotation of the reentrant wave. This mechanism has the advantage that it need not be timed to any particular phase of the rotating wave, nor does it require the electrical stimuli to have any particular spacing in time relative to the dominant period of the rotating action potential waves.

2. Methods

Computer simulations were performed in a three-dimensional, rectangular system meant to represent a portion of one of the walls of the heart. Within this system, we used a simple forward Euler method on a rectangular grid to solve monodomain three-dimensional cable equations of the form,

$$\frac{\partial V}{\partial t} = D\nabla^2 V - I_{ion}(\mathbf{q}, V) \quad (1)$$

$$\frac{\partial \mathbf{q}}{\partial t} = \mathbf{f}(\mathbf{q}, V) \quad (2)$$

Here V is the membrane potential and $\mathbf{q}$ is a vector of the remaining dynamical variables, which consists of ion concentrations and ion channel gating variables. We used the Nygren atrial ion channel model [7] to define the functions

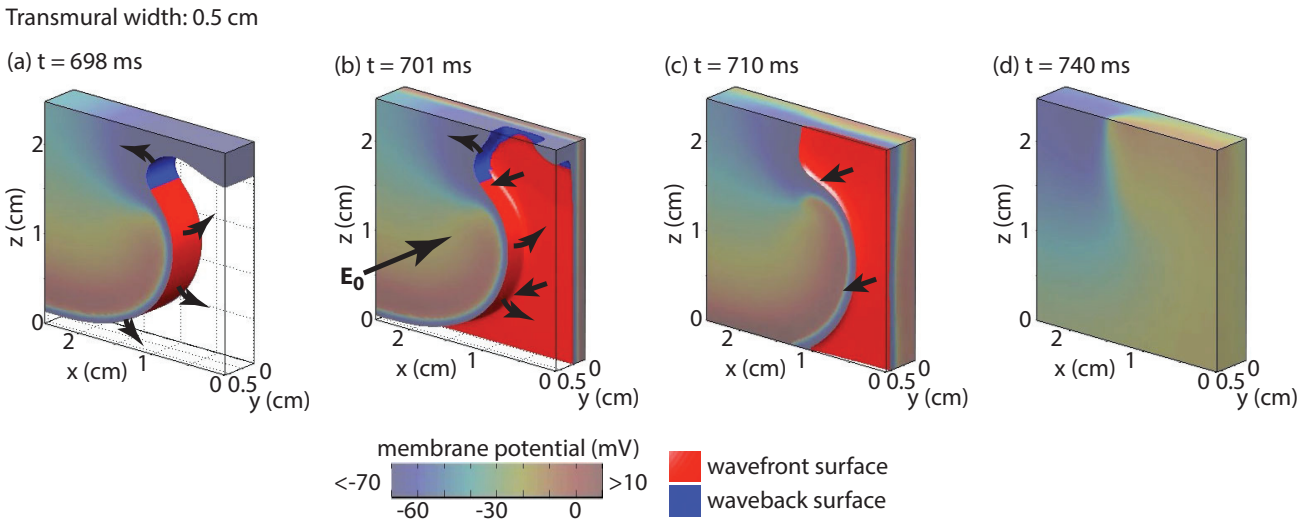

Figure 1. Snapshots of the simulation at four different times for the thin-wall case (wall width = 0.5 cm). The electric field pulse of magnitude 1 V/cm is applied at time $t = 700$ ms for 5 ms. The direction of the field is indicated in panel (b) by the arrow labeled "$\mathbf{E_0}$." The region occupied by the action potential is shown as opaque; the region outside the action potential is transparent. Muted colors on the system surfaces represent the membrane potential values on those surfaces as defined by the accompanying color bar. The bright red surface represents the wavefront of the action potential, while the blue surface defines the waveback. Arrows indicate the directions of motion of these wave surfaces.

$I_{ion}(\mathbf{q}, V)$ and $\mathbf{f}((\mathbf{q}, \mathbf{V})$. Although the Nygren model is an atrial ion channel model, we consider the defibrillation mechanism we find in these simulations to be potentially operative in both the atria and the ventricles. We used this particular model here because it a true ion channel model, and because it supports stable rotating waves, which are, at this stage in our investigation, the central focus of our attention.

The computational grid spacing and timestep were chosen to be 0.025 cm and 0.05 ms, respectively. The membrane potential diffusion coefficient, D, was set to 1.11×10^{-3} cm^2/ms.

Modifications to the Neumann boundary conditions were used to mimic the effect of the defibrillating electric field. Specifically, following the formulation of Pumir and Krinsky (Eq. (5) of [4]), we use as the boundary condition on all surfaces, $\partial V/\partial n = \hat{\mathbf{n}} \cdot \mathbf{E}_0$, where $\hat{\mathbf{n}}$ and $\partial/\partial n$ are the direction and directional derivative normal to the surface, respectively, and $\mathbf{E_0}$ is the applied electric field. Note that this implies "no-flow" ($\partial V/\partial n = 0$) boundary conditions are imposed at all times, except during electric field pulses.

To test the effect of electric field pulses on fibrillatory activity, an action potential scroll wave was initiated in the system with the axis (i.e., the filament) of the wave oriented perpendicular to the two surfaces of the system that we think of as being the surfaces of the heart wall. The filament therefore extends from one of these wall surfaces to the other. The spiral wave was allowed to settle down for 3 rotations prior to the delivery of the effects of the short-pulse, spatially-uniform, external electric field stimulus.

3. Results

We varied the system width in the direction intended to correspond to the direction locally perpendicular to the heart wall surfaces.

The effect of the electric field pulse for the case of a thin wall (0.5 cm) is shown in Fig. 1. Panel (a) shows the pre-existing rotating action potential wave with its axis of rotation (i.e., "'filament") oriented in the transmural direction. In panel (b), 1 ms into the delivery of the shock, we observe expansion of the depolarized region in the form of a propagating action potential plane wave. By $t = 710$ ms (panel (c)), the plane wave has depolarized the entire space not occupied by the rotating wave, killing it, which then leads to repolarization of the entire system, the beginnings of which are shown in panel (d).

In Fig. 2, we see the effect of the electric field pulse on a system with a thicker wall (width: 2.0 cm). As before, a rotating wave occupies the system (panel (a)) prior to the delivery of the electric field shock (in panel (b)). In this case, the trailing (blue) surface of the rotating wave recedes from the original "excitable gap" region before the plane wave has a chance to depolarize it, as shown in panel (c). Instead, the plane wave combines with the original rotating wave to form a modified wave that initially rotates around an L-shaped filament, most easily seen as the boundary between the wavefront surface in red and the waveback sur-

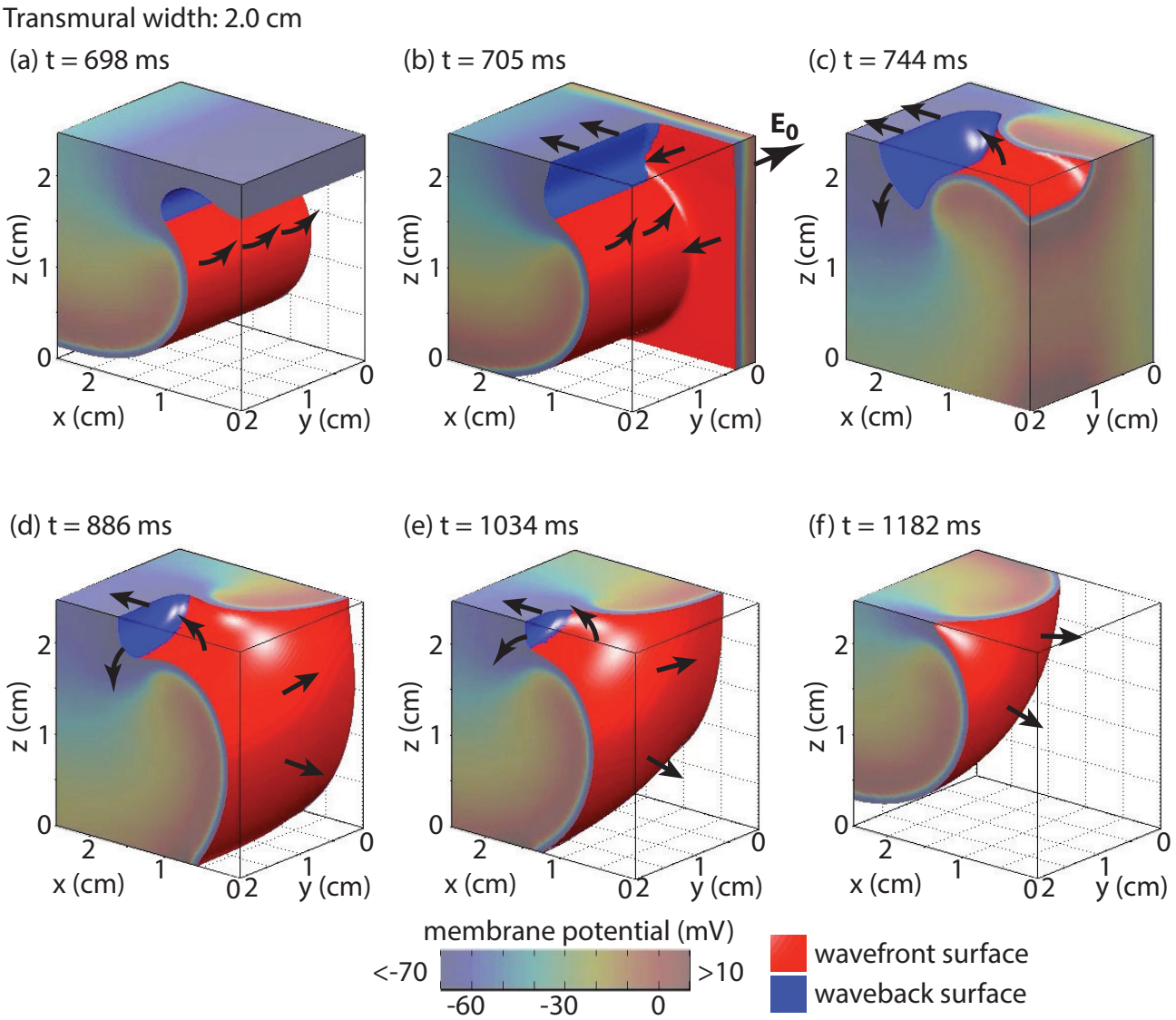

Figure 2. Snapshots of the simulation at six different times for the thick-wall case (wall width = 2.0 cm). An electric field pulse of magnitude 1 V/cm is applied at time $t = 700$ ms for 5 ms. See Fig. 1 for a description of the color scheme and arrows. Note that panels (d), (e) and (f) occur approximately one wave rotation period apart (about 148 ms).

face in blue in panel (b). The L-shaped filament quickly relaxes into a C-shaped filament that runs from the front surface to the top surface of the system, as shown in panel (d). A similar effect was demonstrated previously by Biktashev [8]. This C-shaped filament may be thought of as effectively one quarter of a filament ring, with the rotating wave then being one-fourth of a scroll ring wave. Scroll ring waves tend to self-annihilate as the radius of the ring-shaped filament decreases with time [9]. We observe this type of behavior in this case: panel (e) shows the rotating wave one rotation period later, where we see that the radius of the C-shaped filament is significantly smaller. After one additional rotation period (panel (f)), we find that the filament has disappeared. The wave itself self-annihilates shortly thereafter.

4. Conclusions

We observe that, if the shock electric field can be oriented so that it is perpendicular or nearly perpendicular to as many of the heart wall surfaces as possible, the mechanism shown here can potentially kill all the rotating waves

whose filaments are oriented transmurally. The mechanism clearly works irrespectively of when the shock is applied—essentially the same process described here will occur no matter where the wave is in its rotation. Defibrillation also occurs irrespective of whether the heart wall is thin (as it in the atria) or thick (as in the ventricles). When the walls are thin, the induced wave can depolarize the entire excitable gap, which kills the wave immediately. When the walls are thick, annihilation depends on the shrinking of ring-shaped filaments, which occurs for the case we describe here. Our further investigations will be directed towards determining (a) what happens to rotating waves whose filaments are initially oriented parallel rather the perpendicularly to the wall surfaces, (b) what happens when multiple rotating waves are initially present, (c) what happens when the rotating waves are simultaneously experiencing wave breakup, which, for example, typically occurs in ventricular ion channel models, and (d) what happens when the heart contains regions whose transmural axis is not close to being oriented parallel to the external electric field.

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