HEMODYNAMICS OF CORONARY ARTERY ANEURYSMS IN KAWASAKI DISEASE —

AN IDEALIZED ANEURYSM MODEL

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# Abstract

Kawasaki Disease (KD), a vasculitis of unknown etiology typically occurring in infants and young children, is the leading cause of acquired heart disease in childhood in developed countries. Approximately 20-25% of patients may eventually develop coronary artery aneurysms (CAAs) if not treated within ten days of onset [1-6]. Abnormal hemodynamics within CAAs may trigger thrombosis, resulting in myocardial infarction and sudden death [3-8]. Although prior studies have investigated the use of hemodynamic parameters derived from patient-specific computational hemodynamic simulations for stratifying patient thrombotic risk, such studies have been limited in cohort size. Here, we present a pipeline for systematic analysis of how aneurysm geometry (i.e. aneurysm diameter, length, and position) affects hemodynamics. The pipeline supports the generation of artificial aneurysms from a baseline patient coronary tree, interfacing with the Simvascular open-source software. The pipeline also provides post-processing and data analysis capabilities. Through this systematic analysis, we elucidate the complementary roles of aneurysm diameter, length, and position on hemodynamic parameters such as time-averaged wall shear stress and residence time. We find that multiple combinations of diameter and length can illicit similar flow behavior. Further, we propose that vessel curvature underlies the effects of aneurysm position. Overall, we demonstrate the promise of systematic investigation of the relationship between aneurysm geometry and flow behavior, a general proof-of-concept for systematically evaluating pathological hemodynamics in the absence of clinical data and assessing hemodynamic parameter sensitivity with respect to patient anatomy.

# Introduction

Kawasaki Disease (KD), a vasculitis of unknown etiology typically occurring in infants and young children, is the leading cause of acquired heart disease in childhood in developed countries. Approximately 20-25% of patients may eventually develop coronary artery aneurysms (CAAs) if not treated within ten days of disease onset [1-6]. Unlike cerebral or abdominal aorta aneurysms, which pose a risk of rupture, CAAs are threatening in their potential for inducing thrombus formation, resulting in myocardial infarction and sudden death [3-8].

CAAs resulting from KD can adopt a wide range of shapes and diameters across a range of locations within the coronary tree [1, 2, 4]. Size classification often relies on maximum aneurysm diameter, with potential normalization based on body surface area (Z-score) to label aneurysms as uninvolved, small, medium, or giant. CAAs may also be roughly classified by shape into groups such as saccular, fusiform, and string-of-pearls. A final classification is based on the number of major coronary vessels which bear aneurysms [1, 2]. Coupled with interpatient variability in hemodynamic factors such as blood pressure and cardiac output, wide variability in aneurysm diameter, shape, and position amplifies difficulty in isolating the effects of geometric parameters. Indeed, complex geometry, as well as aneurysm shape, number, and location, may all contribute to abnormal hemodynamics and correlate with patient outcome.

Presently, the American Heart Association offers guidelines for stratification of patient thrombotic risk that rely on aneurysm diameter alone as the criterion for initiating systemic anticoagulation; however studies suggest risk may also depend on hemodynamic parameters such as time-averaged wall shear stress (TAWSS), residence time (RT), and oscillatory shear index (OSI), which are not available through image data alone [9-13]. Patient-specific hemodynamic simulations can non-invasively supply these informative hemodynamic parameters for better thrombotic risk assessment; however, limited patient cohort size impedes understanding of the variability of hemodynamic parameters with respect to patient anatomy and aneurysm geometry [10-13].

With such limitations in mind, existing patient-specific models can be augmented by introducing artificial aneurysms of specified length and diameter, to achieve a systematic evaluation of relationship between CAA shape, diameter, and position on local hemodynamics. Ultimately, elucidating the relationship between hemodynamics and aneurysmal shape characteristics may underlie more powerful risk stratification methods to support clinical decision-making regarding initiation of anticoagulation therapy.

# Methodology

## Idealized Aneurysm Models

A pipeline for generation of three-dimensional idealized aneurysm models has been constructed with dependencies on the Visualization ToolKit Package (VTK). The pipeline workflow consists of the following three steps.

***(1) Baseline Model***An existing patient-specific model of the aorta and coronary vasculature was constructed from coronary CT image data in Simvascular, an open source software which provides a full pipeline for patient-specific cardiovascular simulation [14]. This baseline model is constructed through manual identification of preliminary vessel centerlines, 2D segmentation of vessel contours, and lofting contours to form vessel walls. The union of all vessel walls is taken as the three-dimensional model, represented as a triangular mesh in 3D space.

***(2) Surface Parameterization*** Vessel centerlines are re-extracted from the three-dimensional model in Simvascular, then upsampled from cubic spline interpolation to achieve sufficient spatial resolution in the centerline’s discretization. Points from the 3D model’s wall are expressed in terms of distance along the centerline, , by mapping each wall point to the resampled centerline point that minimizes distance. The Frenet-Serret reference frame is then constructed from the centerline to label each wall point with its angle with respect to the centerline’s unit normal . This combination of position and angle with respect to the centerline enables efficient generation of artificial aneurysms.

***(3) Aneurysm Generation*** Existing AHA aneurysm classification for small, medium, and giant CAAs serves as a guide to proceed from body surface area-normalized Z-score to an estimated maximal diameter for a given coronary vessel. The Z-score formulation presented in Dallaire and Dahdah et al. 2011 [15],

with coefficients estimated separately for each vessel, is used to isolate , the maximal aneurysm diameter. Along with aneurysmal shape index (ASI), defined as the ratio of aneurysmal length to maximal lumen diameter,

and distributions of clinically measured ASI values, we obtain a range of lengths and diameters to guide aneurysm generation [13].

For a given target length and maximum diameter, we use the position and angle parameterizations determined in step *(2)* to deform the vessel wall. Deformation magnitude is given through second-order smooth interpolation of diameter as a function of centerline position, producing radial symmetry. In the case of vessel branching from the desired aneurysm region, all points of the bifurcating vessel and its downstream branches are shifted and rotated according to the average deformation at the bifurcation. Empirically, this method has worked optimally to control bifurcation angle as artificial aneurysms vary in shape and maximal diameter. This process was applied for multiple positions, shapes, and diameters, as exemplified in **Figure 1**.

## Computational Hemodynamics

Generated aneurysm models are further processed in Simvascular using Tetgen, an open source package for mesh generation included in Simvascular [14, 16]. Tetrahedral finite element meshes with boundary layer meshing are generated for each model. The Simvascular solver then computes a numerical solution to the time-dependent Navier-Stokes equations governing blood flow. Blood is modeled as an incompressible Newtonian fluid (density=1.06 g/cc, dynamic viscosity=0.04 dynes/cm2) and walls are assumed to be rigid in all cases.

***Simulation Boundary Conditions*** Systolic myocardial contraction increases intra-myocardial pressure, transiently increasing distal coronary resistance substantially and causing coronary flow to be out of phase with systemic flow. Specialized boundary conditions coupling intra-myocardial pressure to coronary flow are required to replicate this complex physiology in the numerical model. We achieve this by modeling the heart and distal vasculature with a closed-loop, Lumped Parameter Network (LPN), which imposes boundary conditions with tunable parameters to produce physiologically accurate cardiac output, heart rate, blood pressure, and flow distributions. For all idealized models described in this work, LPN parameters are fixed to better isolate the effects of aneurysmal geometry on hemodynamics. **Figures 1** shows the pipeline, proceeding from baseline model to artificial aneurysms and simulation results over isolated aneurysm regions.

## Residence Time Calculation

Although pathways underlying platelet activation and thrombosis remain poorly understood, flow recirculation and stagnation have been hypothesized to contribute to thrombosis [17, 18]. To better understand aneurysm hemodynamics, we can employ Residence Time (RT), which has been shown to be an effective measure of quantifying recirculating flow that traps fluid for an extended duration. RT is computed as described in Esmaily-Moghadam et al. 2013, by first solving the advection-diffusion equation [17]:

where is a measure of time, is the velocity field from solving the Navier-Stokes equation; is the diffusivity, which is set to 0; is the particle source term defined as 1 inside the region of interest, , and 0 otherwise. RT is then computed as:

## Post-Simulation Analysis

In order to quantify local aneurysm hemodynamics we isolate the aneurysm from the larger coronary tree. Aggregate measures of certain hemodynamic quantities, such as WSS, can then be computed either over the aneurysm surface, or over the aneurysm volume. Quantities of interest can also be computed using spatial average over the domain and temporal average over the cardiac cycle. The post-simulation analysis pipeline — including thresholding operations, variable integration, and simulation data visualization — was developed using custom scripts interfacing with Paraview (<https://www.paraview.org/>), an open-source data-analysis and visualization application.

# Results

Idealized aneurysms of 3 representative shape index values (ASI = 2, 4, 6) were generated for 5 diameters (z-score = 6, 8, 10, 12, 14) at three positions along the right coronary artery (RCA) and one position in the left anterior descending (LAD) for a total of 40 cases. Hemodynamic simulation results were isolated over aneurysmal regions to identify the effects of shape, diameter, and position on local hemodynamic conditions.

## Hemodynamic Variations with Shape and Diameter

Hemodynamic parameter distributions at the vessel wall have been hypothesized to be an effective way to assess aneurysm hemodynamics. Broadly, we expect similarly shaped aneurysms to give rise to similar hemodynamic behaviors; further, we expect that increases in Z-score correspond to decreases in fluid velocity and the potential for turbulence or recirculation. Indeed, these expectations are reflected qualitatively in distributions of TAWSS over the vessel surface (**Figure 1**). We observe that surface hemodynamic patterns vary consistently with respect to increasing Z-score, with overall decrease in TAWSS as the diameter increases. Additionally, for each value of ASI, aneurysms appear to bear similar spatial distributions of TAWSS.

To understand hemodynamic variations with geometric parameters, distributions of hemodynamic parameters are quantified in an aggregate manner, revealing that multiple combinations of aneurysm shape and diameter can produce similar hemodynamics. First, the average TAWSS over each aneurysm surface is computed and plotted with respect to ASI, stratified by aneurysm Z-score (**Figure 2**). As shape index increases (i.e. more elongated aneurysms), and as Z-score increases, average TAWSS declines. While in the LAD, the lowest values appear in the longest aneurysms of largest diameter, with relatively steep negative trend overall, average values in the RCA remain relatively similar; the lowest values are observed in aneurysms with ASI = 4.

The fractional aneurysm surface area exposed to TAWSS values less than a critical threshold dserves as another aggregate measure that has been proposed for evaluating CAA hemodynamics and stratifying patient risk [10-13]. A plot of fractional area exposed to low TAWSS as a function of ASI, again stratified by aneurysm Z-score is also given in **Figure 2**. We observe that for aneurysms in both the RCA and LAD, the fractional area under 1 dyne/cm2 increases as either of Z-score or ASI increase. Notably, the longest aneurysms of moderate Z-score (ASI = 6, Z-score = 8) produce similar values to shorter aneurysms of largest Z-score (ASI = 2, Z=score = 14). As with average TAWSS, multiple combinations of aneurysm shape and diameter can produce similar hemodynamics.

Residence Time (RT1), the time a parcel of fluid spends within an aneurysm, can also be used to understand CAA hemodynamics. We observe that RT1 varies consistently with respect to aneurysm shape and diameter (**Figure 3**). Increased RT1 relative to the baseline, computed in the original vessel without artificial aneurysms, indicates that all aneurysms regardless of shape and diameter exhibit fluid accumulation, likely due to recirculation and stagnation. While the small aneurysm diameter (Z-score = 6) produces little variation in RT1 as the aneurysm lengthens, higher values of Z-score tend to also magnify the effects of increasing aneurysm length. As with average TAWSS and fractional TAWSS-exposed area, the relationship between RT1 and aneurysm geometry indicates that multiple combinations of aneurysm shape and diameter can illicit similar hemodynamic behavior.

## Aneurysm Position and Hemodynamics

Aneurysms generated in proximal, medial, and distal positions along the RCA enable comparison of hemodynamics as a function of position. Computing time-dependent flow rate into each aneurysm indicates that for each position, flow into the aneurysm is independent of diameter, but decreases with position along the centerline due to the presence of additional branches diverting blood flow (**Figure 6**). Despite variation in flow rate, **Figure 4** depicts similar levels of TAWSS in aneurysms of the same shape (ASI = 2). Proximal RCA aneurysms exhibit comparatively little variation in average TAWSS as Z-score increases, relative to medial and distal locations. Average TAWSS increases in medial aneurysms relative to proximal and distal positions, though fractional surface area exposed to low TAWSS exhibits little variation with respect to aneurysm geometry.

While average TAWSS and fractional TAWSS-exposed area exhibited inconsistent variation with respect to aneurysm position, RT1 increases consistently with respect to position (**Figure 3**). In particular, the largest, most distal aneurysm (Z-score = 14) has significantly increased RT1 relative to all other aneurysms of the same shape. **Figure 6** indicates that flow rates are identical for aneurysms in the same position regardless of diameter; especially given low variation in baseline RT1 with position, inlet flow rate differences fail to explain increased RT1 distally, as well as changes in RT1 with increased aneurysm diameter. This suggests that increased aneurysm diameter promotes pronounced recirculation in distal aneurysms relative to proximal or medial positions, enabling greater variation in RT1 with respect to aneurysm diameter.

To further investigate hemodynamic mechanisms underlying the non-linear relationship between average TAWSS and aneurysm position, we investigated average WSS over the cardiac cycle. We observe that in proximal and medial aneurysms of ASI=2 in the RCA, values of average WSS are ranked in decreasing order by Z-score over the cardiac cycle (**Figure 4**). However, in aneurysms in the distal RCA, intermediate values of Z-score (8, 10, 12) correspond to consistently increased values of average WSS compared to both low and high values (Z-score = 6, 14) throughout much of the cardiac cycle. This behavior can be understood through visualization of fluid velocity within the aneurysm.

Velocity streamlines through aneurysm cross sections reveal that inflow jet through the aneurysm expansion produces different impingement behaviors against the vessel wall (**Figure 5**). Aneurysms of the same position feature similar inflow jet patterns, producing the similar surface distributions of TAWSS as seen in **Figure 1**. In proximal and medial cases, increases in Z-score did not significantly alter inflow jet impingement area; however, in the distal cases, increases in Z-score alter the angle of the inflow jet, resulting in differing patterns of recirculation. These changes correspond to the WSS trends observed in **Figures 2, 4** and the RT1 trends observed in **Figure 3**. Further, these inflow jet patterns explain how average TAWSS can increase in medial aneurysms without altering the fractional surface area exposed to low TAWSS (**Figure 4**).

Examining average TAWSS, fractional TAWSS-exposed area, and RT1 indicates that aneurysm diameter, shape, and position jointly determine aneurysm hemodynamics. Further, we find that aneurysm position influences hemodynamics by influencing inflow jet patterns, suggesting that local vessel curvature may be an effective low-dimensional predictor of hemodynamic behavior.

# Discussion

An extensive pre- and post-processing pipeline was developed from scratch for systematic generation of idealized aneurysms and analysis of simulation results. Our idealized aneurysm models indicate that Z-score remains a strong predictor of hemodynamic behavior. We can still see the role of diameter – there are consistent trends in average TAWSS and RT1 with respect to Z-score at each level of ASI (**Figures 1-6**). However, Z-score alone is insufficient to determine hemodynamic behavior – shape and position, also, are influential (**Figures 2-4**). Further, we highlight the potential for vessel curvature as an additional geometric parameter influencing hemodynamics (**Figure 5**).

Influence of aneurysm geometry and position on hemodynamic parameters demonstrates potential for low-dimensional predictors of thrombotic risk in KD patients. In previous work, the fractional area of the aneurysm surface exposed to low TAWSS has been used to construct a decision boundary for KD patient risk classification more predictive of thrombosis than aneurysm diameter alone [10, 11]. The distribution of TAWSS over each aneurysm surface, then, may serve as a potential surrogate for hemodynamics within the aneurysm influencing thrombosis likelihood. Indeed, we find that aneurysm shape characteristics strongly influence surface distributions of TAWSS and other measures of aneurysm hemodynamics (**Figure 2**). Aneurysms with constant diameter but with different aspect ratios, or with constant aspect ratio and varying diameter, furnish consistently varying hemodynamic environments. These patterns in variation suggest that a combination of shape parameters may be sufficiently predictive of aneurysm hemodynamics, potentially reducing need for computationally expensive 3D simulations.

Similarity in hemodynamic parameters under multiple combinations of aneurysm shape and diameter inform the use of hemodynamic parameters for clinical decision support. Given that aneurysms with different combinations of shape and diameter may achieve identical hemodynamic parameters, we infer that the prior effectiveness of TAWSS-thresholded area for patient risk stratification lies in its ability to capture critical hemodynamic features that are not strictly dependent on aneurysm shape or diameter alone. In comparison, averaging TAWSS over the surface of the aneurysm also affords a single-dimensional summary of aneurysm hemodynamics, but struggles to capture nuances of shear variation over time and sacrifices knowledge of spatial distribution (**Figures 2, 4, 5**). This suggests that utilization of average TAWSS for clinical risk stratification must be contextualized by the hemodynamic features lost during temporal and spatial averaging.

We also investigated changes in RT1 as aneurysm shape, diameter, and position vary. We find that RT1 increases consistently as Z-score, shape index increase, and as position becomes more distal. A relationship between residence time and thrombosis has been hypothesized for both cerebral and coronary aneurysms [13, 17, 18]. Here, however, RT1 increases intuitively with both aneurysm length and diameter. Given relatively well-developed inlet jets as illustrated in **Figure 5** that may be unrealistic in true patient anatomies, it is difficult to determine whether these trends are a characteristic of the RT1 parameter, or whether geometric similarity between all smooth, symmetric idealized aneurysms in this study inhibits analysis of RT1’s ability to quantify nuanced hemodynamics.

Although we have systematically investigated the role of geometric parameters on aggregate measures of aneurysm hemodynamics, further work should continue to investigate the potential for low-dimensional representations of aneurysm geometry towards predicting CAA hemodynamics as a surrogate for improving clinical predictive value. Known correlations between aneurysm hemodynamic and geometric features suggest potential to link clinical measurements easily obtained from echocardiography or other routine imaging modalities with patient outcome. Such approaches may form the basis for more sophisticated geometry-based risk stratification methods supporting clinical decision- making in assessment of KD patients.

Further work may also investigate fluid-solid interaction simulations. We assumed that hemodynamics secondary to aneurysm anatomy could be captured sufficiently accurately with rigid wall simulations, with local hemodynamic changes due to vessel wall deformation playing only a marginal role. However, investigating the characteristics of wall deformation in CAAs may also guide understanding of thrombosis, improve physiological realism, and produce clinically relevant hemodynamic parameters. Indeed, this framework for systematic variation of aneurysm geometry may be easily coupled with variation in wall material properties to facilitate further study of aneurysm hemodynamics and wall biomechanics.

We demonstrate the potential for modification of patient-specific vascular models and artificial aneurysm generation for systematic evaluation of the relationship between anatomy and hemodynamics. Extending the methodology in [10, 11], we first illustrate that controlled variation in aneurysm shape, diameter, and position may produce predictable variation in hemodynamic parameters. Indeed, assumptions such as axisymmetric aneurysm radius, high degree of surface smoothness, and single aneurysm per vessel limit clinical realism. Even with these simplifying assumptions, we demonstrate that manipulating key geometric parameters such as length, diameter, and position are sufficient to enable systematic variation of aneurysm hemodynamics, allowing closer interrogation of inconsistent variations in hemodynamics. Overall, the use of vessel modification pipeline for systematic understanding of CAA hemodynamics in KD patients forms the foundation for future systematic hemodynamic studies that may inspire insight into disease pathogenesis, aid validation of patient-specific risk stratification methods, and help characterize cardiovascular surgical interventions.

# Acknowledgements

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