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An Approach for Patient-Specific Multi-domain Vascular Mesh Generation Featuring Spatially Varying Wall Thickness Modeling

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Abstract

In this work, we present a computationally efficient image-derived volume mesh generation approach for vasculatures that implements spatially varying patient-specific wall thickness with a novel inward extrusion of the wall surface mesh. Multi-domain vascular meshes with arbitrary numbers, locations, and patterns of both iliac bifurcations and thrombi can be obtained without the need to specify features or landmark points as input. In addition, the mesh output is coordinate-frame independent and independent of the image grid resolution with high dimensional accuracy and mesh quality, devoid of errors typically found in off-the-shelf image-based model generation workflows. The absence of deformable template models or Cartesian grid-based methods enables the present approach to be robust by handling aneurysmatic geometries with highly irregular shapes, arterial branches nearly parallel to the image plane, and variable wall thickness. The assessment of the methodology was based on i) estimation of the surface reconstruction accuracy, ii) validation of the output mesh using an aneurysm phantom, and iii) benchmarking the volume mesh quality against other frameworks. For the phantom image dataset (pixel size 0.105 mm; slice spacing 0.7 mm; mean wall thickness 1.401 ± 0.120 mm), the average wall thickness in the mesh was 1.459 ± 0.123 mm. The absolute error in average wall thickness was 0.060 ± 0.036 mm, or about 8.6% of the largest image grid spacing (0.7 mm) and 4.36% of the actual mean wall thickness. Mesh quality metrics and the ability to reproduce regional variations of wall thickness were found superior to similar alternative frameworks.

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CONFLICT OF INTEREST STATEMENT

Nothing to disclose.

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Keywords

Aneurysm; vascular mesh generation; cardiovascular biomechanics; multi-domain; finite element analysis

INTRODUCTION

Computational analyses of the vasculature have shown potential to improve clinical disease management (Georgakarakos et al., 2014; Raut et al., 2013a). Their accuracy is highly dependent on the 3D mesh reconstruction of the multi-domain anatomy (Raut, 2012). Automated mesh generation strategies are challenging even in well-defined geometries and it becomes more challenging in the case of complex geometries such as anatomical shapes that are typically reconstructed from segmented stacks of 2D images (e.g., blood vessel networks and organs). This manuscript describes a novel framework for the generation of finite element meshes of complex vasculature having random multi-domain geometry with the primary objective of accurately modeling the patient-specific thickness in the annular wall region.

Background

For computational mechanics, volume needs to be discretized into smaller elements avoiding slivers i.e. elements with highly skewed shapes or acute angles. While constitutive modeling (Pierce et al., 2015; Raut et al., 2014; Rodriguez et al., 2009; Simsek and Kwon, 2015), multi-physics (Chandra et al., 2013; Xenos et al., 2015), and sensitivity analysis (Celi and Berti, 2014) in vascular modeling have been explored, image-based modeling for biomechanical assessment can be leveraged for other non-vascular biomechanical analysis as shown by (Shi et al., 2014).

Image-based vascular reconstruction—For topologically simple shapes like an aneurysm surface, contour-based (Sacks et al., 1999) and deformable template (Shim et al., 2009) methods have been previously implemented. Early methods used for abdominal aortic aneurysm (AAA) reconstruction were successful for tubular geometries, but are not preferred for those with bifurcation, high tortuosity, presence of intraluminal thrombi (ILT), and large changes in cross-section. Marching cubes (Lorensen and Cline, 1987), extended marching cubes (Kobbelt et al., 2001), and dual marching cubes (Nielson, 2004; Schaefer and Warren, 2004) are robust methods for surface extraction from image-grid data (Young et al., 2008). However, as these are grid-based approaches, they suffer from primarily i) dependency of the meshing pattern and element quality on the coordinate frame orientation, ii) probability of losing sharp features, iii) uniform mesh density that calls for additional mesh improvement operations, and iv) need for equal grid spacing along all axes for better element quality (Zhang et al., 2005). In the Volume Marching Cube (VoMaC) approach, the Marching Cube algorithm was extended for volume meshing by using tetrahedron templates instead of triangular surface facets (Young et al., 2008). Tetrahedral and hexahedral mesh generation from image-grid data with adaptive sizing was reported by Zhang and colleagues (2005) using dual contour and octree methods. These were further applied for meshing domains with multiple materials (Zhang et al., 2010). However, the framework output has

the inherent dependence on the initial octree grid. Frameworks for image-based lumen segmentation and tetrahedral volume meshing for computational fluid dynamics (CFD) simulations have been previously reported by Antiga et al. (2008) and Cebal et al. (2005). Recently, single-domain hexahedral mesh generation of the vascular lumen was implemented with branching templates for bifurcations (De Santis et al., 2011). However, a template-based approach is limited in three ways: i) skewed elements are likely to occur if target and source geometries are not similar; ii) input data need to match the template mesh topologically; and iii) most of the template-based approaches need few landmark points as input to find correspondence between source and target geometry, and hence the need for user interaction.

Multi-domain mesh generation—Vascular meshing for cardiovascular applications is a challenge due to thin walls (Fig. 1A), the unpredictable number, shape, and location of the ILT domain, and the multiple complex shapes of the interfaces between them (Fig. 1B). At a three-domain intersection, the relative smoothness of either domain can change the geometry noticeably; Fig. 1C shows three possibilities.

Fang and Boas (2009) report on image-based tetrahedral meshing focusing on anatomical shapes. Specifically for AAA, the application A4research® (Vascops, Stockholm, Sweden) reported by Auer and Gasser (2010), can generate hexahedra dominated meshes of the vascular wall and ILT as part of a third-party service for clinical management of AAA. Young et al. (2008) expanded upon the volumetric marching cube algorithm for meshing multiple domains – an approach used for the development of the Simpleware® meshing tools (Simpleware Ltd., Exeter, UK). It carries the same limitations as the marching cube algorithm. In addition, the element density is determined by the smallest feature to be captured, e.g. for an AAA, the aneurysm sac would have to be meshed with small elements as required to model the wall precisely. This demands additional operations for mesh density improvement and the exponential increase in subsequent computational load required for more elements. The approach of using tetrahedral elements to fill the volume defined by boundary surfaces (Fang and Boas, 2009; Shi et al., 2014) may lead to artifacts when modeling thin-walled objects. Reconstruction of the thin vascular wall using commercial tools is also prone to failure (see Fig. 1A). Generating a uniform wall thickness using image dilation and subtraction operations can yield relatively better results. However, a rectangular grid pattern with sliver-like elements and irregularities across the thickness of the wall is often observed (Raut, 2012).

Motivation

The use of small tetrahedral elements to obtain high quality meshes of thin-walled structures, such as blood vessels, is possible; however, it remains a challenge as it yields a large number of degrees of freedom leading to high computational costs for the subsequent biomechanical simulation. The use of a uniform mesh density in the model is not practical since the arterial cross-section is typically large compared to the load-bearing wall. A layered wall mesh is preferable to implement spatially varying material properties across its thickness. In this work, we demonstrate a spatially controlled surface offset approach with hexahedral or wedge shaped elements for computationally efficient aortic aneurysm wall

volume meshes implementing regionally varying, patient-specific wall thickness, which is significant for subsequent computational studies (Martufi et al., 2015; Raut et al., 2013b).

METHODOLOGY

A framework was developed for multi-domain meshing of vascular structures, which imports binary masks resulting from multi-labeled segmented clinical images and outputs volume meshes suitable for finite element modeling and geometry quantification, along with the assessment of mesh quality metrics. Figure 2 shows a schematic of this framework (named “AAAMesh”) in which the main modules are: i) mask operations; ii) surface mesh extraction; iii) wall extrusion; iv) multi-domain meshing; v) quality inspection; vi) geometry quantification; and vii) file export utility. Some of the algorithms used in the surface mesh generation and multi-domain meshing modules were inspired from previous work by Fang and Boas (2009), as reported in the open source code iso2mesh (<http://iso2mesh.sourceforge.net/cgi-bin/index.cgi>).

Image-grid data creation

The framework imports a stack of 4-domain 2D masks (e.g., as seen in Fig. 4, 1-Background abdominal cavity; 2-Wall; 3-ILT (optional); 4-lumen) obtained after image segmentation along with additional DICOM header information, such as pixel size and slice spacing. These masks are converted into three sets of binary domains (BD) representing the lumen, inner, and outer wall surfaces. Three-dimensional smooth distance fields are created such that corresponding surfaces are located at zero isovalues.

Surface extraction and refinement

The framework makes use of the open-source library CGAL [Computational Geometry Algorithms Library, (Guennebaud, 2014)] to extract a surface mesh from the volume data. The strategy for surface mesh extraction is to start with the volume data for the BD and, given an internal point derived from the lumen mask, create sample points on the targeted isosurface. The latter lies at the boundary of two regions in the BD, e.g. internal and external vessel wall, and the points are created using ray tracing in random directions originating from the given point. This initial sample point set is used to obtain an improved triangulated representation of the intended surface using restricted Delaunay triangulation, i.e. new points are added until set criteria for element shape and size are satisfied (Rineau and Yvinec, 2014). The triangulated surface resulting from this method is close to the ideal surface and is capable of providing a good approximation for normals, areas, and curves. The end result is the triangulation of an isosurface for each BD. Sliver elements in the triangulation are identified by the ratio of triangle incircle radius to its circumcircle radius (r-R ratio) and are eliminated by the edge collapse method. Surface mesh smoothing is accomplished by performing Laplacian and Taubin smoothing operations (Taubin, 1995). Percentage changes in volume and surface area are maintained below a specified threshold to bind the distortion occurring during sliver removal and the smoothing operation.

Wall extrusion

Wall-E overview—A wall mesh extrusion module named “Wall-E” was implemented in the AAAMesh framework to model the vasculature with either a *precisely* uniform wall thickness or *patient-specific* variations of wall thickness. Wall-E also functions as a stand-alone script and is shown schematically in Fig. 3. It creates a layered volume mesh by offsetting the surface mesh given as input along the local surface normal. It calculates local nodal normal directions averaging facet normal directions in the immediate neighborhood. The consistency of the normal directions is verified as an added scrutiny of the surface mesh inputs to the Wall-E module. Wedge elements are naturally derived from a triangular mesh. Alternatively, for hexahedra, additional nodes are inserted to convert each triangle into three quadrangles prior to extrusion. The unique capability of Wall-E is to have node-to-node control on the wall thickness during the extrusion process. Wall-E also generates relevant face sets such as inner wall, outer wall, and end faces for boundary condition application in the subsequent finite element model. It is not constrained by the number of inlet and outlet cross-sections.

Local normal estimation—For each surface node on the inner wall surface mesh, the unit normal direction vector is given by Eq. (1),

$$\vec{n}_j = \frac{\sum_{i=1}^{N_e} \vec{e}_i}{\|\sum_{i=1}^{N_e} \vec{e}_i\|} \quad (1)$$

where, $\vec{n}_j$ is the unit nodal normal for the node under consideration (node j), N_e is the number of elements that fall within the q 'th order neighborhood ($G_{neigh_q}^j$) around node j , and $\vec{e}_i$ is the unit normal for the i 'th element. In this approach, q is calculated automatically in Wall-E by calculating the approximate multiple of surface facet segment lengths that can be fitted within a specified radius around each node. This approach is justified as it makes the process of estimating the correct neighborhood level independent of the mesh size.

Thickness assignment and mesh sweeping—In-plane patient-specific thicknesses from 3D point clouds, obtained by stacking thickness data splines from the segmented images (Martufi et al., 2009; Shum et al., 2010), are scaled by the cosine of the angle θ formed by the averaged local normal at each wall thickness data point with each transverse plane (Raut et al., 2013b). This yields the thickness along equivalent local normal directions at the data points, which are then interpolated onto surface mesh nodes. Using local thickness and normal information, additional nodes are created and nodal connectivity is defined to form hexahedral or wedge elements.

Multi-domain meshing

Multi-domain mesh generation involves the operations and algorithms shown in Fig. 4. The following algorithm elaborates on the top-level sequence of operations:

1. Extract surface triangulations for outer wall surface *WallSurfExt* and lumen surface *LumenSurf1* from the distance field;

2. Extrude outer wall surface *WallSurfExt* inward using Wall-E module and patient-specific in-plane thickness data to get *WallMesh*. Label inner wall surface as *WallSurfInt*;
3. Identify edges $Edge_i$ of *WallSurfInt* obtained after wall extrusion;
4. Create two-dimensional triangular meshes at inlet and outlet sections from each $Edge_i$ (proximal and distal ends of the artery) and combine those with *WallSurfInt* to obtain water-tight surface mesh *Surf1LtLumen* for combined lumen and ILT regions;
5. Seed nodes *seedPts* from lumen surface *LumenSurf1* for subsequent volume mesh generation of the space interior to the wall (*ILTLumenUnion*), since the ILT-lumen interface is expected here;
6. Tetrahedralize the interior space of the *Surf1LtLumen* to generate volume mesh *ILTLumenUnion* with nodes of elements at locations coinciding with *seedPts*;
7. Create reference surface *LumenSurf2* by offsetting *LumenSurf1* inward (using Wall-E) by an infinitesimal distance δ and close the ends with necessary mesh patches;
8. For every element $i_ILTLumenUnion$ in *ILTLumenUnion*, if all nodes are outside *LumenSurf2*, $i_ILTLumenUnion$ belongs to *ILT_group*, else $i_ILTLumenUnion$ belongs to *Lumen_group*;
9. Distinguish spatially-separated ILT subgroups from collective *ILT_group* by randomly selecting an element of *ILT_group* and by recursive search and subtraction of elements connected to it from the original *ILT_group* until exhausted to create subgroup (*ILT_subgroup_i*). This will also result in some unintended slivers as in some regions will be marginally close to the *WallSurfInt*. Note that the lumen-ILT interface will be smooth because of the seeded nodes from smooth *seedPts*.
10. Improve quality of volume mesh by i) eliminating non-manifold ILT regions (deformed exterior opposite surfaces) occasionally formed near *ILT_group* edge, and ii) merging the ILT sliver regions with the lumen and remeshing;
11. Store the multi-domain volume mesh following the previously determined conventions for global mesh storage (named as MeshG data structure);
12. Identify and store all the interfaces.

In the proposed approach, the triangular 2D mesh patch to close an inlet/outlet region was created using the Triangle code developed at Carnegie Mellon University (Shewchuk, 1996). The interior volume of the AAA sac (lumen and ILT combined element set, *ILTLumenUnion*) is meshed with TetGen (<http://wias-berlin.de/software/tetgen/>), which is a tetrahedral mesh generator based on boundary constrained Delaunay tetrahedralization. AAAMesh is built on an internal data organization scheme (MeshG) that can handle arbitrary multi-domain structures and store the interfaces in a systematic manner.

Mesh quality assessment and export modules

AAAMesh can evaluate the volume and surface area of the arterial geometry. The mesh quality module also provides statistics on the following three quality metrics – i) Jacobian; ii) Condition number; and iii) Oddy metric (Raut, 2012). The export module in AAAMesh can customize and export a mesh in several available formats, e.g. multi-domain volume mesh for either finite element analysis (FEA) or fluid-structure interaction (FSI) modeling, wall volume mesh for FEA, lumen volume mesh for CFD simulations, triangulation for geometric characterization of 3D surfaces, stereolithography (STL) models, etc.

Verification and validation

The proposed approach was tested for both geometric accuracy and mesh quality by using a three-pronged approach in accordance with the process pipeline: study A (surface reconstruction verification), study B (Wall-E validation), and study C (benchmarking), as shown in Table 1, with Fig. 5 illustrating the respective models. The appendix provides additional details about the testing protocol.

RESULTS

For study A (surface extraction accuracy), using 12543 triangles with an average edge length 1.5593 mm, we found a signed distance error 0.0052 ± 0.1004 mm (using typical/optimal settings) and a maximum error of 0.3506 mm (error assessment accuracy: 2×10^{-4} mm) (Raut, 2012). The error distribution is symmetric about the mean and well spread spatially (Figure 6). This error has an inseparable component originating from pixilation errors in the input mask itself. While higher accuracy can be achieved using a better input image grid resolution and a denser triangulation with shorter edge length, the default settings for AAAMesh were chosen for a computationally inexpensive volume mesh.

Study B (Wall-E validation) is illustrated in Fig. 7, which highlights the local wall thickness modeling accuracy achieved even for a high aspect ratio (~ 7) image grid. The sixteen probes used to measure thickness, as shown in Fig. 7A, are uniformly distributed over the phantom AAA sac. The wall thickness meshed was 1.459 ± 0.123 mm compared to the directly measured wall thickness, 1.401 ± 0.120 mm, in the extracted vertical plane of the image grid. The local normal wall thickness resulted in an absolute error of 0.060 ± 0.036 mm, i.e. about $8.6 \pm 5.1\%$ of the largest image grid spacing (0.7 mm) and $4.4 \pm 2.7\%$ of the mean wall thickness (Figs. 7B–7D).

In study C (Benchmarking), neither 3-Matic or Simpleware could yield a volume mesh with the patient-specific wall thickness distribution. With relaxed mesh quality criteria these applications can generate models with an approximately uniform thick wall. As seen in Table 2, the AAAMesh output is superior in both mesh quality metrics and number of elements used.

DISCUSSION

The uncertainty in locating the bounding surfaces (inner and outer vascular walls) in existing image-based modeling tools is thought to lead to large errors in the computational evaluation

of vascular mechanics using the traditional volume filling and marching cubes approaches. Hence, an extrusion (sweeping) based approach is desirable to achieve an accurate discretization of the vascular wall thickness, high element quality across the wall, and for improved computational efficiency of subsequent FE analyses.

Relevant features

Non-uniform wall thickness—One of the important features of the technique described herein is the ability to model regionally varying wall thickness (see Fig. 8A). It is also devoid of thickness inaccuracies that arise from surface mesh smoothing; the specified thickness is translated accurately to the final volume mesh. Possible sources of error clinical image data are from the input image resolution, variability in the segmentation methods, in addition to approximations used in the conversion of in-plane thickness to the thickness along the local normal direction. To this end, surface triangulation with uniform mesh density (or one with gradual change) is preferred.

Integrated approach—The framework performs operations on 2D mask data and yields 3D volume and surface meshes for multi-domain objects. A user friendly interface (see Fig. 2) facilitates training and meshing can be completed with simple tasks that control the process pipeline, which involves four modules and a complex sequence of operations based on the intended output. Typical computing times for AAA volume meshes operating on a standard Windows desktop computer are: Wall-E, approximately 35 s, and multi-domain meshing, approximately 400 s.

Robustness—The robustness of the framework is due to its i) combinatorial approach, i.e. geometry representation as a collection of simplified groups and elements facilitating their addition and subtraction; ii) automated approach that does not require manual intervention; iii) coordinate frame independence for the ensuing topology; and iv) grid-independent mesh pattern (see Fig. 8). These are essential for handling multiple and unknown number of bifurcations in an arterial network. It does not have the shortcomings that limit applicability of template-based mesh generation methods, such as skewed elements when there is a remarkable difference between the source and target geometries. In the approach, no assumptions are made regarding the shape (except for absence of sharp curvatures), number and location of ILT regions, or number of vessel branches. Due to frame independence and lack of longitudinal mesh sweeping, highly tortuous aneurysms can be modeled without affecting overall mesh quality. In addition, unlike marching cubes, study B highlights with the high accuracy of the approach in spite of input image grid with high aspect ratio. A previous use of 84 patient-specific AAAMesh output volume meshes (Raut et al., 2014; Raut et al., 2013b) demonstrates robustness of the methodology.

Multi-domain mesh generation—As the number of ILT regions, their shapes, and locations are random, the aneurysm sac can present several configurations (see Fig. 1B) - the ILT can be entirely enclosed (ILT-3), or it may be exposed to the inflow or outflow cross-sections (ILT-1 and ILT-4), or multiple ILT regions may have faces exposed to multiple inlets/outlets (ILT-2). In addition to the multi-domain meshing capability, we have

developed an internal data disposition scheme that facilitates systematic registration of the domain interfaces and retrieval of facets for each interface.

Implementation—There are additional advantages with the modularity and parallelized implementation of the framework in MATLAB. The wall thickness data can be input from a source different from the images used for surface extraction, e.g. an advanced imaging technique such as black-blood MRI that provides an accurate spatial distribution of wall thickness. In addition, the framework can be used with any image segmentation application and the layered structure of the wall mesh is more relevant from an anatomical point of view and conducive to modeling the wall with multiple constitutive material properties.

Comparison with other relevant frameworks

Open source codes—The proposed framework eliminates the explicit need for an internal point selection for thin-walled domains, as in Fang and Boas (2009), by taking advantage of the annularity of vascular surfaces and their corresponding representation in terms of multiple binary image sets. The algorithms described in the aforementioned studies work efficiently with bulk volumes. However, when a domain is thin relative to its overall size, the surface mesh of opposite faces is expected to intersect each other forming a non-manifold geometry not compatible with these traditional volume filling mesh methods. Compared to (Fang and Boas, 2009), the limitation of the present approach is that it was specifically developed for vascular structures. Similarly, Antiga and colleagues (2008) developed a complete framework capable of both image segmentation and tetrahedral mesh generation for CFD simulations with a carefully structured validation. Conversely, the focus of the present framework is on multiple domains for solid vascular structures.

Relevant commercial codes—Young et al. (2008) report on the extended Volumetric Marching (eVoMaC) method, which is excellent for versatile applications. However, it is inherently dependent on the image grid orientation and spacing, whereas this proposed framework yields a mesh pattern that is independent of both. In addition, the tetrahedral elements used to mesh a thin wall are computationally inefficient since i) thin walls will demand small tetrahedral elements to meet the target mesh quality metrics resulting in more elements compared to their hexahedral counterparts; ii) at surfaces where displacements are constrained, tetrahedra with one free node represent a challenge for achieving numerical convergence due to the kinematic constraint ensuing from the nearly incompressible behavior of soft tissues. Conversely, wedge or hexahedral elements have extra free nodes facilitating easy convergence with volume conservation. From the benchmarking analysis performed with 3-Matic, it was observed that only the AAAMesh framework can mesh patient-specific, non-uniform wall thickness. For uniform wall thickness, we obtained sliver-free meshes with 3-Matic and Simpleware when we consider the distortion error metric (with truncated iliac arteries). For all other metrics, we found that the AAAMesh mesh quality is superior.

Limitations

Extrusion of a surface along local normals requires that the nodes on the seed surface are sufficiently spaced in comparison to the surface curvature and extrusion thickness. In the

case of sharp changes in surface normal directions and nodes being close to each other, there is the probability of normals intersecting each other and resulting in intersecting volume elements. Hence, we note that for complex vascular trees with multiple branches, outward extrusion is not robust if there are highly curved concave regions, which are typically found at arterial bifurcations; inward extrusion is preferred in these scenarios. Such limitation is mitigated when meshing AAA, which consist of relatively simple vascular geometries with a single aorto-iliac bifurcation. Although not limitation of the approach, in its current implementation, the vessel inlet and outlets are expected at the proximal and distal cross-sections only. In addition, the strategy for node-to-node contact at the interface of the vascular wall with other domains is possible only when the wall is meshed with wedge-type elements.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

Challenges in image-based patient-specific vascular meshing. A) Thin-walled geometry showing typical artifacts found when reconstructing using commercial tools; B) Schematic of cross-section of aneurysm with multiple intra-luminal thrombi (ILT) regions shows that many scenarios are possible for the ILT locations and the ensuing interfaces, underscoring the robustness needed for the implementation of a volume meshing algorithm; C) Cases I, II, and III can occur where the three domains meet by giving the least priority of surface smoothness to the lumen, wall, and ILT, respectively, relative to the other two domains. This highlights how nuances of modeling can affect subsequent simulation results. Case III is the more anatomically relevant and is inherent in the proposed meshing approach with AAAMesh.

Figure 2.

Block diagram illustrating the modules and sequence of operations used for 3D mesh generation in AAAMesh. Surface triangulation extracted from scalar 3D distance field(s) followed by inward extrusion in Wall-E module that can implement node-to-node variation in thickness. Multi-domain mesh generation algorithms are subsequently initiated. All processes connected by black arrows are executed sequentially without any user interaction after meshing parameters are set. Blue arrows show optional modules that can process the finalized mesh from an existing execution file. The AAAMesh GUI facilitates various parameters pertaining to choice of element type, ILT inclusion/exclusion, mesh density, smoothing iterations, etc. Multi-domain meshing can be bypassed in the absence/exclusion of ILT.

Figure 3.

Block diagram describing the operations in the wall mesh generation module (Wall-E) designed to implement spatially variable patient-specific wall thickness. Parameter n denotes the order of the neighborhood in the vicinity of an individual surface element considered for calculating the local normal. The in-plane thickness, input as 3D point cloud, is modified to calculate thickness along the local normal direction and interpolated. An optional uniform thickness can also be implemented precisely; for wedge shaped elements no modification is needed (dashed line); however, for hexahedral meshes, surface triangulation is converted into quadrilateral tessellation (preserving the original nodes). The final volume mesh for the wall is stored in the internal data structure MeshG, which systematically stores not only the volume mesh, but also the interfaces identified as surface sets (details of which are unknown *a priori* and are variable case-to-case). The latter are necessary for prescribing boundary conditions for further multi-domain meshing or output customization.

Figure 4.

Schematic illustrating the multi-domain mesh generation approach. The 4-domain mask input is first converted into respective binary masks for creating the scalar distance field grid, from which isosurfaces are extracted (label convention: 1 → background, 2 → wall, 3 → ILT, 4 → Lumen). Patient-specific in-plane wall thickness is input as a 3D point cloud. The novel pipeline inherently yields anatomically correct interface boundaries without any *a priori* assumption regarding the number and location of ILT regions or number of branches in the vasculature. (Color legend: grey – initial tetrahedralization; yellow – single ILT region; brown and cyan – identified ILT subregions; red – slivers resulting from thin ILT)

Figure 5.

Models used for the three-pronged assessment of AAAMesh. A) Analytical surface model used in study A to test surface extraction accuracy. It has a circular cross section with the radius and center point varying in sinusoidal fashion along the longitudinal direction. ($R_{max} = 1 \text{ cm}$, $R_{min} = 3 \text{ cm}$, and $L = 9 \text{ cm}$). B) Exemplary reverse-engineered input mask (upper) and intermittent binary mask (lower) corresponding to the surface in study A; it also tests the ability of the framework to handle the no-ILT scenario in addition to smooth surface extraction from pixilated images. C) Aneurysm phantom used for μ CT images of the phantom (high contrast, better resolution) that were used as input to AAAMesh and to measure the true thickness. E) Patient-specific geometry randomly chosen from the available patient population. F) Patient-specific CT image (upper) and corresponding segmented images (lower).

Figure 6.

Study A - output surface mesh accuracy. The spatial distribution of the signed distance error shows that the error is not spatially concentrated. The histogram shows that the error distribution is nearly symmetric about zero and few nodes have high distance error. The actual mesh generation error will be smaller since the error shown also takes into account the input image pixilation error component.

Figure 7.

Comparison of phantom wall thickness observed in μ CT images and the final computational model derived from the proposed mesh generation approach. A) Surface model of the phantom depicting the 16 locations used for probing wall thickness by considering four horizontal planes intersecting with two vertical mid-planes; each intersection on the surface locates a measurement point. The extracted grid slice in the vertical plane is shown on the right where variable wall thickness in the phantom can be clearly observed; an exemplary thickness measurement along the local normal direction is shown on the inset below. (B) Probed data are compared using bar charts showing small discrepancies. (C) Regional wall thickness distribution of the phantom (mm). (D) Wall thickness variation along the circumference as a function of the normalized arc length: color $\rightarrow$ horizontal plane of the section (see inset), lines $\rightarrow$ thickness measured in the final mesh, triangles $\rightarrow$ thickness measured in the vertical image section along the local normal direction.

Figure 8.

Exemplary AAAMesh volume meshes. (A) Grid independent pattern at the aorto-iliac bifurcation and a precisely uniform wall thickness. (B) Spatially varying wall thickness with anatomically relevant layered wall mesh with smooth transition. (C) Multi-domain mesh illustrating the wall and ILT at a cross-section through the AAA sac. (D) Mesh with wedge elements demonstrating the ability to handle nearly horizontal iliac branches.

Table 1

Three-pronged assessment of the AAAMesh framework performance. Study A verifies dimensional accuracy of the surface reconstruction; Study B validates the wall-extrusion process using images with relatively high resolution derived from an AAA phantom. Study C benchmarks AAAMesh against popular alternative tools to create image-based multi-domain finite element meshes from patient-specific CT image data. In the image grid resolution described herein, the first two numbers correspond to the pixel size in the image plane while the third number corresponds to the slice spacing.

	Objective	Model	Image source and grid resolution (mm)	Description
Study A	Surface verification	Analytical 3D surface resembling an aneurysm	Reverse-engineered mask input corresponding to intended surface Resolution: $0.75 \times 0.75 \times 3$	AAAMesh output surface mesh was assessed with respect to analytical surface using the minimum signed perpendicular distance calculated at each mesh node as the metric (negative distance $\rightarrow$ node beneath analytical surface)
Study B	Wall-E validation	Aneurysm phantom (non-uniform wall thickness)	μ CT imaging Resolution: $0.105 \times 0.105 \times 0.7$	Thin-walled silicon elastomer phantom with non-uniform wall thickness manufactured with the lost casting technique (Seong et al., 2005), subjected to μ CT imaging. AAAMesh output mesh using this image input is assessed for local wall thickness against that estimated from the μ CT images at 16 well-spaced locations on surface
Study C	Benchmarking and mesh quality assessment	Patient-specific geometry	Contrast-enhanced CT imaging done at Allegheny General Hospital (Pittsburgh, PA) as per the established clinical protocol for AAA treatment (male, 66 year-old patient). Resolution: $0.80 \times 0.80 \times 3$	AAAMesh, Simpleware, and 3-Matic were employed by experienced users of each tool to get the most suitable, anatomically appropriate, 3D mesh; one each for (i) an assumed 1.5 mm uniform wall thickness (Raghavan et al., 2006) and (ii) the patient-specific wall thickness. The meshes must be suitable for FEA simulations with optimal number of elements chosen based on various options available within each application. Mesh quality metrics for all meshes was assessed using Trelis (Csimsoft, American Fork, UT).

Table 2

Benchmarking the AAAAMesh output with that of the leading 3D image-based reconstruction and meshing tools using identical input masks for the patient-specific model (study C) and the objective of creating anatomically realistic, high quality FE meshes with a reasonable number of elements. The quality metrics (unitless, all have ideal value 1) and their recommended ranges are based on the Trelis user manual (Trelis, 2014). The range of the metrics corresponds to the worst 50 elements; the metrics in bold represent the sliver-free status (Trelis, 2014). Both Simpleware and 3-Matic were unsuccessful in modeling the patient-specific wall thickness without artifacts that distorted the geometry or the meshes resulted in a high (> 500,000) number of elements.

Mesh Quality Metric	Range <Possible> (Recommended)		Uniformly thick wall			Non-uniformly thick wall	
	Hex	Tet	AAAAMesh	3-Matic	Simpleware	AAAAMesh	
Distortion	<0,1> (0.6, 1)	<-1,1> (0.6, 1)	0.3250-0.4067	1.0-1.0	1.0-1.0	0.2951-0.4008	
Scaled Jacobian	<-1,1> (0.5, 1)	<-1,1> (0.2, 1)	0.2330-0.3835	0.0199-0.0461	0.0347-0.0875	0.2150-0.3920	
Shape	<0,1> (0.3,1)	<0,1> (0.2, 1)	0.3354-0.4780	0.1088-0.1342	0.1520-0.2733	0.3449-0.4379	
Condition	<1,inf> (1,8)	<1,inf> (1,3)	1.1403-1.1444	14.061-8.286	1.000-1.004	2.5257-3.8151	
Total number of elements	---	---	64,470	109,419	220,171	64,710	
Other element mesh quality metrics specific to Hex elements							
Shear	<0,1> (0.3,1)	N/A	0.2230-0.3835	---	---	0.2150-0.3920	
Skew	<0,1> (0.0,5)	N/A	0.7092-0.8625	---	---	0.6968-0.8750	
Stretch	<0,1> (0.25,1)	N/A	0.2233-0.3268	---	---	0.2052-0.3107	
Aspect ratio	<1,inf> (1,4)	N/A	2.2141-3.4895	---	---	2.7055-4.1056	