COMPUTER SIMULATION OF TRANSESOPHAGEAL PACING WITH CONVENTIONAL AND SELECTIVE LEADS USING 3-D MODELS

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Abstract: Design of a pacing lead for efficient and safe transesophageal pacing is of great importance because the lead configuration determines the current density distribution in the surrounding tissue. Consequently the lead configuration influences the pacing threshold (mA), esophageal thermal injury (if any), and unintentional stimulation of adjacent nerves and muscles. We present two types of transesophageal pacing leads – a conventional one with cylindrical electrodes and a selective lead where electrodes comprise only a part of cylinder's side surface. We analyzed, using 3-D geometrical models, the influence of interelectrode distance (from 2 cm to 10 cm) and central angle of the electrode (from 20° to 360°) on the current density distribution for the case of constant current stimulus of 20 mA. The results clearly show that electrodes with smaller central angle direct the current in the desired direction, but current density at electrode edges rises as well as the voltage between the electrodes. The effect of increased current density in front of the electrodes sharply decreases with radial distance from the lead axis. Increase of interelectrode distance has similar effect as the change from bipolar to monopolar stimulation. Implications of the results for *in vivo* situation regarding the pacing threshold and possible risk of esophageal burn injury were discussed.

Introduction

Transesophageal cardiac pacing is a temporary pacing technique used when other temporary pacing modes (transcutaneous, transvenous, transthoracic) are contraindicated, are too painful, or when they fail for some reason [1].

Numerous reports have confirmed that transesophageal atrial pacing is possible in almost all patients [2-8]. It can be used for prophylactic pacing during general anesthesia [6], to terminate supraventricular arrhythmias [7] and for many studies and diagnostic recordings in cardiac electrophysiology when heart rate has to be increased and precisely controlled [8].

Transesophageal ventricular pacing is more demanding. Due to less favorable position of ventricles to the esophagus than the atria, ventricular capture is harder to obtain, but it is possible [3] even in urgent situations [9]. The price is of course higher pacing threshold than for atrial pacing and possible unintentional stimulation of nerves and muscles. Patient discomfort occurs when higher stimulus currents are used. Hence minimizing pacing threshold is highly desirable [10].

In all clinical studies different bipolar or multielectrode leads were used, having cylindrical or frustum-like electrodes, which produce axially symmetric electric field. Clinicians usually overlook the fact that different electrode shape results in different electric field and current distribution in the tissue surrounding the lead. With the intention of obtaining selective and efficient transesophageal pacing, we proposed a new design of the pacing lead, which can produce directed electric field [11]. The selectivity of this type of leads was evaluated experimentally in a saline bath, and preliminary using a finite-element model [11, 12], but precise calculation of the electric field of those leads has not been performed.

What has been done in this respect is the theoretical analysis of current density distribution for conventional leads based on an analytical solution [8], which assumes that electrodes can be approximated as point sources and neglects actual electrode shape and edge effects. This analytical solution can be extended for spherical electrodes at any distance using method of images, but numerical methods are necessary for more complex electrode shapes like our angular electrodes.

The aim of this study is to simulate and analyze transesophageal pacing with the conventional, nonselective pacing lead and pacing with the selective lead using three-dimensional models. Emphasis is given to three-dimensional visualization of stimulation volume for different central angle of the electrodes and different interelectrode distance.

Materials and Methods

In order to analyze the influence of interelectrode distance and central angle of the electrode we created a precise 3-D model of the multielectrode transesophageal pacing lead (Figure 1). The lead comprises of 5 pairs of electrodes spaced at the distance *d*, having the central angle φ. Figure 1 b) and c) show lead dimensions and dimensions of each angular electrode. Minimal interelectrode distance was 20 mm, and maximal 100 mm. Central angle was varied from 20° to 120° in steps of 20°. We also analyzed angular electrodes with central angle 180° (half cylinder). For the purpose of comparison, central angle was also set to 360°, which corresponds to conventional non-selective lead.

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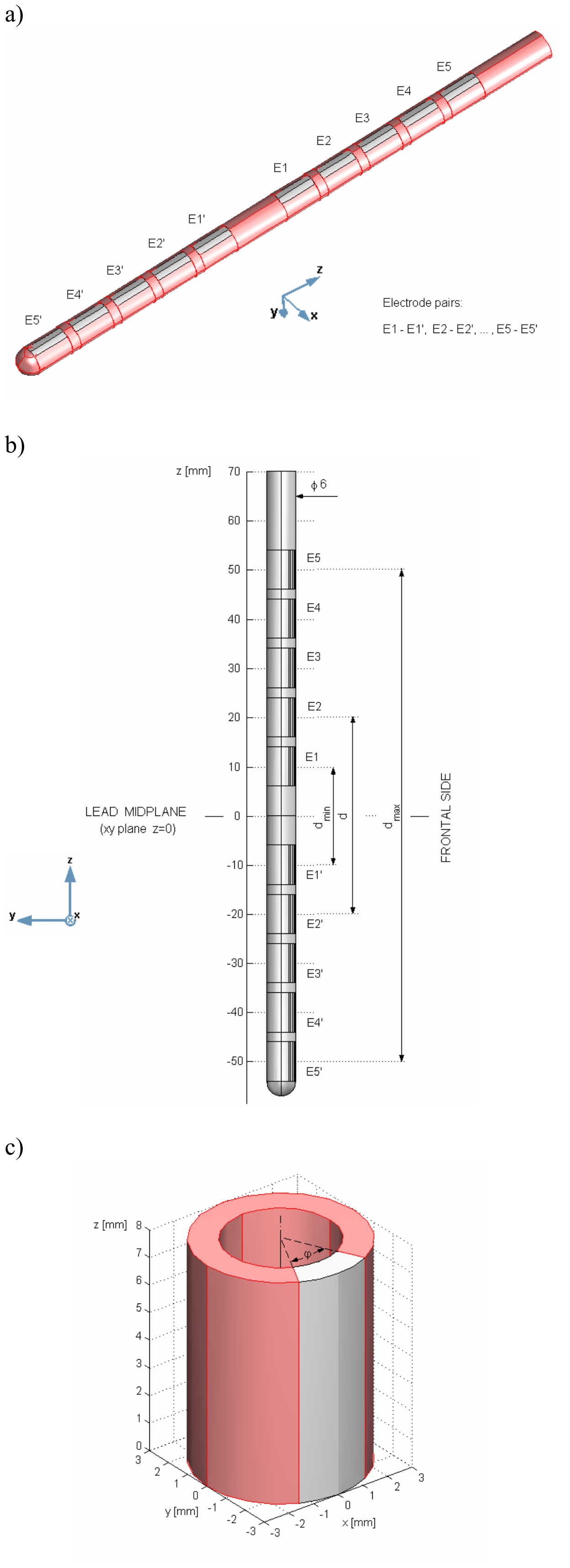

Figure 1: Selective multipolar transesophageal pacing lead. a) 3-D view of the lead; b) lead dimensions; c) dimensions of the angular electrode. d - interelectrode distance, φ - central angle.

Assuming the static conditions, the electric potential distribution associated with the stimulation pulses is governed by an elliptic partial differential equation:

$$-\nabla \cdot (\sigma \nabla u) = 0 \tag{1}$$

subjected to appropriate boundary conditions. Here u is the electric potential and σ is the electrical conductivity. The electric field distribution is then obtained as:

$$\vec{E} = -\nabla u \tag{2}$$

and the current density distribution is given by:

$$\vec{J} = \sigma \vec{E} \tag{3}$$

As we are primarily focused on the constructional aspects of the lead, in our model we positioned the lead in the center of a large cylinder (diameter 240 mm, height 240 mm) having electrical conductivity 0.3 S/m. This value was chosen to be close to the average isotropic conductivity of the myocardium [13].

We used finite-element tool FEMLAB 3 (Comsol, Sweden) for geometry modeling, meshing and solving and MATLAB 6.5 (The MathWorks, USA) for postprocessing. For symmetry reasons we modeled only one forth of the entire geometry ($x \geq 0$, $z \geq 0$, Figure 1). To reduce the error we meshed the geometry with due caution using second order tetrahedral elements. The size of elements near the lead was very small, and then it gradually increased with the distance from the lead axis. The mesh consisted of 20959 nodes and 97695 elements. The number of degrees of freedom was 148468, much larger than the number of nodes, since quadratic elements were used. The same mesh was used for all configurations of the lead. The change of the lead configuration was accomplished by changing the boundary conditions. On the active electrode segment (metallic conductor) we applied Dirichlet boundary condition:

$$u = U_0 / 2 \tag{4}$$

where U_0 is the voltage between the electrodes. On the lead midplane we used an antisymmetry boundary condition:

$$u = 0 \tag{5}$$

Dirichlet boundary condition assigns values of the electric potential to the boundaries. On all other boundaries we applied Neumman boundary conditions:

$$-\vec{n} \cdot \vec{J} = 0 \tag{6}$$

where $\vec{n}$ is the surface normal. This boundary condition specifies a zero current flux, equivalent to an insulating boundary, or a symmetry boundary condition. We sought the solution of linear system generated by the equation (1) and boundary conditions (4)-(6) by using the iterative GMRES solver with AMG preconditioner.

We compared all lead configurations for the case of constant current stimulus of $I_0 = 20$ mA . Therefore we adjusted the electrode voltage U_0 accordingly.

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We calculated the current density along the y-axis ($x = 0$) in the transversal midplane between the electrodes ($z = 0$) and in the transversal plane passing through the middle of the angular electrode ($z = d/2$ i.e. 10 mm, 20 mm, …) (see Figure 1 b).

Additionally, we included in the comparisons of current densities for different lead configurations, the current density obtained by the analytical solution for the lead comprised of two spherical electrodes having the diameter much smaller than their center-to-center distance. In this case spheres can be replaced by point sources. From the basic principles we derived that the current density in the lead midplane resulting from the point current sources + I_0 and –I_0 spaced at the distance d is given by:

$$\left|\vec{J}_{(r,\, z=0)}\right| = \frac{I_0}{4\pi} \frac{d}{\left[r^2 + (d/2)^2 \right]^{3/2}} \tag{7}$$

where r is the radial distance from the lead axis. In the transversal plane passing through the center of the spherical electrodes, the current density is given by:

$$\left|\vec{J}_{(r,\, z=\pm d/2)}\right| = \frac{I_0}{4\pi} \left\{ \left[\frac{1}{r^2} - \frac{r}{(r^2+d^2)^{3/2}} \right]^2 + \frac{d^2}{(r^2+d^2)^3} \right\}^{1/2} \tag{8}$$

For all configurations of the esophageal lead we also calculated the lead resistance:

$$R = \frac{U_0}{I_0} \tag{9}$$

In order to get a three-dimensional impression of the volume that would be stimulated with different configurations of the lead we calculated and visualized the current density isosurfaces 0.5 mA/cm^2, 5 mA/cm^2 and 25 mA/cm^2. The current density level 5 mA/cm^2 was deliberately chosen to correspond to the pacing threshold for the heart [13-15]. Other two levels were chosen to be ten times lower (0.5 mA/cm^2) and five times higher (25 mA/cm^2) than the pacing threshold. It is therefore likely that the volume encircled by the 5 mA/cm^2 isosurface will be stimulated, the volume outside the 0.5 mA/cm^2 is unlikely to be stimulated at all, and the volume encircled by the 25 mA/cm^2 isosurface will be at much higher risk of burn injury caused by the Joule heating.

Results

Figure 2 and 3 show the influence of interelectrode distance and central angle of the electrode on the current density both in the lead miplane (a) and in the plane at the electrode level (b). Vertical dash-dot line represents the lead axis. Negative y values correspond to the frontal side of the lead (see Figure 1).

Figure 4 shows 3-D visualization of 0.5 mA/cm^2, 5 mA/cm^2 and 25 mA/cm^2 current density isosurfaces for different interelectrode distance - central angle pairs.

Figure 5 compares the FEM solutions with the analytical results for the point sources (eqns. (7)-(8)).

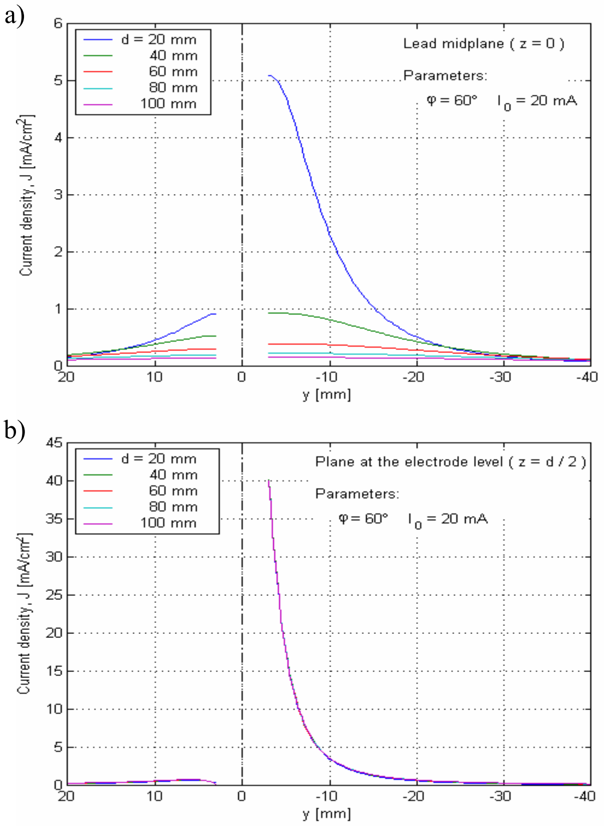

Figure 2: Influence of the interelectrode distance d on the current density. a) lead midplane b) plane at the electrode level.

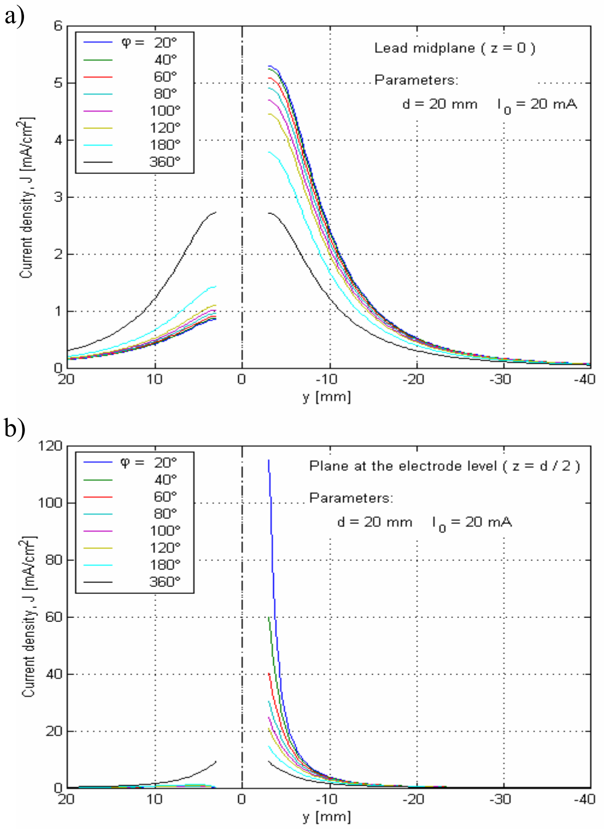

Figure 3: Influence of the central angle φ on the current density. a) lead midplane b) plane at the electrode level.

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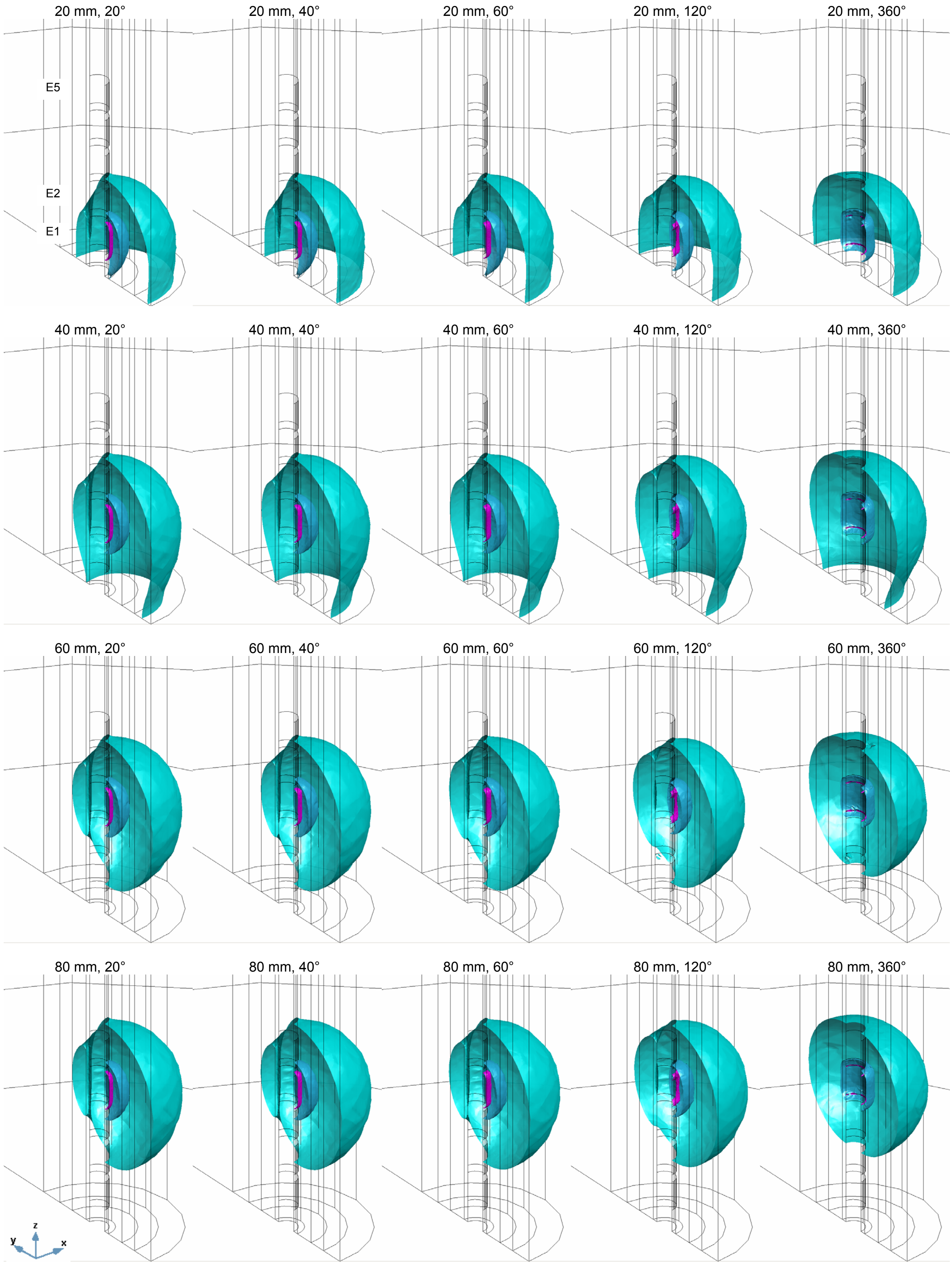

Figure 4: Current density isosurfaces 0.5 mA/cm^2 (cyan), 5 mA/cm^2 (pale blue) and 25 mA/cm^2 (magenta) for different interelectrode distance d (20-80 mm) and central angle of the electrode φ (20°-360°) resulting from a constant current stimulus $I_0 = 20$ mA (d, φ pair above each plot). All plots are symmetric around xy plane $z = 0$ and yz plane $x = 0$. Radius of the auxiliary circular arcs drawn in the lead midplane: 3, 4.5, 10, 15, 22, 72, 120 mm.

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Table 1: Lead resistance R for different interelectrode distance d (20-100 mm) and central angle of the electrode φ (20°-360°). The electrical conductivity of the medium surrounding the lead was 0.3 S/m.

Lead resistance R [Ω]		Interelectrode distance, d				
		20 mm	40 mm	60 mm	80 mm	100 mm
Central angle, φ	20°	614	635	638	640	643
	40°	465	479	485	487	490
	60°	380	396	400	403	404
	80°	323	339	341	344	346
	100°	282	297	302	304	306
	120°	251	265	271	273	275
	180°	189	203	208	211	213
	360°	126	139	144	147	149

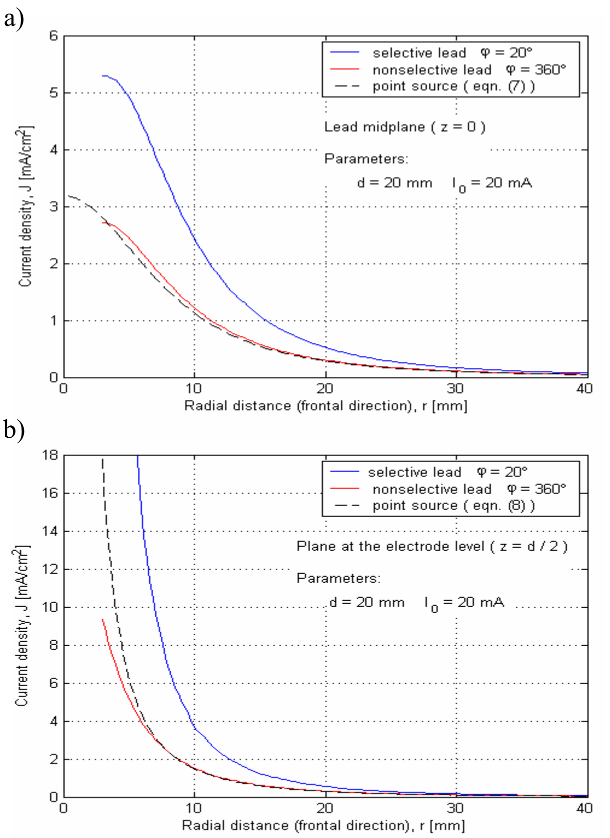

Figure 5: Comparison of FEM and analytical solution. a) lead midplane b) plane at the electrode level.

Table 1 summarizes the lead resistance calculated for different interelectrode distance and electrode central angle.

Discussion

In order to stimulate the heart it is necessary to set the stimulus current high enough such that the current density is above the stimulation threshold. At the same time it is desirable to minimize the stimulus current to avoid pain, hick-ups and other unpleasant effects associated with unintentional stimulation of adjacent nerves and muscles. Another important reason for minimizing the stimulus current is the risk of esophageal burns, especially if pacing is performed for a longer (> 30 min) period of time.

One way of minimizing the stimulus current is positioning the pacing lead in the optimal position. This has been extensively studied experimentally, but there is still no definite agreement between different research groups on whether P-wave recording is necessary or not for setting the optimal lead insertion depth [8, 10, 16]. Another way of minimizing the stimulus current, that we proposed, is the geometrical modification of the lead, specifically the electrodes in a way that the electric field is directed more towards the heart and rejected in the opposite direction.

In this study we focused on analyzing the influence of interelectrode distance and electrode central angle on the performance of the lead. For a given electrode central angle, the increase of interelectrode distance results in a large decrease of the current density in the lead midplane in the vicinity of the lead (Figure 2 a). For larger radial distances curves intersect meaning that the current density for larger distances falls less steeply, but is too small for stimulation. To achieve the stimulation for larger radial distances (i.e. deeper structures) it would be necessary to increase the stimulus current. In the transversal plane at the electrode level current density is, as expected, much higher than in the lead midplane and is insensitive to the change of interelectrode distance (Figure 2 b). For a given interelectrode distance, a decrease of central angle improves the selectivity of the lead (Figure 3), but this positive effect fades for larger radial distances, since the current density falls steeply. In the lead midplane (Figure 3 a) the increase of current density in the frontal direction tends to saturate for smaller central angles (i.e. the change of J for the change of φ from 120° to 100° is larger than the change of J corresponding to the change of φ from 40° to 20°). It is also important to notice that for smaller central angle the current density in the plane at the electrode level increases significantly (Figure 3 b). All this points to the fact that for electrodes with smaller central angle the contribution to the increase of current density in the heart is much smaller than the increase of the risk associated with the esophageal burn injury.

Figure 4 further supports previous observations. We can see that the decrease of central angle increases the penetration of the current field in the frontal direction, as well as that the increase of interelectrode distance has the effect similar to change from bipolar to monopolar stimulation.

Comparison of numerical (FEM) and analytical solution (Figure 5) shows good agreement for the conventional lead with cylindrical electrodes ($\varphi = 360°$), especially for larger radial distances. For a selective lead ($\varphi = 20°$) discrepancy is large.

Lead resistance rises for smaller central angles and larger interelectrode distance (Table 1). This parameter is important for the design of the output stage of the esophageal stimulator. For example, to deliver 20 mA output current, output stage voltage must be able to rise to 12.3 V for (20°, 20 mm) lead, which is almost five times higher voltage than needed for the lead with conventional cylindrical electrodes (360°, 20 mm).

Finally, we point out that many simplifications call for further research (e.g. complex anatomy of the thorax having inhomogeneous conductivity distribution; myocardium is anisotropic; stimulation thresholds are also anisotropic [14]; cardiac sensitivity to electrical stimulation exhibits complex dependence on many factors [15]; esophagus has lower conductivity than the myocardium resulting in an abrupt jump of the current density at the boundary; etc.).

Conclusions

This study presents a comprehensive evaluation of selective transesophageal pacing leads and compares them with conventional non-selective lead. Computer simulations using 3-D finite element model can help optimize electrode central angle and interelectrode spacing of the lead for targeting specific area of the heart.

Results of our simulations bring important practical recommendations for the lead design for more selective and less painful transesophagel pacing. The use of selective instead of conventional electrodes is beneficial, but decreasing the central angle too much, considerably increases the current density in the vicinity of the lead (mainly within the esophageal wall) and has no effect on the current density more distant from the lead (i.e. within the heart where we would like to get higher current density). Thus, the use of electrodes with small central angle inevitably increases the risk of esophageal burn injury. The increase of interelectrode spacing as a method of stimulating more distant structures (i.e. ventricles) is not possible without considerably increasing the stimulus currents. However, this is accompanied by higher current density in the esophagus and increased risk of esophageal burns.

Future work should include evaluation of transesophageal pacing using three-dimensional anatomical thorax model.

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