Effect of anisotropy in myocardial electrical conductivity on lesion characteristics during radiofrequency cardiac ablation: a numerical study

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ABSTRACT

Background: Traditional computer simulation studies of radiofrequency catheter ablation (RFCA) usually neglect the anisotropy in myocardial electrical conductivity (MEC), which is likely an essential factor in governing the ablation outcome. Here, a numerical study of lesion characteristics during RFCA based on an anatomy-based model incorporating fiber orientation was performed to investigate the anisotropy in MEC.

Methods: A three-dimensional thorax model including atria, blood, connective tissue, muscle, fat, and skin was constructed. The myocardial fiber was established through a rule-based method (RBM) based on the anatomical structure of the heart. The anisotropic MEC were 0.40 and 0.28 S m⁻¹ in longitudinal and transverse directions, respectively. The ablation result was compared with the isotropic scenario where the isotropic MEC was the average of the anisotropic conductivities as 0.34 S m⁻¹.

Results: The complexity of fiber architecture varied with that of the local anatomical structure. At RF power of 20 W for 30 s, the tissue temperature and lesion volume were reduced by 2.8 ± 0.1% and 6.9 ± 0.5%, respectively, under anisotropic MEC around the ostium of the pulmonary vein and left atrial appendage. Those for the posterior wall and roof of the left atrium, and the inside of the superior vena cava were 1.9 ± 0.3% and 5.6 ± 1.2%, respectively.

Conclusions: Anisotropy in MEC has a greater reduction effect on lesion volume than on tissue temperature during RFCA; this effect tends to be restrained at positions with more uniform fiber distributions and can be enhanced where significant variation in fiber architecture occurred.

1. Introduction

Atrial fibrillation (AF) is the most prevalent form of cardiac arrhythmia; it has significant morbidity and mortality and affects more than 30 million individuals worldwide [1–3]. AF causes irregular heartbeat and is a risk factor for various associated complications, for example, stroke, heart failure, and coronary syndrome [4]. Radiofrequency catheter ablation (RFCA) is a safe, promising medical modality that eases or eliminates AF-induced problems through RF energy application. An ablation catheter is delivered percutaneously to the target site and RF current is released through an electrode fitted to the catheter tip; in this way, irreversible cardiac tissue necrosis is achieved [5]. To date, numerous computational modeling-based studies have analyzed RFCA characteristics during AF ablation using the finite element method to aid clinical operation and theoretical understanding of AF treatment [6]. However, modeling that fully reflects realistic ablation procedures has not been achieved, as the material geometric constructions and physical properties are usually simplified.

Cardiac muscle is composed of myocardial fibers, which dominantly the contraction of the heart [7]. Fiber orientation has been reported to affect the myocardial electrical conductivity (MEC) such that the electrical current flows preferentially along the fiber direction [8], causing the electrical conductivity throughout the heart to become anisotropic. This phenomenon influences the current distribution and thus the lesion formation of RFCA. In ablation simulation, however, to simplify the model, MEC is usually assumed to be isotropic [6]. As the electrical conductivity governs the ablation result [9], appropriate fiber orientation is essential for good accuracy. The fiber orientation can be determined using an imaging system; however, this process is time-consuming and has limited robustness [10]. Application of a rule-based method (RBM) that mathematically describes the fiber orientation based on historical observations is a more acceptable alternative, as it is more feasible for computational simulation [11–13].

In this study, an anatomy-based atrium model incorporating RBM-based myocardial fiber orientation is developed in...
order to simulate anisotropic cardiac tissue. The effect of anisotropy in MEC on ablation outcome and lesion characteristics is investigated and the result is compared with the isotropic case, that is, the traditional simplification employed in RFCA simulation.

2. Methods

2.1. Construction of the ablation model with fiber orientation

2.1.1. Geometry of the thorax model

The ablation model was constructed using COMSOL Multiphysics 5.4 (COMSOL, Burlington, MA, USA). It consisted of a heart model situated in a portion of the thorax, which was represented as an elliptic cylinder and stratified into four layers: connective tissue, muscle, subcutaneous fat, and skin, from the center to the outer boundary (Figure 1). The chamber inside the heart was defined as the blood domain. The muscle, subcutaneous fat, and skin thicknesses (including both epidermis and dermis) were 18 [18], 6 [19], and 5.7 mm [20], respectively. The thorax-component height X, width Y, and length Z were 250 [14], 400 [15], and 295 mm [16], respectively. The traditional unipolar ablation was configured by attaching a section of the ablation catheter to the heart and defining a grounding pad, which was represented by a rectangular area of 14 × 10 cm [17] on the back. Six ablation locations were determined based on current RF ablation strategies [21,22]. An electrode tip model with a 7-Fr diameter and 4-mm length [23] was added to penetrate the

![Figure 1](image_url). Construction of ablation model. The height X, width Y and length Z of the thorax model were 250 [14], 400 [15] and 295 [16] mm, respectively. To reduce computational complexity, the length Z was determined by the distance from the chest to the upper arm, including the entire heart. The grounding pad was simulated as a rectangular area of 14 (a) × 10 (b) mm [17]. The active electrode penetrated the myocardium to ensure an insertion surface area of 7.85 ± 5% mm² and volume of 1.15 ± 5% mm³.
target myocardium with a surface area and volume of insertion within $7.85 \pm 5\% \text{ mm}^2$ and $1.15 \pm 5\% \text{ mm}^3$, respectively. These measurements can be automatically calculated by COMSOL. Therefore, they were chosen as the two reference criteria to ensure consistent electrode insertion for all ablation sites, as it is difficult to measure and maintain a fixed insertion depth or angle in all cases because of the anisotropic geometry of the heart. The insertion was adjusted to a position as perpendicular to the target site as possible.

As atria constitute the main AF ablation target site, the heart model we adopted was a human atrium model (without ventricular chambers) contributed by the University of Nijmegen as an open-access resource [24]. The geometry of the atrium had already been extracted from MRI images and formatted as a MATLAB file [25] containing mesh information (node locations and corresponding face elements). The node and face information were then collected to recover the geometry that could be imported into COMSOL. The original atrium model left several holes at the locations of four pulmonary veins (PV), superior and inferior venae cavae (SVC/IVC), and mitral and tricuspid valves. (Figure 2(a,b)). Consequently, the domains of connective tissue and blood were unified through these holes and could not be distinguished by COMSOL during the definition of different materials in the ablation model. Therefore, we manually sealed the atrium model and reconstructed the structures of possible ablation targets as PVs and two kinds of vena cava (Figure 2(c,d)).

In previous simulation studies, convergence tests were conducted to determine specific model dimensions and to avoid boundary effects [26,27]; however, we deemed such a test unnecessary here, as the region of interest (ROI) in which ablation occurred was not a simple fragment of myocardium and blood.

2.1.2. Ablation site selection

Figure 3 shows six ablation points selected according to different RFCA strategies for AF treatment: circumferential pulmonary vein isolation (PVI), linear ablation [21], and superior vena cava isolation (SVCI) [22]. Positions 1–5 were located on the PVI pathway. Among them, positions 1, 3, and 5 were...
selected based on novel wide antral PVI. For those cases, the electrode was placed $>1.5$ cm from the PV ostium [28]. Position 1 is located near the left atrial appendage (LAA) ostium, which is beneath the left superior pulmonary vein (LSPV) at the left atrium (LA) anterior side. Position 2 is the traditional ablation site for ostial PVI at the LSPV. Position 3 is on the roofline between the LSPV and right superior pulmonary vein (RSPV); this is an option for linear ablation. Positions 4 and 5 are on the LA posterior wall (PW), where complex fractionated atrial electrograms (CFAEs) may occur; these are a sign of the AF electrophysiological substrate [21]. Position 6 is the only SVC target site in the right atrium (RA). This is the most typical non-PV ectopic focus for AF, and ablation at this site encircles the region approximately 5–10 mm above the cavoatrial junction [22].

### 2.1.3. Development of fiber orientation

To define the anisotropy in electrical conductivity, fiber architecture was generated based on the atrium model. First, the surface node numbers of both the epicardium and endocardium of the atrium model were acquired [29]. Then, the transmural direction of each node was identified in MATLAB as the first step toward generating the fiber orientation. The endocardial-node transmural direction was toward the nearest epicardial node. For the myocardial and epicardial nodes, the transmural direction was consistent with the nearest endocardial node. By rotating the transmural direction $90^\circ$ around the $z$-axis in the anticlockwise direction, the circumferential direction of a given node was obtained.

Considering the complex atrial anatomy, a structure tensor analysis was conducted to smooth the fiber structure. The structure tensor $S$ was defined as follows:

$$ S = G_\sigma \times \left[ \nabla f \times (\nabla f)^T \right] = G_\sigma \times \begin{bmatrix} f_x^2 & f_x f_y & f_x f_z \\ f_x f_y & f_y^2 & f_y f_z \\ f_x f_z & f_y f_z & f_z^2 \end{bmatrix} $$ (1)

where $\nabla f$ is the aforementioned circumferential direction, representing the local gradient of a node in the finite element model (FEM) with the three-dimensional coordinate system $(x, y, z)$, and $G_\sigma$ is a Gaussian filter expressed as

$$ G_\sigma(x, y, z) = \left( \frac{1}{2\pi\sigma^3} \right) e^{-\frac{x^2+y^2+z^2}{2\sigma^2}} $$ (2)

where $\sigma = 0.3$ is the standard deviation of the Gaussian distribution.

Finally, the eigenvalues $(\lambda_1, \lambda_2, \lambda_3)$ of Equation (1) were calculated, and the eigenvector corresponding to the largest eigenvalue denoted the fiber orientation of a given node.
The atrium-model fiber distribution is shown in Figure 4, with detailed displays of the orientations at the six target sites. Taking the OX axis as a reference direction, a diagonal fiber orientation was obtained for position 1 and the majority of position 4. At the PW, the slope angle between the fiber and the OX axis decreased from the inferior to the superior side and, simultaneously, the fiber gradually pointed upward along the OZ axis. The fiber orientation at the inferior wall (IW) boundary was almost parallel to the OX axis (position 5). The fiber orientation at the LA roof was directed upward, but rotated slightly right and left on the posterior and anterior sides of the heart, respectively (position 3). The fiber orientation for the veins was distributed annularly along with the geometric structure (positions 2 and 6).

2.2. Simulation study

2.2.1. Governing equation

Heat transfer in biological tissue is commonly expressed by the Pennes bioheat transfer equation as follows [18]:

$$\rho c_o \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) - Q_p + Q_m + Q_h$$

(3)

where $\rho$, $c_o$, $T$, and $k$ are the density (kg m$^{-3}$), specific heat at constant pressure (J kg$^{-1}$ K$^{-1}$), temperature (K), and thermal conductivity (W m$^{-1}$ K$^{-1}$), respectively, of the tissue. Further, $Q_p$, $Q_m$, and $Q_h$ are the heat loss due to blood perfusion (W m$^{-3}$), the volumetric heat produced by the metabolism (W m$^{-3}$), and the RF current-generated heat source (W m$^{-3}$), respectively. Generally, $Q_p$ and $Q_m$ are ignored because of their insignificant contributions to cardiac ablation compared with the other terms in the Pennes equation. In the cardiac-ablation RF range (~500 kHz), biological tissue can be considered as a total resistive medium. Here, $Q_h$ refers to the resistive heating in the tissue, described by the quasistatic electrical relation

$$Q_h = \sigma_m |E|^2$$

(4)

Here, $\sigma_m$ is the tissue electrical conductivity (S m$^{-1}$) and $E = -\nabla V$ is the electric field intensity (V m$^{-1}$), calculated from the gradient of the associated electric potential $V$ (V). If no internal electric sources exist in the heat transfer system, $V$ satisfies

$$\nabla \cdot \sigma_m \nabla V = 0$$

(5)

In addition, $\sigma_m$ can be considered as the conductivity tensor of each element in the FEM [30], with

$$\sigma_m = \sigma_l + (\sigma_t - \sigma_l)AA^T$$

(6)

where $\sigma_t$, $\sigma_l$, $I$, and $A$ are the myocardial transverse conductivity (perpendicular to the myocardial fiber), longitudinal conductivity (parallel to the myocardial fiber), unit matrix, and direction cosine of the fiber orientation of each element (acquired previously), respectively. Therefore, fiber orientation is substantially reflected by an anisotropic distribution of electrical conductivity throughout the entire heart.

2.2.2. Material properties and boundary conditions

The electrical and thermal properties of the model materials are listed in Table 1 [18,26,27,31–36]. The myocardial electrical and thermal conductivities were assessed at an ambient temperature of 37°C. These two conductivities are temperature-dependent and obey unique piecewise relations [26,27]. The electrical conductivity increased exponentially at a rate of 1.5%/°C below 100°C and then decreased by four orders of magnitude between 100 and 105°C. The thermal conductivity increased linearly at a rate of 0.12%/°C up to 100°C and remained constant thereafter. The electrical conductivity of the connective tissue $\sigma_c$ was adjusted to constrict the global impedance of the model within the range of clinical transthoracic impedance [37]. The MEC in the longitudinal and transverse directions under the anisotropic condition (AC) were determined from two previous experimental studies [34,35]. The longitudinal and transverse
Electrical conductivities were 0.40 and 0.28 S m\(^{-1}\), respectively. The MEC under the isotropic condition (IC) was obtained by averaging the two conductivities under AC (0.34 S m\(^{-1}\)) [36].

Figure 5 demonstrates the electrical and thermal boundary conditions. The electrode was applied with a constant power of 20 W and the grounding pad was applied with 0 V. The remaining model surfaces were described by the Neumann boundary condition (i.e., null electric flux). A constant temperature of 37 \(^\circ\)C was applied to all outer boundaries of the model; this was also the initial temperature of the entire model.

In order to concentrate on the anisotropic effect of MEC on lesion size, fluid dynamics inside the atria were neglected to avoid the shift of temperature distribution and thus the lesion morphology in the myocardium during ablation [23]. The blood-flow cooling effect was simulated by applying two convection coefficients at the myocardium–blood (ht = 708 W m\(^{-2}\) K) and electrode–blood (he = 3636 W m\(^{-2}\) K) interfaces, which were calculated under a medium flow condition of 10.3 cm s\(^{-1}\) [18]. Because of the exclusion of fluid dynamics, the saline irrigation of the electrode was ruled out.

### 3. Results

The electrode–tissue interface was meshed with the finest element size set to 0.1 mm. The simulation time and time steps were 30 s and 0.1 s, respectively. The entire model contained approximately 330,000 tetrahedral elements. The connective-tissue electrical conductivity \(\sigma_c\) was fixed to 0.1886 S m\(^{-1}\) for all the cases of simulations. As a result, the model impedance under different configurations of ablation sites varied from 90 to 120 Ω.

#### 3.1. Ablation performance

We computed the maximum myocardial temperature and effective lesion volume trends at each ablation site for a 30-s RFCA process under AC and IC, as shown in Figures 6 and 7, respectively. An effective lesion was created by a temperature field above 50 \(^\circ\)C [26,27]. The average temperature and lesion volume under IC were 69.3 ± 1.6 \(^\circ\)C and 90.27 ± 24.42 mm\(^3\), respectively. Those for AC were 67.8 ± 1.6 \(^\circ\)C and 84.72 ± 24.08 mm\(^3\), respectively. Higher temperatures and effective lesion volumes were consistently obtained for IC. Table 2 lists the relative changes in the maximum tissue temperature and effective lesion volume. The tissue temperature and effective lesion volume after 30 s of RFCA were reduced by nearly 3% and 10%, respectively when comparing AC and IC. The extent of variation at positions 1 and 2 was relatively larger than at the other positions. Further, these two sites were not effective in growing lesion volume, especially under AC. The average changes in tissue temperature and lesion volume at positions 1 and 2 were 2.8 ± 0.1% and 7.0 ± 0.5%, respectively; the average changes for the other four positions were 1.9 ± 0.3% and 5.6 ± 1.2%, respectively. The larger standard deviations in the results of these four positions were caused by position 6 where the smallest temperature and volume differences were found.

Figure 8(a) compares the maximum epicardial and myocardial temperatures under AC and IC for all positions. The average epicardial temperatures under IC were approximately 0.9 \(^\circ\)C higher than those under AC. Transmural lesions were achieved at all positions; however, at position 2, the transmural effect was weak under both AC and IC, as only an insignificant excess above 50 \(^\circ\)C was reached. The epicardial temperature was reduced compared to the myocardial temperature. Comparatively small decreases in temperature were observed at positions 3, 5 and 6. For position 3, minimum differences of 1.6 and 2.1 \(^\circ\)C between the epicardial and myocardial temperatures were observed under AC and IC.

### Table 1. Thermal and electrical properties of the materials used in the ablation model.

| Material          | \(\rho\) (kg m\(^{-3}\)) | \(c\) (J kg\(^{-1}\) K\(^{-1}\)) | \(k\) (W m\(^{-1}\) K\(^{-1}\)) | \(\sigma\) (S m\(^{-1}\)) | References |
|-------------------|--------------------------|---------------------------------|------------------------------|---------------------------|------------|
| Electrode         | 21,500                   | 132                             | 71                           | \(4.6 \times 10^6\)       | [26,27]    |
| Blood             | 1000                     | 4180                            | 0.540                        | 0.990                     |            |
| Connective tissue | 1000                     | 3200                            | 0.400                        | \(\sigma_c\)              | [31]       |
| Muscle            | 1090                     | 3421                            | 0.490                        | 0.464                     | [18]       |
| Fat               | 900                      | 2222                            | 0.200                        | 0.020                     | [31]       |
| Skin              | 1200                     | 3460                            | 0.345                        | 0.200                     | [32,33]    |
| Myocardium        |                          |                                 |                              |                           |            |
| Anisotropic       | 1200                     | 3200                            | 0.530                        | 0.40 \(\text{longitudinal}\) | [31,34,35] |
|                   |                          |                                 |                              | 0.28 \(\text{transverse}\) |            |

The electrical conductivity of the connective tissue \(\sigma_c\) for all cases of simulations was constant and was determined to match the range of clinical transthoracic impedance [37].

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**Figure 5.** Electrical and thermal boundary conditions of ablation model in posterior view.
Figure 8(b) shows the lesion formation in the myocardium under the two conditions at position 3. The differences in the maximum lesion width and depth between AC and IC were 3.1% and 2.7%, respectively.

According to the characteristics of lesion formation, we identified three types of temperature distributions on the myocardial surface as shown in positions 2, 4 and 6 in Figure 9. The field proximal to the electrode boundary had a lower temperature; this phenomenon was especially apparent at position 2. This result matched the findings of Figures 6–8 that the lowest tissue temperature and lesion volume in the myocardium, as well as the lowest epicardial temperature, were achieved at position 2. At this site, there was a marked transition and gap of lesion formation along the LSPV and atrial wall. As shown in Figure 9, a minor increase in tissue temperature and lesion size was found in IC, especially at positions 4 and 6. From the comparison between the lesion boundary under AC and IC, a slight effect of preference was observed in the lesion growth under AC. At position 2, lesions tended to spread laterally and downward in the PV and atrial wall, respectively. The lesion growth at position 4 was impeded along the bottom-right direction of the electrode, thus demonstrating a diagonal spreading trend. This characteristic was also shared by positions 1 and 3, at which a myocardial-fiber orientation shift occurred. For horizontally oriented myocardial fibers, as in positions 5 and 6, more effective lesions were generated along with the longitudinal orientation. At position 6, the temperature distribution had a
wider vertical range, higher temperatures were generated along the lower boundary of electrode-myocardium contact. This observation agreed with the current flow at position 6, as shown in Figure 10, in which a large amount of current flowed downwards from the electrode. All current was spatially transmitted and eventually pointed to the grounding pad.

The current density distribution on the myocardial surface at positions 2, 4 and 6 are shown in Figure 11. There was a clear similarity in the area of current density distribution under AC and IC. The main difference among these two conditions was caused by areas of higher current density (pink and red areas in Figure 11), more of which was generated under IC. The highest current density was found at position 2, especially along the vertical direction at the electrode-myocardium contact boundary. This phenomenon was also discerned at position 6. In contrast, a relatively uniform distribution of higher current density occurred at position 4. For all cases, the lesion shape, as illustrated in Figure 9, can be

| Positions | Tani (°C) | Tiso (°C) | ΔT (%) | Vani (mm³) | Viso (mm³) | ΔT (%) |
|-----------|-----------|-----------|--------|------------|------------|--------|
| 1         | 67.2      | 69.1      | 2.8    | 67.08      | 73.38      | 9.4    |
| 2         | 65.6      | 67.4      | 2.7    | 61.93      | 68.16      | 10.1   |
| 3         | 67.1      | 68.4      | 1.9    | 73.80      | 78.04      | 5.7    |
| 4         | 70.5      | 72.1      | 2.2    | 123.62     | 131.15     | 6.1    |
| 5         | 67.8      | 69.1      | 1.9    | 77.53      | 82.56      | 6.5    |
| 6         | 68.5      | 69.5      | 1.5    | 104.34     | 108.36     | 3.9    |

Figure 7. The effective lesion volume trends in the myocardium of six target sites under AC and IC for 30-s RFCA.

Table 2. Relative change of the maximum tissue temperature and effective lesion volume inside myocardium under isotropic and anisotropic conditions after 30 s of RFCA.
recognized from the corresponding occupation of current density.

4. Discussion

This simulation study investigated the feasibility and ablation performance of an anatomy-based atrium model with fiber orientation. Models with and without fiber orientation were comprehensively compared, where an RBM was the core method of fiber orientation generation.

4.1. Application of RBM

RBM has been demonstrated to be viable and efficient compared to a conventional modality that requires the manual addition of fiber imaging results [29]. One of the key features of RBM is that the heterogeneous organization of fiber orientation is generated based on the corresponding anatomic structure, with particular focus on the relative positions of an endocardial node and its nearest epicardial node in a heart model. Thus, we believe that anatomy-based models are suitable for using RBM to study the effect of fiber orientation, as a simplified model such as a slab of the myocardium can only generate limited fiber orientation if there is no geometric variation. Although the fiber orientation can be manually adjusted by defining specific spatial MEC without the use of an RBM, the configuration may be unrealistically ideal, and it may take a great deal of time to gather sufficient clues to achieve a reasonably realistic organization of fiber orientation. In the current study, we have labeled two additional positions as Bachmann’s bundle (BB) and the interatrial groove (IG) in Figure 4. Our result shows that the fiber orientations around BB and IG are almost parallel to the OX and OY axes, respectively. For the PW fiber, the orientation ranges from 90° to 120°. In addition, structures such as PVs and the SVC are characterized by circumferential fiber orientation. These results agree closely with previous studies that captured realistic atrial myofibre architecture through imaging techniques [38,39], validating the RBM used for myocardial fiber generation of the atrium model. In addition, we applied a modified RBM combined with structure tensor filtering [29] to smooth the fiber orientation around the junction structure in the atrium model. The post-filtering outcome was quite satisfactory, as apparent from position 2 (Figure 4(a)); thus, this method is adaptable for complex structures.

4.2. Construction of the thorax model

Regarding the building-up of the thorax model as an external factor of the atria, we deem the dimension and stratification of the thorax important in the current study. As can be seen in Figure 11, the boundaries of the model were essential paths for RF current; therefore, the distribution of current and, accordingly, current density [17] depended on the dimension of the thorax. The convergence test would be
Figure 9. Temperature distribution and lesions generated at positions 2, 4 and 6 on myocardial surface. The dashed line represents the boundary of effective lesions under AC. The actual model display of the left atrial (LA) or right atrial (RA) endocardium is shown for each case, with the electrode shown in blue. LAA/RAA indicate left/right atrial appendage; LSPV/LIPV/RSPV/RIPV = left superior/left inferior/right superior/right inferior pulmonary vein; SVC = superior vena cava.

Figure 10. Current flow at position 6 in lateral view. The direction of the current flow is indicated by black arrows.
used to determine the model dimension if the construction of the thorax is not considered. However, the scale of the heart model in the current study was relatively large, and thus, the convergence test may not be able to form an appropriate distance between the electrode and the grounding pad. Further, it is greatly time-consuming for a model at the current scale, especially with the addition of fiber orientation, to run simulation repeatedly for the determination of model dimension. For the stratification of the thorax model, exclusion of these layers underestimated the resulting tissue temperature and lesion volume by nearly 2% and 5%, respectively. As a small increase in temperature will lead to a relatively significant increase in lesion volume, which could be proven by the ablation results under AC and IC, it is suggested to include detailed structures of tissues or organs inside the thorax if the research is focused on the variations of ablation results under different electrical boundary conditions.

4.3. Effect of MEC

Unlike typical RFCA simulation configurations [6], the MEC in a fiber-developed model has different values in the longitudinal and transverse directions and the anisotropic ratio of MEC was 1.43:1 in this study. To carry out a reasonable comparison of the ablation results under AC and IC, the MEC under IC, which was 0.345 m$^{-1}$, was an average of the MEC under AC around 500 kHz [36]. Under this relation, the use of MEC under IC is more likely to cause higher ablation temperature and lesion volume. This finding agreed with the simulation study of Jain et al. [40] where an experimental validation of an RFCA model was performed. The MEC in that computer model was isotropic at 0.355 m$^{-1}$, which was almost the same value used in the current study. Under a similar RF power application of 20 W, their numerical model results showed a more significant overestimation of lesion dimension, which was up to 30% for lesion volume. Apart from the possibility of errors occurring during the experiment and measurement, one possible reason may be that the actual MEC (AC) of the tissue sample used was smaller than in the current study. Different fiber architecture of the myocardium may also be a factor in causing differences in lesion size between studies. Nevertheless, this effect would not be as significant if there was no drastic variation of myocardial structure around the active electrode as in the current results. Both the results from Jain et al. and the current study showed that the isotropic MEC did not absolutely align with the anisotropic one to produce a close outcome.

As established in simulation studies that performed validation of RFCA computer models with isotropic MEC, matching the initial impedance of the overall system is a common way to unify the model with the experimental setup by adjusting the MEC in simulation [37,41]. A proper MEC under IC that can produce results close to the use of AC (i.e., the actual MEC in a real situation) can therefore be simultaneously determined. However, there was an inevitable deviation of the lesion dimension between the two scenarios even if the initial impedances were matched. The lesion depth and width of the computer model tended to be larger and smaller than the experimental results, respectively. The average differences in lesion depth and width between simulation and experiment from these two studies were both approximately 10%. These findings suggest that the

![Figure 11. Current density distribution under AC and IC on myocardial surface for positions 2, 4 and 6. The green dashed line in each IC result represents the occupation of current density under AC.](image)
computational lesion is relatively more spherical because of the isotropic setting of cardiac tissue. If the isotropic MEC is determined by averaging the anisotropic MEC, we found that the initial impedance of the ablation system under IC was always smaller than the one under AC regardless of the ablation sites. This was believed to produce higher tissue temperature and larger lesion volume under IC since the electrical field inside the myocardium was strengthened based on Ohm's law $V = \sqrt{PR}$.

An additional group of tests was conducted with another set of AC values (0.107 and 0.045 S m$^{-1}$ in longitudinal and transverse directions, respectively) acquired from an AF case [42]; the tissue temperature results are plotted in Figure 12. A disruptive decrease in tissue temperature occurred and no effective lesions formed at positions 1 or 2. The low conductivities agreed with the fact that the onset of AF decelerates atrial conduction, and that this conduction delay is more striking in the LA [43]. Thus, increased RF power is required if AF is present, which may explain why the RF power for clinical AF ablation exceeds that of the current study [44].

4.4. Insights into fiber orientation with corresponding target sites

Our results indicate that the fiber orientation fully depends on the anatomic structure of the heart. Locations with the complex anatomic structure were also associated with the complex distribution of fiber architecture such as positions 1, 2 and IG. At position 1 in Figure 4(b), the convergence of different fiber orientations up to approximately 180° is apparent. However, myocardial fibers are bidirectional and demonstrate a relative rotation between fibers in different sections of the heart [45]. We chose four ablation points (positions 1, 3–5) referring to wide antral PVI in this study because it was reported to be more effective than traditional ostial PVI for achieving AF freedom [28], as demonstrated by position 2. Although structures with drastic fiber or anatomic variation (e.g., LAA or PV ostium) may lead to a comparatively higher local current density around the active electrode during RFCA, these structures acutely restrict lesion growth. Ablation at these sites can be applied with sufficient RF power, contact force, and ablation duration. As shown by position 6, uniformly distributed myocardial fibers are more likely to promote lesion volume. Moreover, uniform fiber distribution reduces the temperature and lesion volume differences between AC and IC. Position 6 was the only RA target site corresponding to the SVCI ablation. Considering the rather simple geometric structure and uniform fiber orientation of the SVC, this is a special position for a single-electrode catheter because ablation for common AF triggers is conducted in the LA [22]. The PV may have features that are most similar to those of the SVC, but the recent ablation strategy does not recommend using a single-electrode catheter to proceed into PVs [46]. Based on the RBM, although the LA roof is measured to be the thickest part of the LA [47], the LA-roof myocardial fibers tend to point upward, contributing to transmural lesions; this can be explained by the minimum difference between the epicardial and internal tissue temperatures as well as the lesion depth in Figure 8. Further, as linear ablation at the LA roof is performed to connect encircling lesions of the LSPV and RSPV [48], the time cost could be reduced by employing wide antral PVI.

4.5. Limitations

This study had certain limitations. First, an inevitable error was caused by electrode contact at all ablation sites because of the variable myocardial structure, especially at positions 2 and 6. The former position was a PV-atrial wall junction point and the latter was an annular surface. For these two positions, it was difficult to establish a common agreement regarding the surface area and volume of insertion with the remaining four cases, in which the anatomic structures were relatively simple. Thus, the results of comparing the six target sites may deviate to some extent from the actual tissue temperatures and lesion volumes.

Second, the local fiber orientation trends were a concern when determining the target sites. We chose positions where the variation trend of the anatomic structures of epicardium and endocardium were relatively consistent with each other so that the epicardial fiber orientation almost directly reflected that of the endocardium; this could be confirmed by the RBM. Analysis of the fiber orientation at a complex location, such as the interatrial septum, may be challenging because the fiber orientation around this area is believed to be affected by both the LA and RA.
Third, as mentioned in Section 2.2.2, we employed two convection coefficients instead of fluid dynamics to simulate the effect of blood cooling. Blood flow would be highly turbulent owing to the existence of four PVs in the LA or SVC and IVC in the RA. All of these structures act as blood-flow inlets, and thus the lesion shape is believed to form in an unpredictable manner. However, the inclusion of fluid dynamics is essential to achieve realistic outcomes. When convection coefficients are used, the lesion surface weight and blood temperature are overestimated [23].

Fourth, as we were focusing on the anisotropy in MEC, the thermal conductivity of biological tissue was reported to be anisotropic as well. Heat transfer can also be influenced by the fiber architecture [49]. Unfortunately, specific data of anisotropic thermal conductivity for cardiac tissue are currently lacking. The comparison between the impact of anisotropy in electrical and thermal conductivities is considered to be a meaningful topic for future research.

5. Conclusion

In this study, modeling-based analysis of lesion characteristics during RFCA under the effect of anisotropy in myocardial electrical conductivity was performed, considering the anatomic structure of the human atrium and relevant RBM-based fiber orientation. When RF power of 20 W was applied for 30 s, an anisotropic ratio of MEC of 1.43:1 led to a reduction of tissue temperature and lesion volume for nearly 2.2% and 7.0%, respectively compared to the ablation results under the isotropic MEC which was the average of anisotropic conductivities. The anisotropy in MEC had a greater effect of reduction on lesion volume than on tissue temperature. The extent of this reduction increased with the complexity of fiber variation around the ablation site, which was determined by the local anatomical structure of the heart. For locations such as the PW, LA roof, and inside of veins or venae cavae, the reduction of tissue temperature and lesion volume under anisotropic MEC was approximately 1.9% and 5.6%, respectively. Those for the ostium of PV and LAA were approximately 2.8% and 9.8%, respectively. The current findings indicate that the anisotropic effect tends to be comparatively slight around locations with a more uniform fiber distribution. In contrast, ablation sites with an acute variation of fiber architecture may experience a greater anisotropy impact in MEC, which restrained lesion growth during RFCA.

Disclosure statement

No potential conflict of interest was reported by the author(s).

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