Demonstration of a Novel Catheter Guiding Method for the Ablative Therapy of Ventricular Tachycardia

ME Barley[1], AA Armoundas[3], KJ Choppy[2], AM Galea[2], GB Hirschman[2], RJ Cohen[1]

[1]Harvard–MIT Division of Health Sciences and Technology, Cambridge, MA, USA
[2]Infoscitex Corporation, Waltham, MA, USA
[3]Cardiovascular Research Center, Massachusetts General Hospital, Charlestown, MA, USA

Abstract

Current technologies used to guide the radio frequency ablation of ventricular tachycardia are frequently unable to rapidly and accurately localize the site of origin of an arrhythmia, restricting treatment to patients with hemodynamically stable arrhythmias. We investigate the effectiveness of a new rapid inverse algorithm which is based on a single-equivalent dipole representation of cardiac electrical activity. We conducted experiments in a saline volume conductor that involved the guidance of a moving catheter towards the location of a stationary dipole, using only the dipole positions calculated by the inverse algorithm. The algorithm was able to guide the moving catheter tip to well within 2 mm of the stationary dipole in 29 out of 30 experiments. These results suggest that this method has great potential to direct radio-frequency ablation procedures, including in the currently untreatable patient population with hemodynamically unstable arrhythmias.

1. Introduction

Ventricular tachycardia (VT) following a myocardial infarction may result in as many as 200,000 deaths a year in the United States. (1) The most common cause of VT in this context is the formation of a re-entrant circuit (2) for which the optimal current treatment is Radio-Frequency Ablation (RFA). This involves the guidance of an ablation catheter to the exit site or isthmus of the re-entrant circuit and the administration of high-energy, radio-frequency electricity to the tissue. If the tip of the ablation catheter is successfully placed within 2mm of the re-entrant circuit isthmus, the ablation lesion will transect the re-entrant path and the VT will no longer be inducible. However, current technologies are frequently unable to rapidly and accurately localize the optimal site for ablation.

Numerous methods have been proposed for the accurate identification of the optimal site for ablation. One class of such methods attempts to solve the inverse problem, in which the instantaneous pattern of cardiac electrical activity is characterized with a model that can best reproduce the potentials measured on the body surface. Our approach to solving the inverse problem characterizes the instantaneous electrical activity with a Single Equivalent Moving Dipole (SEMD) model (3). We call the method of obtaining the SEMD parameters from potentials recorded at the body surface, the *inverse algorithm*.

The inverse algorithm is first used to calculate the trajectory of the dipole over one beat of VT. Then, the dipole corresponding to the exit site of the re-entrant circuit (the *arrhythmogenic dipole*) is identified through analysis of this trajectory. Next, the location of an ablation catheter tip that has been engineered to produce a current dipole of specific orientation is calculated using the same inverse algorithm. The ablation catheter is guided towards the exit site of the re-entrant circuit using only the calculated positions of the arrhythmogenic and catheter tip dipoles.

The model used by the inverse algorithm assumes an infinite and homogenous volume conductor (4). This model simplification allows the dipole solution to be calculated in real-time. Nevertheless, by ignoring torso inhomogeneities, boundary effects and other non-idealities, the *estimated* position of the arrhythmogenic dipole is displaced from its *true* position by an error vector whose magnitude and direction are dependent on the specific nature of the non-idealities (5). However, if the ablation catheter dipole is also calculated using the same inverse algorithm, when the catheter tip dipole is superposed with the arrhythmogenic dipole, their calculated positions will be displaced from their true positions by the same error vector and their calculated positions will also be the same.

In this paper we evaluate the inverse algorithm as a guidance method with experimental studies conducted in a saline tank. A current dipole at the tip of a moving catheter was guided towards a target stationary dipole (representing the arrhythmogenic dipole) using only the calculated dipole positions, within both a homogeneous and inhomogeneous tank model. This provided an initial assessment of the feasibility of using the inverse algorithm to guide an ablation catheter to the exit site of a re-entrant circuit.

2. Methods

2.1. Experimental setup

Experiments were conducted in an open cylindrical tank of radius 14.6 cm and height 46 cm, approximating the dimensions of an average male torso. The full experimental setup is shown in Figure 1. The tank was filled with standard saline solution. 60 Ag-AgCl electrodes were arrayed over the vertical wall of the tank in a rhombic lattice configuration. One electrode at the base of the tank was chosen as the reference electrode. A rigid vertically-oriented catheter bearing two platinum electrodes 5mm apart at its tip was mounted on an *x-y-z* positioner with a resolution of 0.2 mm and placed inside the tank. The range of the positioner allowed the catheter to be moved within a 20-cm sided ‘heart volume’ approximately centered within the tank, as illustrated in Figure 1. The catheter tip electrodes were connected to an electrically isolated BK Precision model 4011A function generator such that a sinusoidal signal of frequency 100 Hz and amplitude 10 V was created between them. This signal amplitude was chosen in order to generate potentials that ranged from 1 to 10 mV at the surface of the tank, equivalent to the range of a surface ECG. The differential signal between the 59 measurement electrodes and 1 reference electrode was amplified using World Precision Instruments ISO-DAM8 isolated bioamplifiers and passed to the data acquisition system. The entire experimental setup was enclosed within a Faraday cage. A graphical user interface was developed to display the calculated positions and moments of the stationary and catheter tip dipoles.

Torso inhomogeneities were simulated using three cylindrical plastic objects (of dimensions *r*=7.5 cm, *h*=37 cm; *r*=9 cm, *h* = 28 cm; *r* = 8.3 cm, *h*=8 cm) placed in random configurations within the tank.

2.2. Inverse algorithm dipole estimation

The amplitude of the sinusoidal signal in each of the 59 data acquisition channels was acquired by filtering the raw data with a 5th order Butterworth band-pass filter centered at 100 Hz. Due to the proximity of the reference electrode to the measurement electrodes, the estimated potential, $\boldsymbol{\phi}^i$, at the i^{th} channel due to a single dipole in an

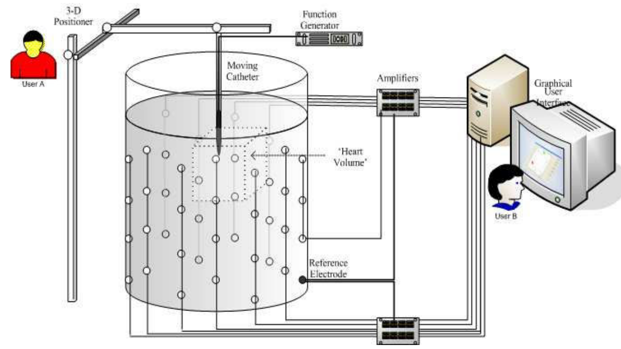

FIG. 1: Experimental Setup

infinite, homogeneous volume conductor is given by:

$$\varphi^i = \frac{\mathbf{p}\cdot(\mathbf{r}-\mathbf{r}_i')}{4\pi g\left|\mathbf{r}-\mathbf{r}_i'\right|^3} - \frac{\mathbf{p}\cdot(\mathbf{r}-\mathbf{r}_{ref}')}{4\pi g\left|\mathbf{r}-\mathbf{r}_{ref}'\right|^3}$$

where $\mathbf{r}_i$’ represents the i^{th} electrode location, $\mathbf{r}$ the dipole location, $\mathbf{r}_{ref}$ the reference electrode location, $\mathbf{p}$ the dipole moment, and g the conductivity of the volume conductor. Due to the linear dependence of ϕ^i on $\mathbf{p}$, the moment can be solved for analytically using the *three-plus-three* parameter optimization algorithm described by Armoundas et al (3). An objective function describes how well a dipole’s estimated surface potentials correlate with the measured ones. The torso volume is searched to find the dipole whose parameters minimize the objective function for each time sample of the measured data.

2.3. Experimental protocol

The experimental aim was to assess whether an operator can reliably guide a catheter tip to the position of a stationary dipole using only the positions calculated by the inverse algorithm. Two operators were needed for each experiment since it was imperative that no knowledge of the real positions of the catheter tip and stationary dipole be used to guide the catheter tip. At the beginning of each experiment, Operator A moved the catheter to a random position within the tank and noted both the real 3-D coordinates of the catheter from the *x-y-z* positioner (the *stationary dipole* location) and the calculated 3-D coordinates of the catheter from the graphical user interface (the *stationary dipole image* location). Operator A then moved the catheter in all three dimensions by a minimum of 5 cm to a second random position within the tank, and recorded both the real 3-D coordinates of the catheter from the positioner (the initial *moving dipole* location) and the calculated 3-D coordinates from the interface (the initial *moving dipole* *image* location). The stationary dipole image location was displayed concurrently with the moving dipole image location on the graphical user interface.

Using only the graphical user interface, Operator B then determined in which direction to move the catheter in order to guide the moving dipole image towards the stationary dipole image. The actual catheter tip was subsequently moved in this direction by between one-half and two-thirds the distance between the dipole images. This distance range was chosen since it represented the most likely size of movement that would be made by a cardiologist: that is, large enough to ensure the rapidity of the procedure but slow enough to minimize the possibility of damage to the ventricular wall and delicate heart structures. After each movement of the catheter, Operator B noted the positions of both the moving dipole and the moving dipole image. Once the moving and stationary dipole images were superposed on the graphical user interface, Operator A recorded the final locations of both the moving dipole and its image.

For the homogeneous tank, 15 trials were conducted, each time using a different stationary dipole location and different initial moving dipole location. A further 15 trials were conducted for an inhomogeneous tank, each time with the inhomogeneities arranged in a different configuration.

3. Results

3.1. Dipole estimation accuracy

Since the inverse algorithm ignores torso inhomogeneities, boundary effects and other non-idealities, the *estimated* position of the dipole is displaced from its *true* position by an error vector whose magnitude and direction are dependent on the specific nature of the non-idealities. By recording the positions of both the dipole and dipole image after each movement of the catheter, the magnitude of this error vector could be evaluated. For every recorded position of the catheter (n=115 for the homogenous case, n=119 for the inhomogeneous case), the distance between the positions of the actual moving dipole and its image was calculated. The error was found to be only slightly greater (p<0.01) for the inhomogeneous tank (2.609 ± 0.835 cm) than for the homogeneous tank (2.306 ± 0.675 cm).

3.2. Real and image path comparison

Next, we compared the path taken by the real moving dipole with that taken by the moving dipole image, for both the homogenous and inhomogeneous tank experiments. Figure 2 illustrates one such set of real and image paths for a normalized end-point. The similarity of

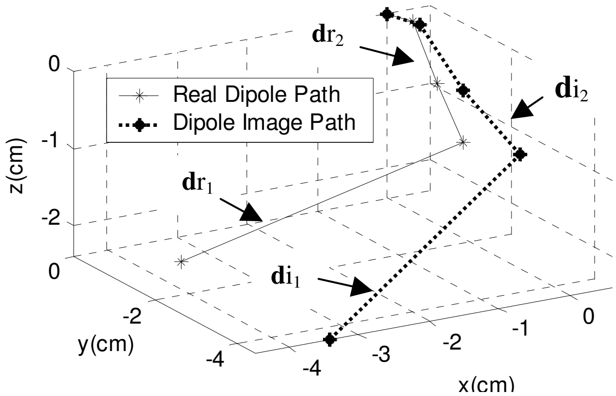

FIG. 2: Real Dipole path (solid) and Dipole Image path (dotted) for a normalized end-point.

the real and image paths was quantitatively assessed by comparing the distances and directions moved by the real and image dipoles. For each movement j of the catheter, the error between the distance moved by the real moving dipole on the catheter tip ($|\mathbf{d}_r,j|$) and the distance moved by the moving dipole's image during the same step ($|\mathbf{d}_i,j|$) is given by:

$$E_{d,j} = \left\| \mathbf{d}r_j - \mathbf{d}i_j \right\|$$

The mean and standard deviation of E_d vs. distance of the moving dipole from the stationary dipole is shown in Figure 3 for both the homogeneous and inhomogeneous tank experiments. The mean and standard deviation of E_d are a fraction of the length of a single catheter movement. Furthermore, as the stationary dipole is approached the error becomes smaller, which is an advantage of this method.

The directions moved by the real dipole and dipole image were also compared, by calculating the angle between $\mathbf{d}_{r,j}$ and $\mathbf{d}_{i,j}$ for each movement of the catheter. The angle between the vectors moved in real and image space was 3.89 ± 1.41 degrees and 3.35 ± 1.28 degrees for the homogenous and inhomogeneous tanks respectively. These results indicate that movements made by the real catheter and moving dipole image coincide both in distance and direction even when significant inhomogeneities are present in the volume conductor. .

3.3. Accuracy of dipole superposition

Lastly, we examined the accuracy with which the moving and stationary dipole positions could be superposed in both the homogenous and inhomogeneous tank experiments. The operator was able to superpose the images of the moving and stationary dipoles in every experiment; the number of steps required to achieve image convergence was 7.33 ± 2.09 for the homogenous

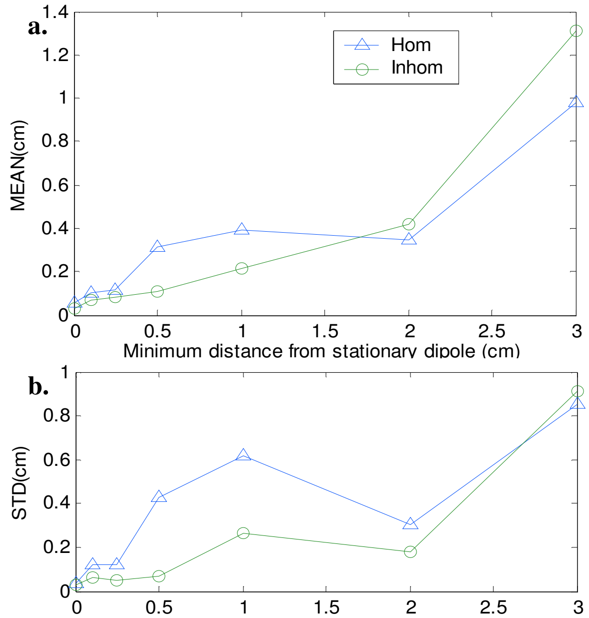

FIG. 3: a). Mean and b). standard deviation of the distance error, E_d, vs. minimum distance of the catheter tip from the stationary dipole for both homogeneous and inhomogeneous tank.

case, and 7.81 ± 2.14 for the inhomogeneous case. The distance between the final actual catheter tip position and actual arrhythmogenic dipole (the *end-point accuracy*) was found to be 0.61± 0.43 mm and 0.55 ± 0.39 mm for the homogeneous and inhomogeneous trials respectively. This mean and standard deviation for the inhomogeneous tank accuracy excludes a single outlier with an end-point accuracy of 28.2 mm. It was noted that in this trial, when the dipole images were superposed, their moments were *not* aligned. This indicates that occasionally, in the presence of significant inhomogeneities or other sources of systematic error, multiple real dipole locations may map to dipoles with the same location in image space but slightly different moments. However, while the inverse algorithm may sometimes falsely indicate the positional superposition of two dipoles, it is able to superpose two dipoles with excellent, sub-2mm, accuracy.

4. Discussion and conclusions

We have demonstrated that the inverse algorithm provides guidance of a catheter tip to the site of a stationary dipole with an accuracy well within the required clinical accuracy of 2mm for radio-frequency ablation. Convergence is achieved in fewer than 10 steps even in the presence of significant inhomogeneities. The accuracy of the superposition of the catheter tip with the stationary dipole was significantly less than 2 mm in all 15 experiments in a homogeneous tank and in 14 out of 15 experiments in an inhomogeneous tank. The algorithm may however falsely indicate the superposition of two dipoles. We are working to characterize the situations under which these false solutions occur; current and future work will also include the development of a method to prevent this (6). However, the speed and accuracy of the inverse algorithm demonstrated in this study suggest that it may provide a new method for guiding radio-frequency ablation procedures both more accurately and to a much wider segment of the population affected by VT.

Acknowledgements

This work was funded by NIH Grant # 1 R44 HL079726-01 and an AHA Scientist Development Grant Award (#0635127N) to AA Armoundas.

References

[1] Thom T, Haase N, Rosamond W, Howard VJ, Rumsfeld J, Manolio T, et al. Heart disease and stroke statistics--2006 update: a report from the American Heart Association Statistics Committee and Stroke Statistics Subcommittee. Circulation 2006;113(6):e85-151.

[2] Zipes DP. What have we learned about Cardiac Arrhythmias? Circulation 2000;102(IV):52-57.

[3] Armoundas AA, Feldman AB, Sherman DA, Cohen RJ. Applicability of the single equivalent point dipole model to represent a spatially distributed bio-electrical source. Med Biol Eng Comput 2001;39(5):562-70.

[4] Armoundas AA, Feldman, AB, Mukkamala, R, and Cohen, RJ., et al. Statistical Accuracy of a Moving Equivalent Dipole Method to Identify Sites of Origin of Cardiac Electrical Activation. IEEE Trans Biomed Eng 2003;50(12):1360-1370.

[5] Armoundas AA, Feldman, AB, Mukkamala, R, and Cohen, RJ. A Single Equivalent Moving Dipole Model: An Efficient Approach for Localizing Sites of Origin of Ventricular Electrical Activation. Annals of Biomedical Engineering 2003;31:564-576.

[6] Barley M, Cohen, RJ. High-Precision Guidance of Ablation Catheters to Arrhythmic Sites using Electrocardiographic Signals. In: Engineering in Medicine and Biology Conference (EMBC); 2006; New York: IEEE; 2006.

Address for correspondence

Maya Barley
E25-335, Massachusetts Institute of Technology
77 Massachusetts Ave. Cambridge, MA, 02139, USA
mbarley@mit.edu