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High-Precision Guidance of Ablation Catheters to Arrhythmic Sites using Electrocardiographic Signals

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Abstract – **Hemodynamically unstable ventricular arrhythmias are frequently untreatable with radio-frequency ablation due to the difficulty of rapidly and accurately localizing the site of origin of an arrhythmia with current technologies. We demonstrate a new catheter guidance method that will direct the tip of an ablation catheter to the site of origin of an arrhythmia and reduce the time needed to locate the site such that a patient need only be maintained in the arrhythmia for a few beats. The algorithm, based on a single-equivalent moving dipole (SEMD) model, is used to identify the bioelectric dipole corresponding to a site of origin of an arrhythmia. If a current dipole is produced at the ablation catheter tip, the tip position may also be calculated using this algorithm, and the catheter can be guided towards the site of origin of the arrhythmia. We present a method to compensate for the effect of systematic non-idealities, such as boundary effects, on the accuracy of this algorithm. In simulations, this method is able to guide the catheter tip to within 1.5 mm of the arrhythmic site at any location within the model torso with almost 100% success and with a realistic number of movements of the ablation catheter. These results suggest that this method has great potential to direct radio-frequency ablation procedures, especially in the significant patient population that is currently untreatable.**

1. INTRODUCTION AND OVERVIEW

Of the 900,000 deaths each year in the US from heart disease [1], at least 36% are sudden and presumed to be caused by ventricular tachyarrhythmias (VT) [2]. 80% of sudden cardiac deaths attributed to tachyarrhythmias are presumed to be due to the after-effects of a myocardial infarction (MI) [3]. The most common etiology of VT in the presence of infarcted tissue is the formation of a re-entrant circuit [4]. Radio-frequency ablation (RFA) treatment of arrhythmias involves the guidance of an ablation catheter to the isthmus or exit site of the reentrant circuit and the administration of high-intensity radio-frequency current to the tissue. If the site has been accurately identified and ablated, the necrotic tissue that remains will transect the isthmus of the re-entrant circuit. The arrhythmia will then be non-inducible.

The *inverse problem* in electrocardiography seeks to characterize the instantaneous pattern of electrical activity within the heart. The *inverse algorithm* solution calculates the single equivalent moving dipole (SEMD) model of the instantaneous electrical activity using a single time sample of a non-invasive, multiple-lead body surface electrocardiograph (ECG) [5, 6]. Since the SEMD is a single dipole, it is an accurate and realistic representation at those times when cardiac electrical activity is well localized, for example when a wave of depolarization is emerging from the exit site of a re-entrant circuit [6]. We shall term the dipole resulting from activity at the exit site the *bioelectric dipole*.

RFA requires the accurate guidance of an ablation catheter to this exit site. If an ablation catheter is designed to deliver bipolar currents from electrodes at its tip (at a frequency above the range of biologically-generated signals), the resultant body-surface potentials may be used by the inverse algorithm to calculate the catheter tip location. Thus, the algorithm enables an operator to visually advance an image of the catheter tip on a graphical user interface towards the image of the selected bioelectric dipole for the accurate delivery of ablative energy.

The *forward problem* used by the inverse algorithm to estimate the torso surface potentials due to the equivalent source, assumes an infinite and homogenous volume conductor. This simplification of the volume conductor model allows the inverse algorithm solution to be calculated in real-time. However, the effect of ignoring all torso inhomogeneities, boundary effects and inaccuracies in electrode positions introduces a systematic error into the calculation; the position of the bioelectric dipole image seen by the cardiologist will be displaced from its *true* position by some error vector whose magnitude and direction are dependent on the specific nature of the non-idealities.

This error becomes important when we attempt to match the locations of the catheter tip and bioelectric dipole. If two dipoles are matched in *both* location and orientation, systematic error has been found to distort the two dipole images to locations within 3 mm of each other [6, 7]. Further work, however, has indicated that the images of two perfectly superposed yet *differently-oriented* dipoles may be greater than 5 mm apart in the presence of systematic error. This is far greater than the 2-3 mm accuracy required for ablation.

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This paper presents a method to compensate for this effect of dipole orientation in the presence of sources of systematic error. The method utilizes a special catheter tip design on which four electrodes are placed to produce three independent dipoles. If these dipoles are stimulated consecutively at the same location, they produce three sets of torso surface potentials. These torso potentials are weighted and summed to produce a set of potentials $\boldsymbol{\phi}_{new}$, such that the error between $\boldsymbol{\phi}_{new}$ and the measured potentials from the bioelectric dipole is minimized. When the catheter tip is superposed with the cardiac dipole, $\boldsymbol{\phi}_{new}$ is identical to the signal from the cardiac dipole to within the level of Gaussian noise, and therefore the dipole solutions will be equivalent regardless of physical tip orientation. Computer simulations will be used to test the method's ability to detect superposition of two dipoles in the presence of significant systematic error, and also its ability to guide a moving catheter tip dipole towards a fixed bioelectric dipole. This will provide an initial assessment of the method's ability to guide ablation therapy.

II. METHODS

A. *The inverse algorithm*

In the application of the inverse algorithm, voltage measurements are made at 60 or more electrodes on the torso surface over the course of a cardiac cycle. For a single time sample, the inverse algorithm calculates the parameters of the SEMD that would best reproduce the potentials recorded on the body surface. For a given dipole location, an unbounded volume conductor estimation is used to estimate the potential at the i^{th} body-surface electrode, $\phi_e{}^i$:

$$\phi_e{}^i = \frac{\vec{p}\cdot(\vec{r}-\vec{r}_i\,')}{4\pi g|\vec{r}-\vec{r}_i\,'|^3} \qquad (1)$$

where $\boldsymbol{r_i}$**'** represents the i^{th} electrode location, $\boldsymbol{r}$ the dipole location, $\boldsymbol{p}$ the dipole moment, and g the conductivity of the volume conductor. Due to the linear dependence of $\boldsymbol{\phi_e^i}$ on $\boldsymbol{p}$, the moment can be solved for analytically using the three-plus-three parameter optimization algorithm described by Armoundas et al [8]. An objective function, χ^2 per degree of freedom, describes how well the estimated dipole reproduces the measured voltages:

$$\chi^2/dof = \frac{1}{dof}\sum_{i=1}^{I}\left(\frac{\phi_e{}^i-\phi_m{}^i}{\sigma^i}\right)^2 \qquad (2)$$

where σ^i is the standard deviation of the Gaussian noise in lead i, I is the number of electrodes and $dof = I\text{-}6$. A 'brute force' search method is used to find the SEMD parameters that minimize the χ^2 function for a single time sample of the measured data.

B. *Catheter Design*

To compensate for the effect of dipole orientation, we propose a special catheter that can generate three independent dipoles at its tip. The three resulting dipole surface signals recorded at each electrode i, $\boldsymbol{\phi_k}^i = [\phi_1{}^i, \phi_2{}^i, \phi_3{}^i]^T$, may be weighted and summed to produce a voltage $\phi_h{}^i$, using a weighting vector, $\boldsymbol{\lambda}= [\lambda_1, \lambda_2, \lambda_3]$:

$$\phi_h{}^i = \vec{\lambda}\cdot\vec{\phi}_k{}^i = \frac{\vec{\lambda}\cdot P\cdot(\vec{r}-\vec{r}_i\,')}{4\pi g|\vec{r}-\vec{r}_i\,'|^3} \qquad (3)$$

where P is a matrix in which the rows are formed by the three dipole moments. $\phi_h{}^i$ is the signal that would have resulted if a dipole of orientation $\boldsymbol{h} = \boldsymbol{\lambda}\cdot P$ were generated at the catheter tip. By recording the three signals generated consecutively at the same catheter tip location by the three independent dipoles, and then choosing a single $\boldsymbol{\lambda}$ to be used for all I electrodes, we can reproduce the set of potentials that would have resulted from a dipole of orientation $\boldsymbol{h}$ placed at the catheter tip. This allows us to simulate the surface potentials of a dipole of any moment at the catheter tip, regardless of physical tip orientation.

Let us now assign the instantaneous surface potentials due to the bioelectric dipole to $\boldsymbol{\phi_h}=[\phi_h{}^1, \phi_h{}^2,\ldots, \phi_h{}^I]$ and the orientation of the bioelectric dipole image to $\boldsymbol{h}$. *If the catheter tip is perfectly superposed with the bioelectric dipole, a* $\boldsymbol{\lambda}$ *can be found to create a* $\boldsymbol{\phi_{new}}=[\phi_{new}{}^1, \phi_{new}{}^2,\ldots, \phi_{new}{}^I]$ *that is indistinguishable from* $\boldsymbol{\phi_h}$ (to within the random noise level). If a brute-force search is then applied to $\boldsymbol{\phi_{new}}$, the SEMD solution will be equivalent to that for $\boldsymbol{\phi_h}$.

C. *Finding the catheter tip location*

We define the error, E, as the sum over all body-surface electrodes of the difference between the potentials recorded at each surface electrode due to the bioelectric dipole, $\phi_h{}^i$, and a weighted sum, $\tilde{\phi}^i = \vec{\lambda}\cdot\vec{\phi}_k{}^i$, of the potentials recorded at each electrode due to the dipoles at the ablation catheter tip:

$$E = \sum_{i=1}^{I}\frac{\left(\phi_h{}^i-\tilde{\phi}^i\right)^2}{(\sigma^i)^2} \qquad (4)$$

For a set of three catheter signals expressed in the matrix $\phi_k=[\boldsymbol{\phi}_1, \boldsymbol{\phi}_2, \boldsymbol{\phi}_3]^T$ where $\boldsymbol{\phi}_1=[\phi_1{}^1, \phi_1{}^2,\ldots, \phi_1{}^I]^T$ etc. we find the $\boldsymbol{\lambda}$ that minimizes E, $\boldsymbol{\lambda}_{\min}$. The signal $\boldsymbol{\phi_{new}}$ is then:

$$\vec{\phi}_{new} = \vec{\lambda}_{\min}\cdot\phi_k \qquad (5)$$

There are then two ways to perform the brute force search to find the best-fit dipole model for $\boldsymbol{\phi_{new}}$: Method I, in which the moment of the SEMD solution is chosen by three-plus-three parameter optimization (both the dipole location and moment are optimized); and Method II in which the SEMD solution is restricted to a dipole of moment $\boldsymbol{h}$ regardless of catheter tip position (and therefore only the dipole location is optimized).

At catheter tip positions close to the bioelectric dipole, $\boldsymbol{\phi_{new}}$ is very similar to $\boldsymbol{\phi_{ch}}$, the signal that would be

recorded if a dipole of orientation $\boldsymbol{h}$ were placed at the catheter tip, but not identical. We hypothesize that Method I may fail to direct the final superposition of the catheter tip on the location of the bioelectric dipole, because the homogenous, unbounded model lacks the complexity to accurately fit two sets of surface potentials from slightly different locations in a non-ideal torso. The parameter χ^2 provides insight into the degree of systematic error, since its magnitude increases as the degree of 'mismatch' between the unbounded, homogenous model and the bounded, non-ideal torso increases. The higher the χ^2 value of the bioelectric dipole, χ_h^2, the larger the uncertainty in the SEMD estimation, the greater the possibility that Method I will fail, and the greater the distance from the bioelectric dipole that this is likely to happen.

If the dipole moment is fixed during the brute-force search (Method II), the method will only indicate superposition if the dipole images correspond in both location and moment. However, at distances far from the bioelectric dipole, ϕ_{new} bears little resemblance to ϕ_{ch}. Therefore, the position of the dipole calculated using Method II may be significantly displaced from the actual location of the catheter tip. As the catheter and bioelectric dipole images are brought closer and ϕ_{new} begins to resemble ϕ_{ch}, the position calculation will become more accurate.

Methods I and II appear to work best in complementary regimes, and we hypothesize that the ideal algorithm is a robust combination that most efficiently converges on the correct solution.

E. Combined Method

The Combined Method employs each Method in the regime in which it operates most effectively. The distance at which Method I errors might occur depends greatly on the nature of the systematic error and the location of the bioelectric dipole. Therefore, our approach is based on a comparison of χ_h^2 with a parameter that measures the fit of φ_{new} to the bioelectric dipole signal, φ_h. This is the minimum value of the parameter, E, defined in *Eqn 4*. If the minimum value of E, E_{min}, is less than χ_h^2, the homogenous, unbounded model does not adequately discriminate between φ_{new} and φ_h, and may not accurately calculate the distance between the cardiac and catheter dipoles. Hence, Method I may fail. For $E_{min} < \chi_h^2$, Method II is gradually introduced to a degree determined by the ratio of E_{min} to χ_h^2. When $E_{min} \geq \chi_h^2$, Method I is used.

A weighting function, $W(R)$, where $R = E/\chi_h^2$, is used to weight the contributions from Methods I and II to the final solution when $E_{min} < \chi_h^2$. We empirically chose a weighting function $W(R) = R^{\alpha}$ where α is a constant whose value we optimized at 0.415. The dipole image locations calculated by CSC Method I (r_{MethodI}) and Method II (r_{MethodII}) are weighted to produce a new dipole image location, r_{combine}:

$$r_{\text{combine}} = W \cdot r_{\text{Method I}} + (1\text{-}W) \cdot r_{\text{Method II}} \qquad (6)$$

F. Computer Simulations

Methods I and II and the Combined Method were tested in simulations of a homogenous spherical torso model of radius 12.5 cm, simulating boundary effects, inaccurate electrode positioning and Gaussian noise. 60 electrodes were distributed on the sphere surface in a 12.5 cm by 12.5 cm square pad of 25 electrodes (inter-electrode separation = 3.125 cm), with the other 35 electrodes distributed randomly over the spherical surface. The pad was centered at the point on the torso surface closest to the bioelectric dipole. This layout takes into consideration the eventual application of this work in RFA procedures. The bioelectric and catheter tip dipoles were placed randomly within a spherical 'heart compartment' of radius 4.5 cm, centered 8 cm from the torso center. The dipoles were of sufficient magnitude to generate maximum surface potentials on the order of 0.1 mV.

Systematic error in the form of both boundary effects and inaccurate electrode positioning was modeled to simulate measured data. Boundary effects were simulated using an equation derived by Frank [9] defining the potentials on the surface of a bounded spherical torso due to a dipole within the torso volume. Inaccurate electrode positions, on the other hand, were generated by adding errors drawn from a Gaussian distribution ($\mu = 0$, $\sigma_e = 0.5$ cm) to the *x-y-z* co-ordinates of each correct electrode location (this error is severe compared to the accuracy of technology currently available to acquire electrode positions). Finally, we added Gaussian white noise ($\mu = 0$, $\sigma_m = 0.01$ mV) to the measured surface potentials to account for a realistic level of measurement noise.

To compare the guidance capabilities of Methods I and II and the Combined Method, the catheter tip was steered towards the bioelectric dipole using only the positions of the dipole images as would be seen on a user interface by the cardiologist. For each method, the actual catheter tip was moved towards the bioelectric dipole by a fraction X of the residual inter-image distance vector. At superposition, the distance between the actual catheter tip and actual bioelectric dipole (the *end-point accuracy*) was documented for that Method. If the end-point accuracy was less than 1.5 mm, the number of steps required to reach this effective superposition was noted. 100 simulations were conducted for values of X ranging from one-eighth to the full inter-image distance, each time using different randomly-chosen bioelectric dipole and initial catheter tip locations, and different electrode layouts.

III. RESULTS

A. Image Convergence

The percent image convergence is high for all three methods and for all values of X. Method I and the Combined Method have image convergence rates of 98.75% averaged over all values of X. Method II has 98.25% image convergence.

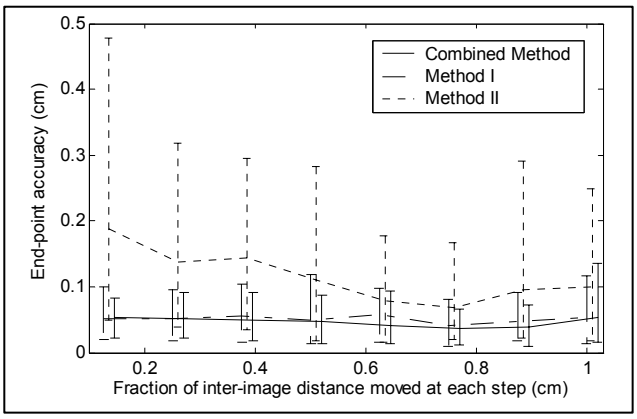

FIG 1: Mean and 95% confidence intervals for the end-point accuracy of each method vs. the fraction of the inter-image vector moved in real space by the catheter

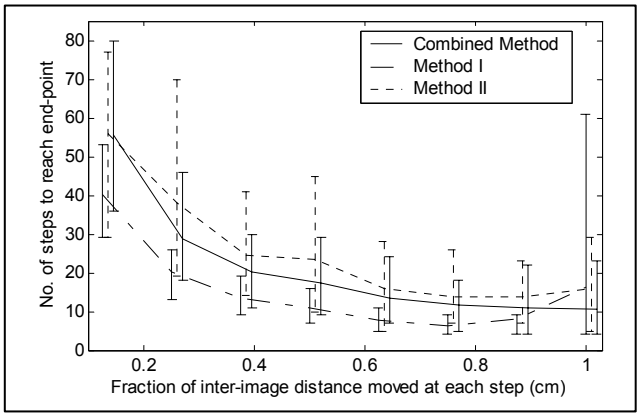

FIG 2:. Mean and 95% confidence intervals for the number of steps required by each method to reach the end-point vs. the fraction of the inter-image vector moved in real space by the catheter tip.

B. End-point Accuracy

The mean end-point accuracy for the Combined Method is slightly smaller than that of Method I for almost all values of *X,* due to distortion of the Method I results by one or more outliers greater than 1.5 mm (caused by dipole-model fitting errors). The Combined Method is not susceptible to these errors and offers the smallest end-point accuracy of the three methods (less than 1.5 mm with almost 100% confidence). Method II does not provide adequate accuracy as a guidance algorithm.

C. Number of Steps Taken

The number of steps required by Method I to achieve effective superposition is substantially lower than the Combined Method, except at $X = 1$ (see Fig. 2). This is due to the introduction of Method II into the calculation of $r_{combine}$; although the Combined Method will take longer to guide a catheter tip to the bioelectric dipole, it will also 'fine-tune' the relative dipole positions and prevent the significant errors allowed by Method I. Using either Method I or the Combined Method, we expect that the number of steps required to achieve effective superposition will not exceed 35 for the majority of step sizes. It is unlikely that the difference in the number of steps between the two methods would cause a significant difference in usability.

IV. DISCUSSION AND CONCLUSIONS

We have presented a method to compensate for the effect of dipole orientation in the presence of sources of systematic error such as boundary effects on the accuracy of the inverse algorithm. We tested three variants of the method (Methods I and II, and the Combined Method) in simulations of a homogenous, bounded spherical torso. Two forms of systematic error were used (boundary effects and inaccurate electrode positioning) in addition to 0.01 mV Gaussian noise. Using each of the three methods in turn, the virtual catheter was guided so as to reduce the distance between the images of the catheter tip and bioelectric dipoles on a user interface, with no knowledge of their actual locations.

Results indicate that the Combined Method and Method I are both excellent guidance algorithms for directing a catheter tip towards a bioelectric dipole. Their percent image convergence is close to 100%, their end-point accuracy is less than 1.5 mm with 95% confidence, and the number of catheter movements the cardiologist can expect to make is very reasonable. Given that dipole-fitting errors may occur using Method I guidance, the Combined Method is the slightly preferred option. This Method allows the inverse algorithm to accurately guide a simulated ablation catheter tip to the site of a bioelectric dipole. The rapidity and accuracy of this Method may allow radio-frequency ablation treatment to be administered both more accurately and to a much wider segment of the population affected by VT.

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