Impact of surrounding tissue on conductance measurement of coronary and peripheral lumen area

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Parallel conductance (electric current flow through surrounding tissue) is an important determinant of accurate measurements of arterial lumen diameter, using the conductance method. The present study is focused on the role of non-uniform geometrical/electrical configurations of surrounding tissue, which are a primary source of electric current leakage. Computational models were constructed to simulate the conductance catheter measurement with two different excitation electrodes spacings (i.e. 12 and 20 mm for coronary and peripheral sizing, respectively) for different vessel–tissue configurations: (i) blood vessel fully embedded in muscle tissue, (ii) blood vessel superficially embedded in muscle tissue, and (iii) blood vessel superficially embedded in muscle tissue with fat covering half of the arterial vessel (anterior portion). The simulations suggest that the parallel conductance and accuracy of measurement is dependent on the inhomogeneous/anisotropic configuration of surrounding tissue, including the asymmetric dimension and anisotropy in electrical conductivity of surrounding tissue. Specifically, the measurement was shown to be accurate as long as the vessel was superficial, regardless of the considerable total surrounding tissue dimension for coronary or peripheral arteries. Moreover, it was shown that the unfavourable impact of parallel conductance on the accuracy of conductance catheter measurement is decreased by the combination of a lower transverse electrical conductivity of surrounding muscle tissue, a smaller electrode spacing and a larger lumen diameter. The present findings confirm that the conductance catheter technique provides an accurate platform for sizing of clinically relevant (i.e. superficial and diseased) arteries.

Keywords: parallel conductance; impedance catheter; asymmetric surrounding tissue; anisotropic electrical conductivity

1. INTRODUCTION

The conductance catheter has been widely adopted for sizing the chamber volume of ventricles [1–5] and the luminal diameter of aorta [6,7] and medium-size arteries [8–11]. Two bolus injections of saline solutions with known electrical conductivities (e.g. normal and half-normal) to transiently displace blood and to effectively minimize the haemodynamics-induced blood conductance alterations were introduced by Kassab et al. [9,10] for analytical determination of vessel cross-sectional area (CSA) and the electric current leakage through the vessel wall and surrounding tissue (parallel conductance).

Parallel conductance (the extent of electric current leakage through surrounding tissue) is a function of vessel–tissue configuration and the distance between excitation electrodes. Both geometry (tissue dimension and/or relative vessel position in tissue) and electrical conductivity of tissue are fundamental parameters that affect parallel conductance and hence measurement accuracy. Anisotropic features of tissue electrical conductivity also affect parallel conductance. In fact, parallel conductance has been treated as a potential source of offset in conductance measurements [6,8–10,12–14]. In addition, the parallel conductance mediated by heterogeneous electrical properties of aortic wall has been examined in terms of aortic diameter measurements [12].

Here, we used two excitation spacings (12 and 20 mm) to measure coronary (diameter range 2–5 mm) and peripheral (diameter range 4–10 mm) vessels of clinical interest (reference vessels for sizing of a diseased segment), respectively. Accurate determination of vessel lumen size is important for clinical interventions such as balloon angioplasty and stenting. A larger excitation spacing for peripheral vessels produces a greater parallel conductance, which can affect the accuracy of lumen sizing. Accordingly, the objective was to determine how

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the conductance catheter performance is affected by the geometrical/electrical vessel–tissue configuration for measurement of coronary and peripheral CSA, using the two-injection approach. This is important because the anatomical positions of blood vessels and surrounding tissue structures are diverse. Although diseased vessels tend to be superficial [15], it is necessary to systematically investigate the effect of non-uniform surrounding tissue configuration on the potential measurement error. We computationally assessed how the accuracy of conductance catheter measurement is affected by various vessel–tissue configurations and validated the numerical results with ex *vivo* measurements.

2. COMPUTATIONAL METHODS

2.1. Geometry and computational domain

All computational simulations were carried out in a three-dimensional geometry of circular vessel embedded into the tissue to various degrees, as depicted in figure 1. In order to model the anatomically relevant vessel location within the muscle for coronary and peripheral vessels, different blood vessel positions relative to the surrounding tissue were investigated (figure 1). For a coronary artery, two different configurations were explored: (i) a blood vessel largely embedded into a cardiac muscle (positioned in the centre of cardiac muscle tissue; config. C-I in figure 1b) and (ii) a superficial blood vessel with fat covered on the anterior surface of vessel (positioned 1–2.5 mm below the fat surface; config. C-II in figure 1b). The outer tissue of fat layer represents a simplified exterior cardiac structure encompassing the pericardial sac and fluid, which is a more anatomically realistic environment of coronary arteries. For a peripheral artery, three different configurations were examined: (i) a blood vessel profoundly embedded into skeletal muscle (positioned in the centre of skeletal muscle tissue; config. P-I in figure 1c), (ii) a blood vessel superficially positioned into tissue (positioned 1 mm below the tissue surface; config. P-II in figure 1c), and (iii) a superficial blood vessel with fat covered on the anterior surface of vessel (config. P-III in figure 1c). Two different excitation electrode spaces of conductance catheter (0.9 mm diameter corresponding to the annulus CSA to the annulus CSA (area between the muscular vessel segment (30 mm) was set to be sufficient to encompass the electric field by four electrodes of conductance catheter (0.9 mm diameter corresponding to 0.065°). The computational domain in the radial direction, including blood vessel lumen and surrounding tissue, is more anatomically relevant, while the domain was assumed to be cylindrically axisymmetric in the previous conductance catheter studies [7,9,12].

2.2. Governing equation and numerical methods

In order to obtain the electric field in lumen and tissue, we solved the Poisson equation:

\[ \nabla \cdot (\sigma \nabla v) = I, \]  

(2.1)

where \( \sigma, v \) and \( I \) respectively denote electrical conductivity, electric potential and driving current. The values of various parameters used in the simulations are listed in table 1. The values chosen for tissue electrical conductivity were based on the range of measured values in the literature [16–25]. At the outer boundary of computational domain, a zero spatial derivative of electric potential (i.e. no electric current across the boundary) was assumed similar to that of the previous conductance catheter study [9]. As described in detail in the previous studies [9,10], CSA and parallel conductance can be analytically (lumped model) determined, using two injections of different saline solutions with the following equations:

\[ \text{CSA} = L \cdot \frac{G_2 - G_1}{\sigma_2 - \sigma_1}, \]  

(2.2a)

and

\[ G_p = \frac{\sigma_2 \cdot G_1 - \sigma_1 \cdot G_2}{\sigma_2 - \sigma_1}, \]  

(2.2b)

where \( G, G_p \) and \( L \) respectively denote total conductance, parallel conductance and distance between detection electrodes, and subscripts 1 and 2 designate two different injections of saline solutions. The underlying assumptions for the derivation of equations (2.2a) and (2.2b) are that a catheter–vessel–tissue configuration is modelled as two parallel resistors and that the parallel conductance remains unchanged by two different injections. This allows for two equations associated with two injections to solve for CSA and parallel conductance simultaneously (i.e. two equations for two unknowns). The determination of per cent error (%error) in lumen diameter takes into account the presence of conductance catheter size by adding the conductance catheter CSA to the annulus CSA (area between the catheter and vessel) indicated by equation (2.2a).

The finite volume commercial package *ANSYS FLUENT* (v. 12.1, *ANSYS, Inc.*.) was used to solve the governing equation with the user-defined codes. For all simulations, the computational domain consisted of hexahedral elements. All simulations were performed on a Dell Studio XPS machine containing two Intel Core i7 processors and memory of 9 GB.

2.3. Electrical conductivity model

For the lumen, the electrical conductivity values of normal and half-normal saline solutions were specified as listed in table 1. For the surrounding tissue region, an electrical conductivity tensor \( \sigma \) at a given point of the muscle tissue was assigned. In the present study, the muscle fibres in surrounding tissue region were assumed...
to lie in the planes parallel to the $x-z$ plane (figure 1). In addition, the surrounding muscle tissue was assumed to have anisotropic electrical conductivity (i.e. different electrical conductivity values are assigned for axial, $\sigma_L$ and transverse direction, $\sigma_T$, of the fibre; table 1) in accordance with the previous studies on anisotropic tissue conductivity [16–25]. According to the muscle fibre orientations on the planes, the conductivity

Figure 1. Schematic of the computational domain for simulations that mimics the tissue element having a blood vessel embedded in it. (a) A multi-lateral view and dimension of the domain (left and centre panels) and schematic configuration of ex vivo conductance measurement (right panel). $L_T$, $L_{TD}$, $L_{EE}$ and $D_L$ respectively denote the length of blood vessel segment, the thickness of tissue element, the depth of blood vessel in tissue, the excitation electrodes space and the diameter of vessel lumen. $\phi$ denotes the circumferential angle of tissue element, which is set as $\pi/40$, $\pi/20$ and $\pi/10$ for $L_T$ of 18, 33 and 60 mm, respectively. The white solid line and the black square dots in the centre panel, respectively, demonstrate the conductance catheter and four electrodes (inner pair: detection electrodes, outer pair: excitation electrodes) of the catheter. The width of each electrode and distance between detection electrodes were assumed to be 1 mm. A variety of surrounding tissue configuration depending on the anatomic position of blood vessel embedded in (b) cardiac muscle tissue and (c) skeletal muscle tissue. The left panels in (b) and (c) indicate a deeply embedded blood vessel ($L_{TD} = (L_T - D_L)/2$) that gives rise to a uniform configuration of surrounding muscle tissue (configs C-I and P-I). The centre panel in (c) illustrates a superficial blood vessel ($L_{TD} = 1–2.5$ mm) that gives rise to an asymmetric dimension of surrounding muscle tissue (config. P-II). The right panels in (b) and (c) illustrate a superficial blood vessel ($L_{TD} = 1–2.5$ mm) where half of the vessel is surrounded by a thin layer of fat tissue (configs C-II and P-III). The dark grey area in tissue indicates the fat layer, which is 1–2.5 mm thick. The outer tissue of fat layer in the right panel in (b) represents a simplified exterior cardiac structure encompassing pericardium. Note that the superficial vessel position has different anatomic configurations respectively in coronary and peripheral arteries (catheter diameter was set as 0.9 mm corresponding to 0.035”).
tensor can be transformed as follows, similar to the previous study [28]:

\[ \bar{\sigma} = \begin{bmatrix} \sigma_T \cos^2 \theta + \sigma_L \sin^2 \theta & 0 & (\sigma_L - \sigma_T) \sin \theta \cos \theta \\ 0 & \sigma_T & 0 \\ (\sigma_L - \sigma_T) \sin \theta \cos \theta & 0 & \sigma_T \sin^2 \theta + \sigma_L \cos^2 \theta \end{bmatrix}, \]

(2.3)

where \( \theta \) denotes the muscle fibre angle with the z-axis (figure 1). A thin layer (1–2.5 mm) of fat tissue with lower electrical conductivity different from muscle fibre (table 1) was assumed to cover the superficial region of surrounding tissue to model an anatomically relevant, surrounding tissue configuration (figure 1b,c).

### 2.4. Ex vivo experiments

Isolated segments of carotid and iliac arteries were obtained from either canine or swine. The vascular segment was rinsed with saline and all arterial branches were ligated. Both ends of the segment were cannulated and secured. The segment was then placed in a plastic container filled with saline at a depth of 30 mm (right panel in figure 1a), which is commensurate with the computational domain in figure 1 (i.e. \( L_T = 33 \) mm). A variety of saline bath concentrations were made by combining 0.9 per cent saline and deionized water so as to mimic the surrounding tissue of conductivity. For each bath concentration, the diameter of the vessel segment was measured using the conductance catheter at various depths of the segment (i.e. 1, 8 and 15 mm below the bath surface). For two conductance measurements, the segment was perfused with normal and half-normal saline solutions. After conductance measurements, the segment was perfused with cardiac muscle having an electric conductivity of mid-anisotropic ratio (AR = 2.2) as listed in table 1.

The results show that the error in predicted diameter for the fully embedded configuration for the same size vessel is significantly higher for the larger electrodes space (i.e. approx. –30% for config. P-I in figure 2b versus approx. –10% for config. C-I in figure 2a) despite the difference in electrical conductivity value of each muscle tissue. In fact, specifying an identical electrical conductivity value of muscle tissue for both electrodes spaces (i.e. the smaller electrodes space with skeletal muscle versus the larger electrodes space with cardiac muscle) led to a more amplified error (data not shown). The finding is reasonable because the larger electrodes space provides a more distributed electric field, leading to an elevated current leakage through surrounding tissue and hence an increased error at a given vessel–tissue configuration. The results also demonstrate that the error decreases as the position of blood vessel becomes superficial (config. C-I→ config. C-II and config. P-I→ configs P-II and P-III) regardless of total tissue thicknesses considered for both electrodes spaces. The extent of reduced error with the vessel depth, however, was shown to depend on the degree of parallel conductance induced by the combination of vessel–tissue–catheter configuration. Likewise, the results indicate that the layer of fat covering the anterior surface of blood vessel seems to improve the accuracy of measurements but the extent of improvement also depends on the parallel conductance.

### 3. RESULTS

#### 3.1. Effect of geometrical configuration of surrounding tissue

The anatomical positions of arterial blood vessels and their surrounding tissue structures are diverse throughout the cardiovascular system. The electric field is therefore modulated by the geometrical configuration of tissue. As apparent from equation (2.2), this can affect the accuracy of conductance catheter measurements for lumen size. Figure 2 depicts the %error in predicted diameter for the relative vessel–tissue configurations with two different electrodes spaces (\( L_{ES} \) of 12 and 20 mm) as shown in figure 1. The error was depicted for three different surrounding tissue dimensions considered (i.e. total tissue thickness \( L_T \) of 18, 33 and 60 mm) and for each configuration. In figure 2a, the conductance was measured for a 3 mm diameter vessel by a catheter of 12 mm electrodes spacing, assuming that the vessel is surrounded by the cardiac muscles having an electric conductivity of mid-anisotropic ratio (AR = 2.2) as listed in table 1. In figure 2b–d, the conductance was measured for the vessel diameters of 3, 6 and 9 mm by a catheter of 20 mm electrodes space, assuming that the surrounding tissue is a skeletal muscle with an electrical conductivity of mid-AR (i.e. AR = 6.6) as listed in table 1.

The results show that the error in predicted diameter for the fully embedded configuration for the same size vessel is significantly higher for the larger electrodes space (i.e. approx. –30% for config. P-I in figure 2b versus approx. –10% for config. C-I in figure 2a) despite the difference in electrical conductivity value of each muscle tissue. In fact, specifying an identical electrical conductivity value of muscle tissue for both electrodes spaces (i.e. the smaller electrodes space with skeletal muscle versus the larger electrodes space with cardiac muscle) led to a more amplified error (data not shown). The finding is reasonable because the larger electrodes space provides a more distributed electric field, leading to an elevated current leakage through surrounding tissue and hence an increased error at a given vessel–tissue configuration. The results also demonstrate that the error decreases as the position of blood vessel becomes superficial (config. C-I→ config. C-II and config. P-I→ configs P-II and P-III) regardless of total tissue thicknesses considered for both electrodes spaces. The extent of reduced error with the vessel depth, however, was shown to depend on the degree of parallel conductance induced by the combination of vessel–tissue–catheter configuration. Likewise, the results indicate that the layer of fat covering the anterior surface of blood vessel seems to improve the accuracy of measurements but the extent of improvement also depends on the parallel conductance.

Because the peripheral vessels of clinical relevance have a wide array of lumen diameters (4–10 mm), the %error was also assessed for three different diameters. Figure 2b–d demonstrate that the error is much lower for larger vessels (i.e. \( D_L = 3 \text{ mm} \) versus \( D_L = 6 \text{ and} \)}
(9 mm). Specifically, the results depict that the level of error is less than approximately 3 per cent for the vessel diameter 6 and 9 mm, and the variation in error with vessel–tissue configuration is negligible compared with the vessel diameter of 3 mm. This is due to the fact that the lumen conductance by saline solutions dominates over the current leakage for the larger vessels.

3.2. Effect of anisotropic electrical conductivity

Anisotropy in electrical conductivity of muscle tissue has been extensively investigated for the cardiac [16–19] and skeletal muscle [18,20–25] suggesting a wider spectrum for skeletal than for a cardiac muscle (i.e. \( \text{AR} = \sigma_L/\sigma_T = 1.1–3.4 \) for cardiac and 2.7–15 for a skeletal muscle, table 1). Therefore, it is essential to understand how such a wide spectrum of anisotropic electrical conductivities affects the conductance measurement accuracy. The top panels of figure 3(a,b) depict the absolute value of %error in predicted diameter with an increase in anisotropy of the electrical conductivity of cardiac (figure 3a(i)) and skeletal muscle (figure 3b(i)) surrounding a 3 mm diameter vessel at a given tissue dimension (i.e. \( L_T \) is 33 mm).

For the cardiac muscle, the significant correlation between error and AR is not found for the superficial vessel–tissue configuration (i.e. config. C-II) while the error decreases with increasing AR for the centred-vessel position (i.e. config. C-I). Specifically, the results indicate that the majority of electric current is leaked
Figure 3. (Caption opposite.)
through surrounding tissue for the lowest AR considered (i.e. AR = 1.1) so that the %error is not assessable (data not shown). For the skeletal muscle, however, the results demonstrate that the error decreases as the anisotropy increases for the three relative vessel positions. For the centred-vessel position (i.e. config. P-I), similar to the cardiac muscle, the excessive electric current leakage was observed for the lowest AR investigated (i.e. AR = 2.7, data not shown). However, the error decreases with the higher ARs. For the more superficial vessel positions (i.e. configs P-II and P-III), the level of error is drastically reduced and decreases with the increasing AR as well. Specifically, the error falls below 10 per cent when the AR increases beyond three for config. P-III.

The results indicate that the error has a strong dependency on the transverse electrical conductivity of muscle tissue especially for the centred-vessel position. The middle panels of figure 3a,b depict data similar to the top panels such that the error is plotted relative to the transverse electrical conductivity. Despite the different spectrum of transverse electrical conductivity and AR for cardiac and skeletal muscle tissue, the results demonstrate that the error monotonically decreases as the transverse electrical conductivity decreases for both cardiac and muscle tissue when the vessel is fully embedded in surrounding tissue. For the superficial vessel position, however, the results demonstrate that the dependency of error on the transverse electrical conductivity is less sensitive. In fact, for the superficial coronary vessel position, the error was shown to virtually not change with the transverse electrical conductivity of muscle tissue since the axial and transverse direction of tissue varies depending on how the muscle fibres lie relative to the blood vessel. The effect of muscle fibre orientation on the measurement accuracy, however, was shown to be negligible for the superficial vessel position for both cardiac and skeletal muscle tissue.

3.3. Effect of tissue directionality

The error in diameter was assessed for a series of angles that the muscle fibres and the blood vessel make under the centred (configs C-I and P-I) and the superficial (configs C-II and P-III) configurations of blood vessel. Figure 4 depicts the absolute value of %error in diameter for a variety of muscle fibre orientations relative to blood vessel, which range from 0° (parallel with blood vessel) to 90° (perpendicular to blood vessel). Figure 4a,b demonstrates the error with the angle of cardiac and skeletal muscle fibres which have the ARs of 2.2 and 6.6, respectively.

The results demonstrated that muscle fibre orientation can affect the electric current through the tissue since the axial and transverse direction of tissue varies depending on how the muscle fibres lie relative to the blood vessel. The effect of muscle fibre orientation on the measurement accuracy, however, was shown to be negligible for the superficial vessel position for both cardiac and skeletal muscle tissue.

3.4. Validation

The effect of relative vessel position in tissue on conductance catheter performance was assessed using isolated vessel segments ex vivo. Figure 5 depicts the %error in measured diameter with the vessel depth in the saline bath of different concentrations for two different size blood vessels (i.e. $D_h = 4.3$ mm canine carotid artery and 6.8 mm swine iliac artery). The conductance catheters with shorter and longer spaced excitation electrodes (i.e. 12 and 20 mm, respectively) designed for the coronary and peripheral arteries were used for the measurements of smaller and larger size vessels, respectively. The results demonstrate that at the bath concentration of less than or equal to 0.2 per cent (most physiologically relevant), the level of error in diameter is less than 7.5 per cent regardless of vessel position and size. 0.2 per cent and 0.1 per cent bath concentrations correspond to the mid-transverse electrical conductivity of cardiac muscle and the highest limit of transverse electrical conductivity of skeletal muscle listed in table 1, respectively. However, the dependency of the catheter measurement accuracy on the vessel
depth in bath becomes more prominent as the bath concentration increases (i.e. as the surrounding media becomes more conductive). Specifically, at the highest bath concentration made (i.e. 0.3% which is closer to the highest limit of transverse electrical conductivity of cardiac muscle listed in Table 1), the extent of decrease in %error magnitude of diameter with vessel position was shown to be significant for the deeper vessels. This

![Figure 5](https://example.com/figure5.png)

Figure 5. The %error in diameter measured for an isolated (a) canine carotid artery segment ($D_L = 4.3$ mm) and (b) pig iliac artery segment ($D_L = 6.8$ mm) submerged in saline bath with different concentrations (i.e. 0.1%, 0.2% and 0.3%). The black, grey and white bars denote respectively the submerged depth of 1, 8 and 15 mm. 0.3% and 0.1% bath concentrations correspond to the electrical conductivity values closer to the highest limit of transverse electrical conductivity of cardiac and skeletal muscle listed in Table 1, respectively. The excitation electrode spacing is 12 and 20 mm for the carotid and iliac artery segment, respectively.

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is attributable to a large amount of electric current leakage through surrounding saline bath, which is consistent with the simulations depicted in figures 2 and 3.

4. DISCUSSION

A novel approach using two bolus injections of saline solutions with known electrical conductivities has been previously proposed in order to analytically determine arterial CSA and parallel conductance [8–11]. The parallel conductance or electric current leakage through surrounding tissue is one of the most challenging issues in determination of arterial CSA by electric conductance measurements. Although the capability of conductance catheter technique has been extensively investigated using computational simulations, the majority of the vessel–tissue models were idealized to be homogeneous and axisymmetric around blood vessel [7,9,12,29]. A systematic assessment of the role of non-uniform surrounding tissue configuration on the parallel conductance in non-axisymmetric geometries was the focus of this study.

The major finding of this study is that the level of %error in diameter measurement decreases as the position of blood vessel becomes superficial in tissue, regardless of entire surrounding tissue dimensions considered (i.e. config. C-I → config. C-II and config. P-I → configs P-II and III). In fact, the variability of %error in predicted diameter depending on the surrounding tissue dimension is negligible: i.e. the variability over all the surrounding tissue thicknesses considered (i.e. 18, 33 and 60 mm) are ±0.5 per cent and ±1.1 per cent, respectively, for configs C-I and C-II in the case of cardiac muscle and ±1.9 per cent, ±0.6 per cent and ±0.3 per cent, respectively, for configs P-I, P-II and P-III in the case of skeletal muscle for the smallest diameter explored (figure 2a,b).

Although the role of heterogeneous/anisotropic electrical properties of aortic wall on conductance catheter measurement has been studied [12], the effect of wide spectrum of anisotropic characteristics in electrical conductivity of surrounding tissue on the accuracy of conductance measurements has not been fully addressed. The present simulations demonstrated that the level of %error in diameter significantly and monotonically decreases as the transverse electrical conductivity decreases especially for a fully embedded vessel configuration (the middle panels in figure 3) while no significant correlations were found between the error and axial electrical conductivity (the bottom panels in figure 3). Furthermore, the results from figure 4 demonstrated that the variability of %error in diameter is negligible for the superficial vessel configuration (configs C-II and P-III) while the %error in diameter seems relatively more sensitive to the muscle fibre angle for the centred-vessel configuration (configs C-I and P-I) for both cardiac and skeletal muscle tissue.

In fact, the high aspect ratio of electrical conductivity (e.g. skeletal muscle, the top panel of figure 3b) can give rise to the significant variations in transverse and axial component of electrical conductivity of muscle fibre relative to blood vessel so that the altered relative component of transverse electrical conductivity have an impact on the parallel conductance and hence measurement accuracy. The results, however, demonstrated that the influence of muscle fibre orientation on measurement error is generally not as significant as that of the superficial portion of blood vessel. This was especially true when the vessel–tissue–catheter configuration leads to an elevated parallel conductance (e.g. small lumen diameter plus large excitation electrodes space, figure 4b).

The electric current leakage through surrounding tissue seems to be dominantly determined by the superficial portion of surrounding tissue rather than by the entire surrounding tissue dimension, despite the variations in geometrical/electrical configuration of the tissue surrounding coronary and peripheral arteries as illustrated in figure 1b,c, and table 1. The results also suggest that the favourable influence of superficial vessel position on CSA measurement accuracy is likely to be the most prominent for the condition that contributes to the elevated parallel conductance (i.e. smaller lumen diameter + larger electrodes space + higher electrical conductivity of tissue), while such impact seems to be mitigated for larger lumen diameters owing to the predominance of luminal conductance over parallel conductance (figures 2 and 3).

The effect of superficial vessel position on CSA measurement accuracy was also verified by the ex vivo experiments (figure 5), demonstrating that the decrease in the level of %error is more evident at the higher saline bath concentration (greater than 0.2% saline) commensurate with higher electrical conductivity for both electrodes spaces (i.e. 12 and 20 mm) and for both lumen diameters (i.e. 4.3 and 6.8 mm) considered. The level of error is generally lower for the larger lumen diameter, even at the large electrodes space with high electrical conductivity of tissue (figure 5a versus 5b at 0.3% saline bath concentration).

The physiologically/pathologically relevant anatomy of arterial vessel and surrounding tissue is even more complicated than that depicted in the model. The vessel–tissue model in present study may be thought of as a simplified model that can be further refined. As an example of more anatomically relevant and complex tissue structure, a multiple layers of tissue with different electrical conductivity value of each layer were modelled exterior to the fat layer of coronary vessel. The difference in predicted error of lumen diameter induced by the multiple layers of tissue, however, was shown to be negligible (data not shown). This is attributable to the fact that the electric current leakage is primarily regulated by the fat layer exterior to vessel so that the added complexity in surrounding tissue configuration minimally affects the conductance catheter measurements.

Using various imaging modalities such as echocardiography, computed tomography and magnetic resonance imaging, the anatomical features of epicardial/pericoronary fat have been determined [30–36]. The thickness of epicardial fat layer on the surface of the heart has been shown to vary from 1 to 16.5 mm [30,33,37]. The fat layer thickness adopted in the present study (i.e. 1–2.5 mm) is commensurate with a
previous study [36] and can be considered as a conservative configuration since larger fat volumes will induce a more insulative environment. Other details in structure such as various layers of vessel wall, including endothelium and vascular smooth muscle can be modelled to have heterogeneous electrical properties similar to a previous study [12].

Despite a wide variety of anatomical structures and compositions of arterial blood vessels and surrounding tissues, the present numerical simulations and ex vivo experiments suggest that the error in lumen diameter measurement is sufficiently small for the clinically relevant vessel–tissue–catheter configuration (i.e. superficial anatomy; [15,38]). Specifically, the present findings confirm that for the coronary arteries, the combination of small excitation electrodes spacing and existence of anterior superficial fat can render the conductance measurement for CSA determination a reliable approach despite the high electrical conductivity of cardiac muscle tissue (table 1). Moreover, it has been shown that in the two injection approach, the cardiac cycle–excitation frequency-induced nonlinear complexity in conductance measurement for coronary arteries can be effectively minimized by the transient displacement of blood by two bolus injections of saline solution [8,10,11]. On the other hand, for the peripheral arteries, the low transverse electrical conductivity of skeletal muscle tissue plus relatively large lumen diameter can play a favourable role in measurement accuracy.

The comparative studies on the accuracy of conductance catheter measurements of artery lumen CSA have been previously performed [8,10,11]. In phantoms of known diameters, the error of conductance catheter and intravascular ultrasound (IVUS) measurements were shown to be within 3 per cent and 13 per cent of actual dimensions, respectively [11]. A previous report [39] of IVUS measurements in phantoms using various IVUS catheters and systems reported CSA errors of 19 per cent depending on the imaging systems and catheters used where the differences in CSA measurements reach up to 27 per cent in some systems. In in vivo validations, the root mean square error in diameter for the conductance catheter measurements was shown to be 5 per cent in comparison with B-mode US [10] and 10 per cent [11] in comparison with IVUS. In a recent pilot human study that included atherosclerotic vessels [8], the error in diameter of conductance catheter versus IVUS and QCA (quantitative coronary angiography) was shown to be 14.3 per cent and 25.8 per cent, respectively. IVUS and QCA are in general agreement for normal concentric cross-sectional vessels but deviate significantly in atherosclerotic vessels which typically have eccentric geometry [40,41].

As described in the previous studies [9,10], one of the assumptions of two injection approach is that the parallel conductance remains unchanged by two different injections, which enables the analytic determination of lumen CSA and parallel conductance simultaneously. Such assumption can be justified by the fact that the saline reaches the vessel of interest before it fills the capillaries and potentially changes the myocardial parallel conductance. The small error in diameter prediction by current approach can be viewed as the assessment of the reliability of such assumption. More elaborate assessment of the impact of parallel conductance altered by various anatomical circumstances of parallel vasculature should be assessed with three-dimensional full branching vascular model taken into account in future studies. Nevertheless, the measurement accuracy of vessel lumen CSA is found to be predominantly dictated by the surrounding tissue structure in vicinity of the vessel of interest rather than the dimension of entire surrounding tissue where the parallel vasculature resides.

The underlying assumptions associated with the present model need further validation in line with clinical conditions, e.g. potential incomplete displacement of blood by hand injection in larger size blood vessels, and the non-uniform electric field near detection electrodes generated by irregular vessel-catheter configurations (transition from aorta to coronary arteries, vicinity of bifurcations, etc.). The range of applicability of the present simple model would enhance the utility of the conductance catheter methods for blood vessel sizing in the clinic.

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