HEMODYNAMIC VORTEX ANALYSIS AS A MEANS OF INTRACRANIAL ANEURYSM RUPTURE PREDICTION

By

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Dedication

To my famliy and friends

who

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Preface

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Acknowledgments

I would like to thank all the members of my committee and my advisor Dr. Jingfeng Jiang. Their leadership, support, knowledge and motivation not only helped me throughout my research, but helped drive me to become a better scientist and to never stop learning.

Special thanks are also needed for Dr. Autumn Schumacher, who was willing to take a gamble on a brand new scientist fresh out of their undergraduate education. Her and expertise (and many hours of manuscript editing) were invaluable in getting me to where I am today.

I would also like to thank my friends for their boundless confidence in me which helped push me through my PhD work. Last but not the least, I would of course like to thank my family. All of their love and support helped make this thesis possible.

Definitions

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List of Abbreviations

ACA	Anterior Communicating Artery
AFI	Aneurysm Formation Indicator
CFD	Computational Fluid Dynamics
DICOM	Digital Imaging and Communications in Medicine
DVO	Degree of Volume Overlap
ENR	Elastic Net Regression
IA	Intracranial Aneurysm
ICA	Internal Carotid Artery
MCA	Middle Cerebral Artery
MLR	Multiple Logistic Regression
NSC	Nearest Shrunken Centroid
OSI	Oscillatory Shear Index
PC-MRI	Phase Contrast Magnetic Resonance Imaging
ROC	Receiver Operator Characteristic
STA-WSS	Spatiotemporaly Averaged Wall Shear Stress
TA-WSS	Temporally Averaged Wall Shear Stress
VMTK	Vascular Modeling Toolkit
VTK	Visualization Toolkit

WSS	Wall Shear Stress
WSSG	Wall Shear Stress Gradient
λ_2	Lambda$_2$
ACL	Access Control List
AIB	Add-In Board
ALE	Arbitrary Lagrangian Eulerian
AMANDA	Advanced Maryland Automatic Network Disk Archiver
AMBER	Assisted Model Building with Energy Replacement
AMD	Advanced Micro Devices
AMOLED	Active-Matrix Organic Light Emitting Diode
AMPI	Adaptive Message Passing Interface
ANL	Argonne National Laboratory
API	Application Program Interface
ASCII	American Standard Code for Information Interchange
ATLAS	Automatically Tuned Linear Algebra Software
b_eff	effective bandwidth Benchmark
BIOS	Basic Input/Output Operating System
BLAS	Basic Linear Algebra Subprograms
BOMD	Born-Oppenheimer Molecular Dynamics
BP	Bootstrap Protocol
CCSR	Center for Computer Systems Research

CentOS	Community enterprise Operating System
CFD	Computational Fluid Dynamics
CHARMM	Chemistry at HARvard Macromolecular Mechanics
CHAMBER	CHarmm ↔ AMBER
CMake	Cross Platform Make
CODINE	Computing in Distributed Networked Environments
CP2K	Car-Parrinello 2000
CPMD	Car-Parrinello Molecular Dynamics
CPU	Central Processing Unit
CSS	Central Security Service
CTM	Chemical Transport Model
CUDA	Compute Unified Device Architecture
CUDPP	CUDA Data-Parallel Primitives Library
DAE	Differential Algebraic Equation
DARPA	Defense Advanced Research Projects Agency
DAE	Delay Differential Equation
DFT	Discrete Fourier Transform
DFT	Density Functional Theory
DGEMM	Double Precision GEneralized Matrix Multiplication
DHCP	Dynamic Host Configuration Protocol
DMCA	Digital Millennial Copyright Act

DOD	Department of Defense
DOE	Department of Energy
DRM	Distributed Resource Manager
DRMAA	Distributed Resource Manager Application API
EFF	Electron Force Field
EVL	Electronic Visualization Laboratory
FCA	Fabric Collectives Accelerator
FEA	Finite Element Analysis
FFT	Fast Fourier Transform
FFTW	Fastest Fourier Transform in the West
FLOPS	Floating Point OPerations per Second
FPU	Floating Point Unit
FSI	Fluid Structure Interaction
FTDT	Finite Difference Time Domain
FTP	File Transfer Protocol

Abstract

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Chapter 1

Introduction

Subarachnoid hemorrhage is a potentially devastating pathologic condition in which bleeding occurs into the space surrounding the brain. One of the prevalent events that result in subarachnoid hemorrhage is the rupture of an intracranial aneurysm (IA). IAs are degenerative, irregular expansions of areas in the cerebral vasculature that occur in an estimated 3-5% of the global population [100, 198, 247], with an estimated 0.15% - 0.7% of the global population suffering the rupture of an IA each year [114]. Mortality rates due to the rupture of IAs are estimated between 45-50%, with survivors suffering significant neurological damage, including physical and cognitive impairment [156, 247]. Yet the disparity between the number of ruptured and unruptured IAs indicate that not all IAs are at high rupture risk. From a clinical perspective, improved medical imaging techniques have led to an increase in the detection of unruptured IAs, and novel surgical intervention methods have aimed to reduce the instances of IA rupture and their subsequent impacts of patient health [140, 175]. Yet surgical treatments are not without a high healthcare costs: surgical repair (without complications) of IAs can be an estimated $25,000. Data has also shown that, while treatment risks are relatively low, similar outcomes of morbidity and mortality can occur in the event of treatment complications [45, 153, 164]. Surgical complications also result in a marked increase in costs: between an estimated $35,000 - $70,000 depending on possible complications (varies with surgical intervention) [28]. Additionally, continued economic burden is placed upon patients and their families if long term rehabilitation, home care, or in-hospital treatment is required post-surgical complications. This surgical vs rupture risk has led to significant interest in elucidating the processes and conditions which impact IA rupture. Proper identification of IAs at high risk of rupture could significantly benefit risk stratification, improve patient selection and aid treatment planning as to best determine a course of action for the clinical management of IAs.

Current Rupture Prediction Metrics

Research has shown that a wide array of risk factors may impact IA development, growth and rupture potential [67, 141, 188, 220], with said factors generally separated into three categories:

– IA morphological characteristics

– Vascular hemodynamics

– Patient genetic and health factors

Morphological characteristics of IAs are often first considered when determining the severity of an unruptured IA and its potential to rupture. Factors such as IA size (volume), shape, aspect ratio, IA to parent vessel angles, etc. NEEEEED REFERENCES have been identified in a number of studies as a possible metrics to assess IA rupture potential. However, many of these parameters have been shown to generate conflicting results between studies, varying in their strengths toward rupture prediction. As per example, IAs >10mm in size are often associated with rupture risk, with IAs >25mm thought to be at the greatest rupture risk leading the majority of IAs at these sizes to undergo surgical intervention[173, 258, 264] even though not all large IAs rupture. If this thought process would be applied clinically, many small IAs (<4mm) may be spared surgical intervention due to the thought that the complications from surgical interventions would be of greater risk than the possibility of small IA rupture. Yet small aneurysms have been shown to be at a non-insignificant risk of rupture [78]. In addition to assessing rupture risk solely on IA size may overestimate the risk of large IAs (and subsequently underestimating small IAs), size characteristics may be of limited usefulness in assessing rupture risk in medium-sized IAs (4-10mm). Additionally, may of the other morphological characteristics of IAs have had varying degrees of strength toward predicting IA rupture[10, 29, 124]. To overcome the limitations of relying on morphological characteristics for assessment of IA rupture potential, hemodynamic characteristics within the IA sac have been investigated.

Hemodynamic stressors are thought to play a significant role on the initiation, growth and possible rupture of IAs. Tangental fluid stressors along the vascular wall, known as wall shear stress (WSS), and its derivatives are often of investigated when generating models for IA rupture prediction(s)[38, 68, 124, 264]. These near-wall forces have been shown to act as a biological stimulator, eliciting various changes to both vasculature endothelial and smooth muscle cells, inducing changes in gene expression, phenotypical and mechanical properties, and protein expressions which dictate cellular activity [13, 15, 32, 41, 69, 148, 202, 241]. While WSS has been assessed to trigger vascular cell changes, discrepancies exist to which specific hemodynamic characteristics impact IA rupture prediction and their overall predictive strength, similar to IA morphological characteristics. It has been shown that vascular cells maintain healthy physiological characteristics while exposed to a range of WSS between 5 and 20 dynes/cm^2 yet studies are in conflict if WSS lower [27, 172] or higher [69, 213] from said preferred range is of greater impact on IA rupture risk. It has been suggested that both high and low WSS trigger differing impacts on vascular cells [169], meaning both may play a significant role in IA development and potential rupture. Yet even focusing on one extrema of WSS values and determining their usefulness on prediction

IA rupture risk has proven difficult. In a 2016 meta analysis by Zhou, the impact of low wall shear stress on predicting IA rupture varied between studies (Fig. **??**). In addition to WSS extrema as a predictive metric toward IA rupture, fluctuations in WSS gradients and changes in WSS directionality (Oscillatory shear index (OSI)) are thought to cause pathologic changes to the vasculature are often applied to IA rupture prediction models. Its has been theorized that fluctuations in WSS directionality can trigger varying changes to vascular cells, weakening the overall strength of the vasculature and triggering possible IA development or rupture.

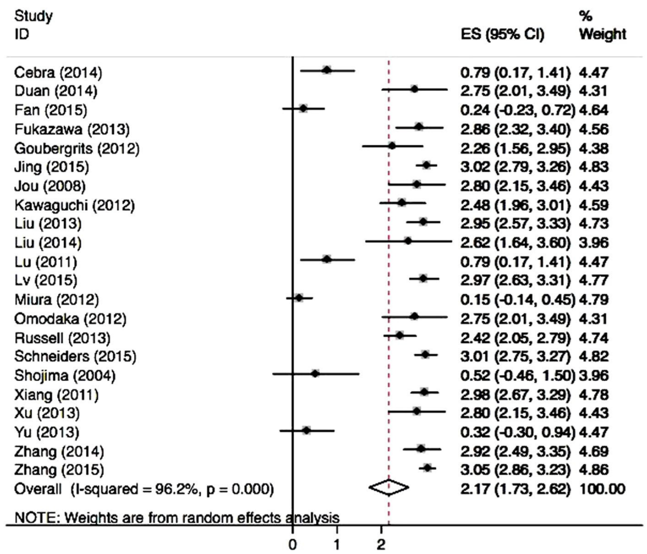

Figure 1.1: Zhou 2016 Meta-analysis of the reported low WSS rate of rupture aneurysms and the Odds Ratio for low WSS in predictive modeling

It is worth noting, that the clinical measurement of WSS characteristics *in-vivo* is possible, yet remains difficult especially in smaller blood vessels and small IAs. Improvements in computational fluid dynamics (CFD) can help overcome the limitations of *in-vivo* WSS measurement and is being adopted as a clinical tool for assessment of rupture potential and planning of IA treatments [222, 240].

While the development of IAs have been related to hemodynamic factors, a number of genetic and patient health-factors have also been associated with structural weakness in the arterial wall indicative of IAs[174, 229]. The increased liklehood of IA development in a person whose family member(s) have also presented with an IA gave early insight towards a possible genetic component to IAs.

A number of patient health factors have also been associated with IA development/rupture and have been incorporated into prediction models. Patient age, gender and smoking habits, which have bee linked with changes to the mechanical properties of the vasculature, have been incorportated into IA rupture prediction [10, 29, 166]s. Increases in age and smoking[11, 97, 141, 167] have been linked with a breakdown of the collagen of the vasculature, weakening the overall mechanical strength of the vasculature. There is also a breakdown in the elastin complexes of the vasculature as a function of aging, decreasing the compliance of the vessel (to hemodynamic conditions) and increasing vessel stiffness[79, 252]. Additionally, hypertension has also been linked with an increase in the likelihood of aneurysm development and potential rupture[193, 226]. Hypertension is thought to weaken vessels by increasing mechanical stresses while also causing a subsequent increase in vascular inflammation, leading to possible remodeling of the vascular wall[207].

Combining the assessment of IA morphological characteristics, WSS and genetic/patient medical information have been applied to a number of models aiming to predict the likelyhood of IA rupture. Yet the indices chosen and the resultant strength of predictions models can vary significantly between studies (1.1).

Table 1.1

Recent articles for aneurysm rupture prediction

Author-Year	Parameters for Analysis	Aneurysm Location(s)	Ruptured IAs	Unruptured IAs	AUC or Accuracies
Qin et al., 2017[194]	W, H, L, NW, Age, AR, Dmax, HW, BF, WSS, LSA, EL	MCA	36	31	0.931
Bijlenga et al., 2017[25]	S, Age, Hyp, Race, SAH	MCA, ICA, ACA/PcomA/-Post	598	243	0.681-0.756
Detmer et al., 2018[65]	Age, Gender, S, OSIm, NSI	ACA, AcoM, BA, ICA, MCA, PCom, VA	66	183	0.82
Kocur et al., 2019[136]	S, H, W, NS, AR, BF, H/W, SR, A	AcoM, ICA, MCA, Post	146	285	0.55-0.64
Wang et al., 2019[250]	R, DD, NW, W, D, Dia, AR, DW, BF, SR, F, LD, SD,	AcoM	214	147	0.846
Varble et al., 2018[245]	Age, SR, AR, UI, EL, NSI, WSS, OSI, WSSG, RRT, LSA, MWSS, PLc, EL	ACA, ICA, MCA, PcoM, Post	102 (Train) 14 (Test)	311 (Train) 115 (Test)	0.767
Jiang et al., 2018[123]	L, H, AR, SR, UI, EI, D, NSI, NWSSa, WSSm, WSSa, NWSSa, NWSSm, WSSG, LSA, OSI, RRT, Pm, Pa, NPa, NPm	ACA, AcoM, ICA, MCA, PcoM	167	167	0.81
Anertieor communicating artery (AcoM), Anterior communicating artery (ACA), Internal carotid artery (ICA), Middle cerebral artery (MCA), Posterior communicating artery (PcoM), Posterior inferior cerebral artery (Post), Aneurysm angle (A), Area (A), Aspect Ratio (AR), Bottleneck factor (BF), depth/width (DW), Diameter (D), Energy Loss (EL), Flow angle (F), Height (H), Hypertension (Hyp), Irregular shape (IR), Large daughter artery angle (LD), Lateral angle ratio (LAR), Length (L), Low wall shear area (LSA), Mean diameter (MD), Mean wall shear stress (MWSS), Non-sphericity index (NSI), Normalized pressure average (NPa), Normalized pressure max (NPm), Normalized wall shear stress area (NWSSa), Normalized wall shear stress mean (NWSSm), Oscillatory shear index(max) (OSIm), Pressure average (Pa), Pressure loss(coefficient) (PLc), Pressure max (Pm), Relative residence time (RRT), Size ratio (SR), Small daughter artery angle (SD), Surface (S), Temporally averaged WSS (TAWSS), Undulation Index (UI), Vessel angle (V), Width (W), Wall shear stress (WSS), Wall shear stress gradient (WSSG), Wormsley number (WN), Wall shear stress mean (WSSm).					

These variations between study outcomes suggest that novel assessment methods may help improve the understanding of IA characteristics indicative of rupture. Of recent attention are the assessments of broader flow patterns occurring within the aneurysm sac and parent vessel for their relationship to aneurysms. While WSS characteristics assess the 'near-wall' flow within the vasculature, this overlooks the reaming flow characteristics within the vasculature that may help improve prediction models (Fig. 1.2).

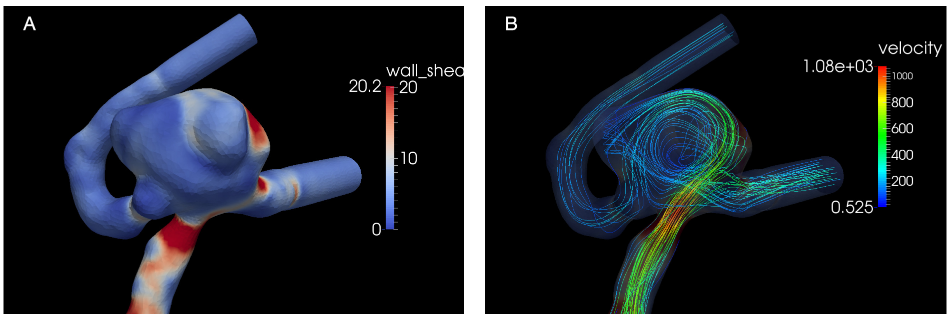

Figure 1.2: The wall shear stress (N/mm^2) distribution on the aneurysm wall(A) and blood flow pattern inside the aneurysm visualized with streamlines (mm/sec)(B). Wall shear stressors may overlook the disturbed/swirling flow patterns (white arrow) that are generated in the IA sac.

Early investigations into the nature of IAs uncovered the non-random distribution of areas of the vasculature susceptible to IA development: vessel bifurcations and areas of significant vasculature curvature (Fig **??**). These areas of likely aneurysm development were shown to correlate with areas of disturbed, swirling hemodynamic flow patterns (vortex/vortices). In much of the vasculature, blood maintains laminar flow characteristics in which fluid travels smoothly in regular paths with little to no mixing. Vortices are antithetical to laminar flow, with areas of significant mixing/recirculation, and causing a significant drop in WSS values. Not only are said vortex patterns present in specific areas of the vasculature, they are also present within IA sacs. These findings led research toward understanding the impact that these vortices may have on altering the vasculature, triggering IA development and rupture[39, 53, 155]. As hemodynamic forces help regulate the physiological characteristics of the vasculature, vortex flow conditions are thought to impart pathologic cellular changes to vascular cells.

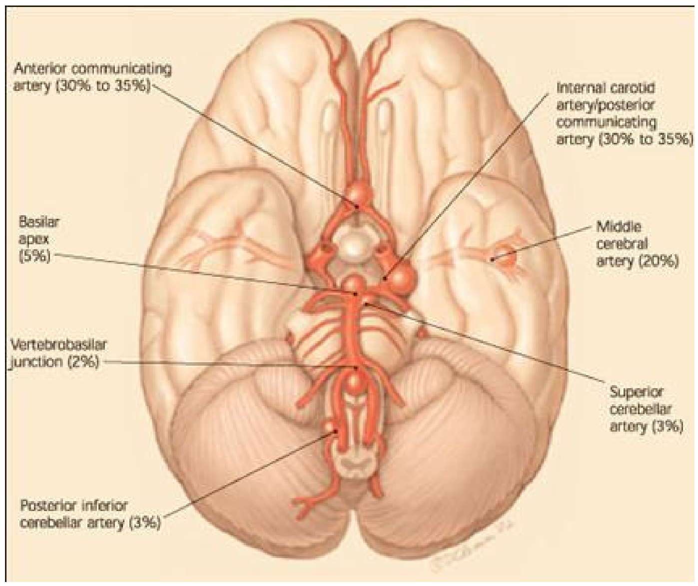

Figure 1.3: Areas of likely cerebral IA development: Original image from [209].

Yet the analysis of disturbed flow is typically done from a visual qualitative assessment of computational fluid dynamic (CFD) based flow patterns or through limited computational analysis of a vortex's centroid region. These methodoligies can only result in a restrictive assessment of the characteristics of vortices, and could limit their usefulness in predicting IA rupture. As per example, a study by Varble et al[244]. was unable to find a strong correlation between vortices and IA rupture, but said study only focused on identifying the centroid region of vortices. A more comprehensive analysis of vortices may give novel insight towards IAs and could provide a useful tool to help improve the assessment of IA rupture potential.

The **focus of this thesis** is to apply novel computational technique to assess the spatiotemporal characteristics of broader structure of hemodynamic vortex patterns and determine their usefulness in identifying cerebral IAs rupture risk.

While a number of metrics (geometric, hemodynamic, and health factors) have aided in understanding possible mechanisms triggering IA development and rupture, I Investigations of hemodynamic vortices in both *in-vivo* and *in-vivo* and have been shown to elicit endothelial cell dysfunction [17, 160, 180].

Section 1

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Objective

Although there exists a number of studies[34, 245, 277] and methodologies[83, 97] that attempt to assess IAs at a high risk of rupture, inconsistencies between study outcomes leave the development of an ideal predictive model out of reach. In addition, many of these previous studies assess the geometric[2, 130, 245] and/or hemodynamic wall stressors[34, 172, 277] as a means to predict IA rupture, with limited quantitative assessment of the hemodynamic flow conditions within the aneurysm. **The primary objective** of this work is to assess the viability of adapting quantitative analysis of hemodynamic flow patterns, specifically swirling flow pattern(s) (vortex), within IAs to improve the prediction and understanding of IA rupture. In this work, an overview of recent theories concerning

Methodolgy

For the initial focus of this work, image-based computational fluid dynamics models of patient-specific IA geometry will be constructed from 3D phase contrast magnetic resonance imaging (PC-MRI). Computational fluid dynamic (CFD) simulations will be performed on the computational models to generate realistic 3D hemodyanmic velocity and flow pattern data. From said data,

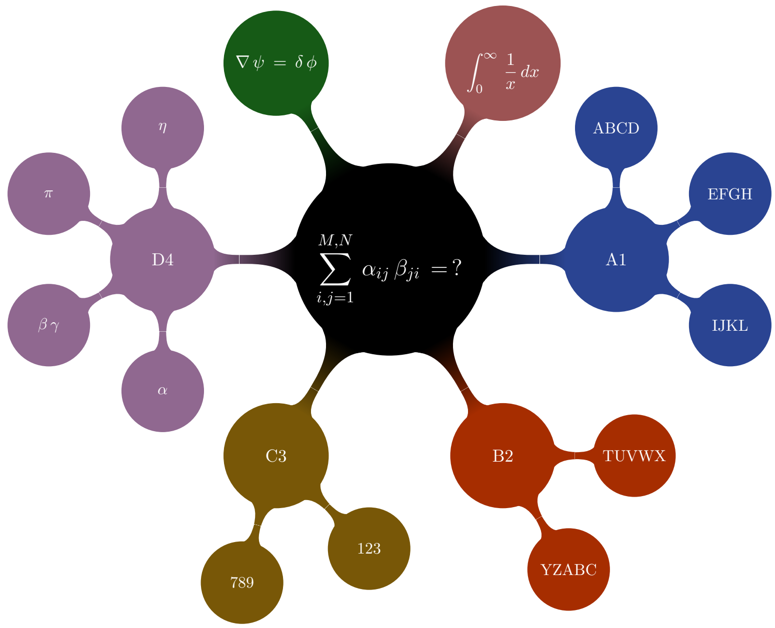

Figure 1.4: Schematic representation of our universe

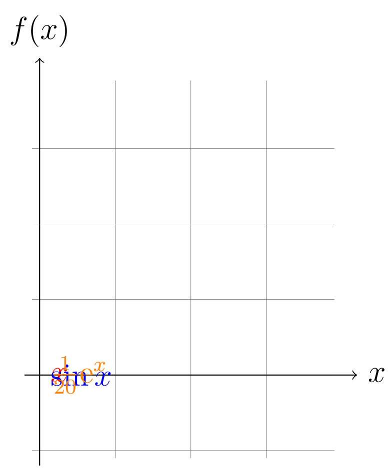

Figure 1.5: Mathematical functions plotted using TikZ package

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Aneurysm Geometic Characterisitcs

All aneurysm geometries were taken from the finalized computational mesh generated for simulations. The aneurysm sac was manually isolated from the parent vessel and the resultant cut plane was capped and identified as the IA ostium using an in-house script written in VMTK. Geometric measurements were either taken directly from the values reported in the Aneurisk dataset, or were calculated using in-house scripts in VMTK.

Aneurysm Surface Area and Volume: Measured directly from the isolated IA geometry before and after (respectively) ostium capping. A number of studies have eluded to an increase in IA size as a risk for both IA growth and rupture. [11, 29, 97, 245]. A meta-analysis performed by Brinjikji et al reported that IA $\leq$ 10 mm in size (diameter) grew at a rate $<$ 2.9% per year, while IAs $>$ 10 mm were associated with growth rates of 9.7% per year. This growth was also reported with an associated IA rupture rate: 3.1% per year compared with 0.1% per year for stable (non-growing) aneurysms ($p \leq 01$). From a clinical perspective, the overall size of an aneurysm is often a characteristic used to determine course of IA treatment (or lack thereof) [140, 259]. Yet while large IAs are thought to increase the likelihood of rupture, a not-insignificant number of small IAs ($<$5 mm diameter) also have been shown to rupture [126, 130, 141]. This disparity between sizes of ruptured IAs suggest that the assessment of additional factors in tandem with IA size may improve rupture prediction.

Aneurysm Height: The length of the centerline of the IA sac is measured, following the IA shape, as opposed to measuring a straight line from the ostium centroid directly to the highest IA point. The radius of the maximum inscribed sphere at the centerline's furthest point is added to the length measurement to fully measure the IA height. This is a modified version of the typical IA height measurement: a straight line of the maximum stretch from the ostium centroid to the IA dome [78, 158].

Vessel Diameter: The parent artery diameter value is computed at locations close to the aneurysm ostium. For terminal aneurysms, the vessel diameter of the common branch was measured at the point prior to centerline splitting between the daughter arteries, and both daughter arteries' diameter were measured at the point one (common artery) diameter away from the IA ostium cut. The average of the three values was used as the value of the vessel diameter.

Inlet Cross-sectional Area: The beginning of the inlet vessel was cut square in the 3-matic software package, the resultant cross-sectional area of the inlet vessel was calculated.

Aspect Ratio*: A modified calculation of the commonly defined aspect ratio (aneurysm hight/ostium diameter) was used by adapting the length of the center-line of an IA as the IA height (IA$_{height^*}$), and the area and circumference of the ostium since ostium diameter is rarely uniform for an IA [189].

$$AspectRatio^* = \frac{IA_{height^*}}{4 * (Ostium_{area}/Ostium_{circumfrence})} \tag{1.1}$$

The aspect ratio of an IA has been shown to be correlated with levels of hemodynamic stressors and has been used as an ease-of-use method to assess conditions within an IA [272].

Aneurysm Hemodynamic Characterisitcs

Wall Shear Stress: The calculation of wall shear stress (WSS) is performed by the ANSYS-FLUENT commercial finite-element solver (ANSYS v17.0). The value is defined as the normal velocity gradient against the (vessel) wall:

$$\tau_w = \mu \frac{\partial v}{\partial n} \tag{1.2}$$

with μ as the fluid dynamic viscosity (0.004 kg/m-s).

The spatial-temporally averaged value of the aneurysm's WSS was calculated alongside its temporally-averaged WSS minimum and temporally-averaged WSS maximum. In a similar manner as IA volume, research differs on wither high [69] or low [273] wall shear stress is a better predictive metric for IA rupture potential. In a study by Meng et. al., both high and low WSS were associated with IA rupture potential, yet causing differing cellular changes [169].

Kinetic Energy Density: The kinetic energy density (KED) within the IA dome was calculated as follows:

$$KED = \frac{\frac{1}{2}\rho \sum v^2}{n} \tag{1.3}$$

Where v is the velocity values, ρ is the mass density of blood, and n is the number of voxels within the IA. The KED at each time-step (along the cardiac phase) was calculated, as well as the Temporally averaged KED (TA-KED) for all cases.

Disturbed Flow on Vascular Endothelium

The vascular endothelial cell (EC) layer forms the innermost lining of blood vessels, directly interacting with hemodynamic stressors and helping to maintain homeostatic functions of the vasculature[51, 125]. The mechanotransduction capabilities of this initial vascular layer help maintain a selective macromoleuclar barrier, trigger vascular remodeling, regulate vascular smooth muscle cell contraction[242], and help control vascular inflammatory responses[43]. The degradation of vascular homeostatsis, resultant from disturbed hemodynamic flow patterns, has been associated with an array of vascular pathologies: aneurysms[39, 156], atherosclerosis[155], and thrombosis[53, 239]. Due to the life threating nature of IAs, improved quantitative methods to characterize hemodynamic patterns and to what degree they impart EC pathologic changes, could prove essential to further our understanding of the disease's initiation and progression.

The morphology and cytoskeletoal organization of EC have been shown to be susceptible to non-laminar flow conditions[249]. Typically, EC morphology aligns along flow directionality, forming organized parallel actin stress fibers and giving the cells an elongated structure[17, 125, 230]. Disrupted flow patterns resulting in vortex flow and altered WSS, show a differential change in EC characteristics: a rounded morphology with marginally located short actin stress fibers[53, 70, 239]. These changes have been associated with a number of structural-fucntional changes in vasuclar cells, such as increased permeability to macromolecules, increased expression of adhesion molecules (ICAM-1, VCAM-1), decreased endothelial cell regeneration and increased smooth muscle cell proliferation/migration.

Additionally, inflammatory processes within vasculature has been shown to be a significant actor in the pathogenesis of IA development and potential rupture [43, 107, 215]. In a typical physiological setting, the vascular EC layer maintains antiatherogenic characteristics, inhibiting platelet adhesion and aggregation along the vascular wall, as well as limiting cellular pro-inflammatory pathways[4]. In the occurrence of IA pathology, a breakdown of the EC inflammatory-limiting capabilities is noted: small aneurysm shown to have intimal thickening and diffuse macrophage/lymphocyte infiltration, whereas chronic atherosclerotic lesions with embedded macrophages and lymphocytes have been noted in larger aneurysms[142, 238]. Upon leukocyte and macrophage infiltration, the matrix metalloproteinase enzyme is released which digests extracellular matrix proteins leading to additional pathologic damage to the vascular wall[7, 234]. The remodeling of the vascular wall, impart due to inflammatory pathogenic activities, lead to an overall loss vessel mechanical strength and a possible ballooning out of the impacted area

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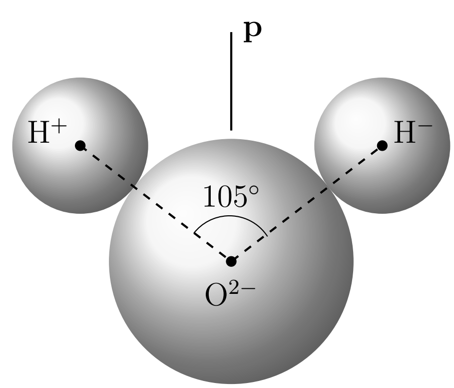

Figure 1.6: Schematic representation of a water molecule

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Chapter 2

Cerebral Vasculature and the Impact of Disturbed Flow

The human vasculature is a system aimed at carrying blood and lymph through the body. The arteries of the vascular system help deliver oxygenated blood, nutrients, and components such as inflammatory markers and hormones through the body, while the veinous structures help take cellular and tissue waste matter to organs such as the lungs, liver and kidneys for removal as well as carrying deoxygentaed blood back to the heart. The lympahtic vessels transport lymph (fluid containing water and blood cells) to help maintain hemodynamic pressures withing the body. For the purpose of studying IA, the arterial portion of the vasculature will be focused on in this work as the majority of IAs occur within the arterial system. The innermost layer of arteries, known as the tunica intima, is made of a monolayer of endothelial cells (EC) supported by a layer of collagen and elastin. This intimal layer comes into direct contact with the hemodynamic flow environment of the lumen (hollow cavity of the arterial system in which blood flows). Underlying the intima layer is the the tunica media, or media, a layer composed of smooth muscle cells, elastic connective tissue and collagen fibers. The main purpose of the media layer is to contract or dilate the arterial vasculature in response to (as signaled by the intima layer) differing hemodynamic conditions as a means to regulate circulation within the body[26, 170]. The outermost layer of arteries, the tunica externa/adventitia, is composed of collagen fibers and elastic tissue helping to maintain the mechanical properties of the vasculature, while the collagen having a secondary purpose of anchoring the vessel to surrounding tissues (improving vessel stability)(Fig. 4.1). In terms of vascular changes concerning aneurysm development and rupture, most focus is on changes to the tuinca intima's ECs, and the vascular smooth muscle cells (vSMC) of the tunica media.

Endothelial Cells

ECs are specialized cells that are directly exposed to hemodyanamic environment and its circulating components, with a basolateral surface separated from surrounding tissues by a glycoprotein membrane. Due to ECs direct contact with the hemodynamic

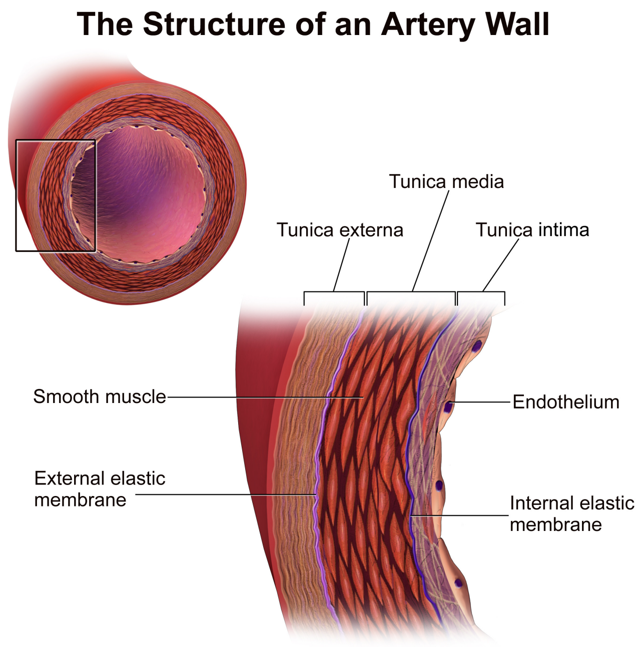

Figure 2.1: Layers of the vasculature. Original image from [218]

environment, their ability to respond to changes in flow conditions is of primary importance. Cytoskeleton actin stress fibers help resist fluid stressors, maintain cellular shape, and aid in cellular mechanotransduction. In laminar flow conditions, actin fibers maintain an oblong orientation, aligning themselves with the direction of flow and forming parallel fibers across the cell to minimize the shear stress exerted by the flowing blood. In addition, ECs act as mechanotransducers, converting mechanincal flow stressors into chemical-cellular signals triggering a variety of cellular processes within the vasculature[48]: Cell-cell junctions[75], caveolae[212], integrins[51] and the endothelial glycocalyx[35]. Platelet endothelial cell adhesion molecules (PECAM-1) is shown to create bound complexes with integrins[52, 176], vascular endothelial growth factor receptor 2(VEGFR-2) and VE-cadherin to transmit shear stressors to cellular responses[46]. Activated integrins trigger multiple complex of non-receptor tyrosine kinases (FAK, c-Src, Shc, paxillin, and p130CAS), adaptor proteins (Grb2, Crk), and guanine nucleotide exchange factors (Sos, C3G), thereby activating Ras family GTPase.2.2. EC glycocalyx, membrane-bound proteoglycans, also take part in sensing and transduction mechanical forces into biochemical signals[56, 171**?**]. Under physiologic laminar flow conditions, glycocalyx triggers an increase in eNOS expression and subsequent NO production[35, 238, 271]. NO helps regulate vascular dilation/tone and limits vascular inflammation through the inhibition of NF-B[42, 165] which in turn limits the expression of intercellular adhesion molecule-1 (ICAM-1), a membrane protein involved in vascular adhesion and migration of leukocytes.

Another role of vascular ECs is the establishment and maintenance of a selective permeability layer within the vasculature[33, 54]. The outward adventitial interstitial pressure (outside the vasculature) is low, roughly 10mmHg, with a much greater intraluminal arterial blood pressure at 130/80mmHg (in humans)[219]. This pressure gradient would create a prominent unidirectional, unchecked mass transport of molecules from the lumen of the vasculature into the surrounding media layer without an additional permeability regulator. The internal EC monolayer helps control macromolecule transport. Typically, maintenance of EC intercellular junctions helps limit macromolecules such as lipoproteins and leukocytes from passing into the intimal layer[125, 178]. The main groups of cell-cell complexes that control permeability are adherens junctions**NEED CITATION** and tight junctions[274], membrane proteins

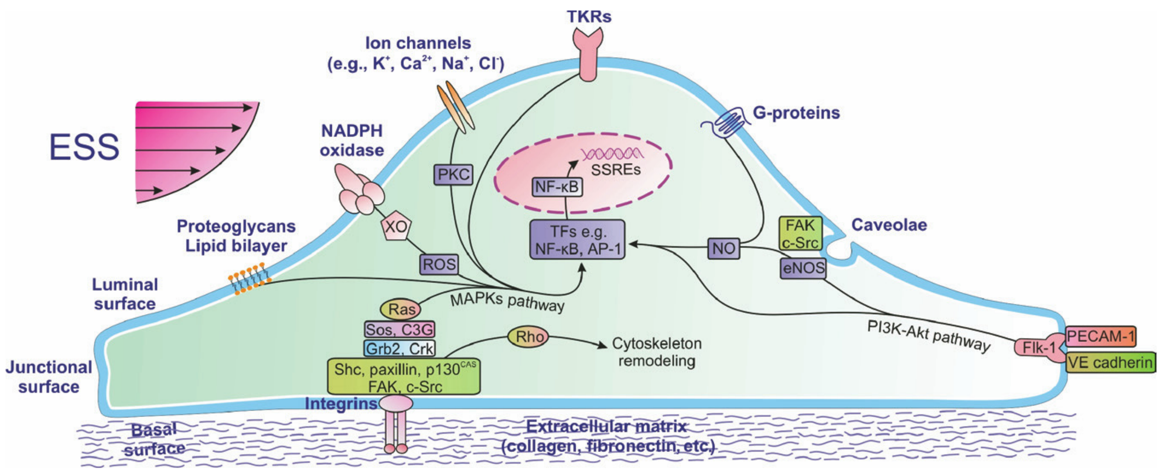

Figure 2.2: Mechanoreceptors of ECs: ion channels (K^+, Ca^2+, Na^+, Cl^-), G-proteins, caveolae, tyrosiekinase receptors (TKRs), nicotinamide adenine dinucleotide phosphate (NADPH) oxidase, xanthine oxidase (XO), inegrins, and heparan sulfateproteoglycan. Signals are transmitted through the cytoskeleton to the basal or junctional endothelial surface. Integrin mechanosensory complexes consist of platelet endothelial cell adhesion molecule-1 (PECAM-1) and Flk-1. When activated they initiate downstream signaling cascades. Activated integrins trigger multiple complex of non-receptor tyrosine kinases (FAK, c-Src, Shc, paxillin, and p130CAS), adaptor proteins (Grb2, Crk), and guanine nucleotide exchange factors (Sos, C3G), thereby activating Ras family GTPase. Active Ras plays a pivotal role in intracellular transduction of signals as it triggers various parallel downstream cascades of serine kinases; each of these activate downstream signals, ultimately activating mitogen-activated protein kinases (MAPKs). Shear stressors also activate a number of other downstream signaling pathways that impact reactive oxygen species (ROS): NADPH oxidase, activation of protein kinase C (PKC), activation of Rho family small GTPases (mediate the EC remodeling), release of endothelial nitric oxide synthase (eNOS) and activation of phosphoinositide-3 kinase (PI3K)-Akt cascade. These signaling pathways lead to phosphorylation of transcription factors (TFs) such as nuclear factor-kappa(NF-$\alpha\beta$) and activator protein-1(AP-1). Original figure from[48]

connected to the cytoskeleton through transmembrane and cytosolic proteins. The proteins of claudins and occludins make up a significant portion of the tight junction between cells[47, 274]. Tight junctions prevent the passage of molecules and ions through the space between plasma membranes of adjacent cells, so materials must pass enter the cells (by diffusion or active transport) in order to pass through the tissue helping to maintain osmotic balance within the vasculature. Adherens junctions such as vascular endothelial (VE)-cadherin also help cell-cell adhesion (and subsequent EC permeability) well as maintaining an inhibition of endothelial cell growth. A breakdown in these connections leading to *decreased* regulation of permeability within the EC monolayer as 'gaps' now exist between cells, allowing the greater degree of macromolecule infiltration to subsequent layers of the vasculature. While not directly tasked with the maintinance of EC permeability, the regulation of cellular turnover and/or apopotosis has a secondary impact on vascular permeability[195]. Areas of apoptotic cells, or areas of cellular turnover can change the permeability of the vasculature.

Activated integrins such as β-1 integrin and integrin $\alpha 5$, transmits additional cellular signaling to proteins such as Src homology domain 2-contaning kinase (Shc), which can activate NF-$\kappa\beta$ causing subsequent increases cell proliferation, apoptotic signaling through Jun-amino-terminal kinase (JNK) and altered cellular response to cytokines through p38 mitrogen-activated protein kinase (MAPK). Activated NF-$\kappa\beta$ also triggers downstream activation of a number of genes related to macrophage recruitment and vascular inflammation: MCP1, MMP2, MMP9, IL1β, and inducible nitric oxide synthase(iNOS).

The tunica media of the vasculature is primarily made up of vSMC and produce a significant portion of the vascular extracellular matrix proteins (elastin, collagen, proteoglycans, and fibrillin). This mulit-layered portion of the arterial system help regulate blood vessel integrity and vascular tone in response to hemodynamic forces and are key components in maintaining contractile functionality. Under physiologic conditions, vSMC maintain a nonproliferative phenotype with an abundance of contractile fibers, intermediate filaments, microtubules and actin. Intermediate filaments such as vimentin and desmin aid in the maintenance of vSMC structure[31], whereas microtubules **DO WHAT**. Actin filaments transmit mechanical signals dispersed within the vSMC cytoplasm and aid in the transmission of cellular signalling[24]. Changes in cellular tension alter actin cytoskeletal organization, regulating cell contraction, migration and cellular viability[33]. The myosin components within vSMC aid in cellular contraction (similar to actin fibers) and are shown to be altered in the event of vascular injury.

Impact of Disturbed Flow on Vascular Cells

Broadly speaking, the change from a healthy artery to aneurysm development and possible rupture follow an (assumed) overall path: alterations to the ECs, increased macrophage infiltration into the vessel wall, breakdown of proteins (proteolysis) in the extracellular matrix[197, 204], loss of internal elastic lamina[203, 253], and decreased collagen synthesis[135, 154]. Continued pathophysiological conditions can lead to apoptosis of vSMC causing continued weakening of the vascular wall and possible aneurysm formation. In the event of prolonged degradation to the aneurysm wall, an intraluminal thrombosis[66, 132, 150] or aneurysm rupture may occur.

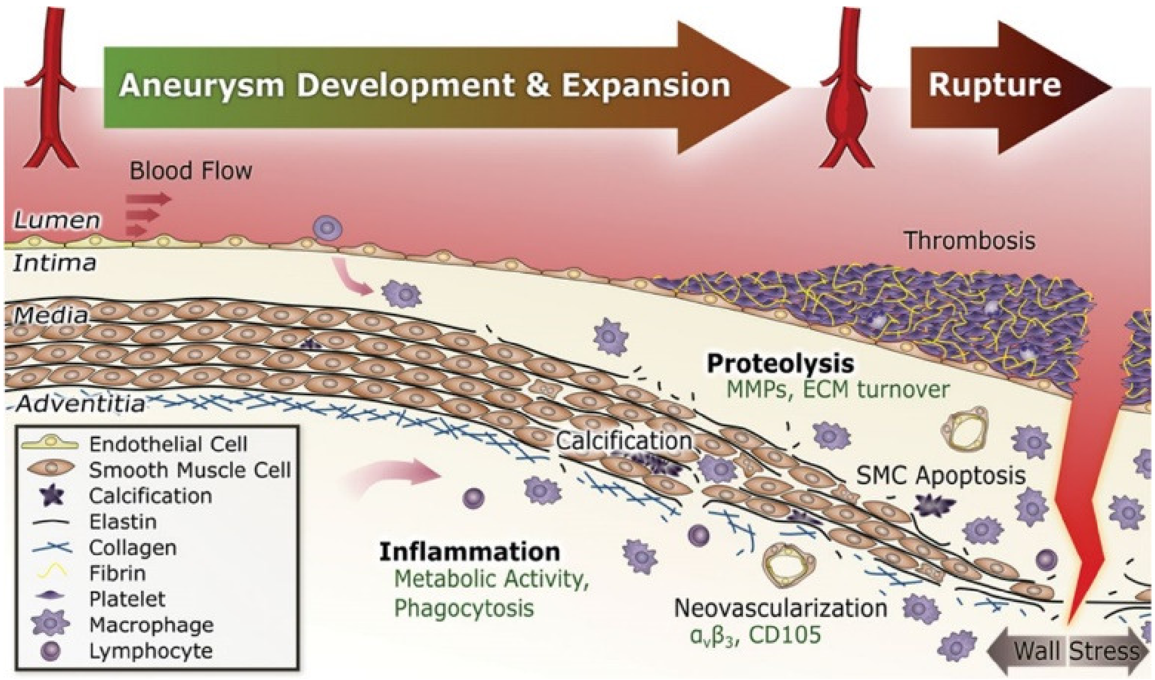

Figure 2.3: Schematic representation of abdominal aortic aneurysm (AAA) pathophysiology. Inflammation, proteolysis (breakdown of proteins by enzymes), smooth muscle cell (SMC) apoptosis, neovascularization, calcification, and intraluminal thrombosis may be targeted by molecular imaging. ECM indicates extracellular matrix; and MMP, matrix metalloproteinase.

As mentioned, laminar flow promotes the oblong orientation of actin stress fibers aligned with the direction of flow across the cell in ECs. Low WSS ($<$ 5 dynes/cm^2) disturbed flow conditions generated in in vitro, result in altered cytoskeletal orientation with actin fibers forming non-random accumulation at cell peripheries[17, 149, 261]with ECs taking on a round or polygonal shape. Additionally, changes in structural matrix smooth muscle actin expression is reduced in areas of disturbed flow, with lower actin expression in aneurysms versus healthy vessels and further reduction in actin expression within ruptured versus unruptured [**?**]. These alterations to EC morphology

Table 2.1
Disturbed Flow's Impact on Vascular Cells

Cellular Functionality Cellular component/protein	**Resultant Change due to Disturbed Flow**	***Model* Stimulus**	**Reference**
Endothelial permeability			
VEGF	Increased		[9]
	Increased	*In-vivo*: CaCl2 induction	[152]
VEGFR2	Increased	*In-vitro* Orbital Shaker	
VE-cadherin	Decreased	*In-vivo*:ligated arterial segment	[52]
	Decreased	*In-vitro*: Magnetic twisting	[235]
Atherosclerosis			
PECAM-1	Increased	*In-vivo*:ligated arterial segment	[52]
Mechanotransduction			
β1-Integrin	Increased	*In-vitro* PPFC	[262]
Integrin α5	Increased	*In-vitro*: Orbital Shaker	[227]
Glycocalyx	Decreased	*In-vitro*: PPFC *In-vivo*:Arterial Ligation	[104]
	Decreased	*In-vivo*:Arterial Ligation	[171]
	Decreased	*In-vitro*: Curved flow chamber	[56]
Inflammatory Response			
ICAM-1	Increase	*In-vitro*: PPFC with occulusion	[17]
	Increased	*In-vivo*: Arterial LIgation	[211]
	Increased	*In-vitro*: Cone-plate viscometer	[96]
	Increased	*In-vitro*: PPFC	[16]
VCAM-1	Increased	*In-vivo*: Arterial Ligation	[262]
	Increased	*In-vitro*: PPFC	[16]
ECM Remodeling			
MMP2	Increased	*In-vitro*: PPFC with occlusion	[17]
	Increased	*In-vitro*: Stenosis flow chamber	[57]
MMP-9	Increased		[56]
Oxidative Stress / Reactive Oxygen Species			
NOX	Increased	*In-vivo* genetic knockout/Ang II infusion	[216]
IL-6	Increased	*In-vivo*: Arterial Ligation	[211]
	Increased	*In-vitro*: Cone-plate viscometer	[96]
IL-1β	Increased	*In-vitro*: Cone-plate viscometer	[186]
IL-5	Increased		[18]
eNOS	Decreased	*In-vitro*: PPFC with occlusion	[104]
	Decreased	*In-vitro*: Cone-plate viscometer	[24]
iNOS	Decreased	*In-vitro*: Orbital Shaker	[161]
Apoptosis			
Caspase 12	Increase	*In-vitro*: Cone-plate viscometer	[186]
Caspase 3	Increase	*In-vitro*: Cone-plate viscometer	[109]
Transcription Factors			
NF-$\kappa\beta$	Increased	*In-vitro*: PPFC	[14]
	Increased	*In-vitro*: Cone-plate viscometer	[96]
Nrf2	Decreased	*In-vitro*: Orbital Shaker	[163]
JNK	Increased	*In-vivo*: Arterial ligation	[251]
	Increased	*In-vitro*: Oscillating stretch of cells on silicone foundation	[55]

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Table 2.2

A portrait table: first column represents the year in which the Nobel prize in physics was awarded; second column indicates the name of the scientist and the third column is the work for which the Nobel prize was awareded

Year	**Scientist(s)**	**Nobel Work**
1901	W. C. Röntgen	X-rays
1902	H. A. Lorentz	Influence of magnetism on radiation
	P. Zeeman	Influence of magnetism on radiation
1903	A. H. Becquerel	Spontaneous radioactivity
	M. Curie	Radiation phenomena discovered by Becquerel
	P. Curie	Radiation phenomena discovered by Becquerel
1904	J. W. Strutt	Argon
1905	P. E. A. von Lenard	Cathode rays
1906	J. J. Thomson	Electrical conductivity of gases
1907	A. A. Michelson	Spectroscopic and metrological investigations
1908	G. Lippmann	Photographic reproduction of colours
1909	K. F. Braun	Wireless telegraphy
	G. Marconi	Wireless telegraphy
1910	J. D. van der Waals	Equation of state of gases and liquids
1911	W. Wien	Laws governing heat radiation
1912	N. G. Dalèn	Automatic regulators for lighting coastal beacons and light buoys

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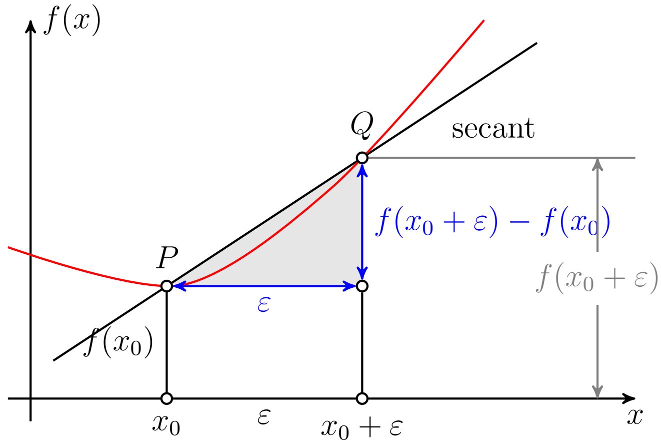

Figure 2.4: Fancy mathematical plots using TikZ package

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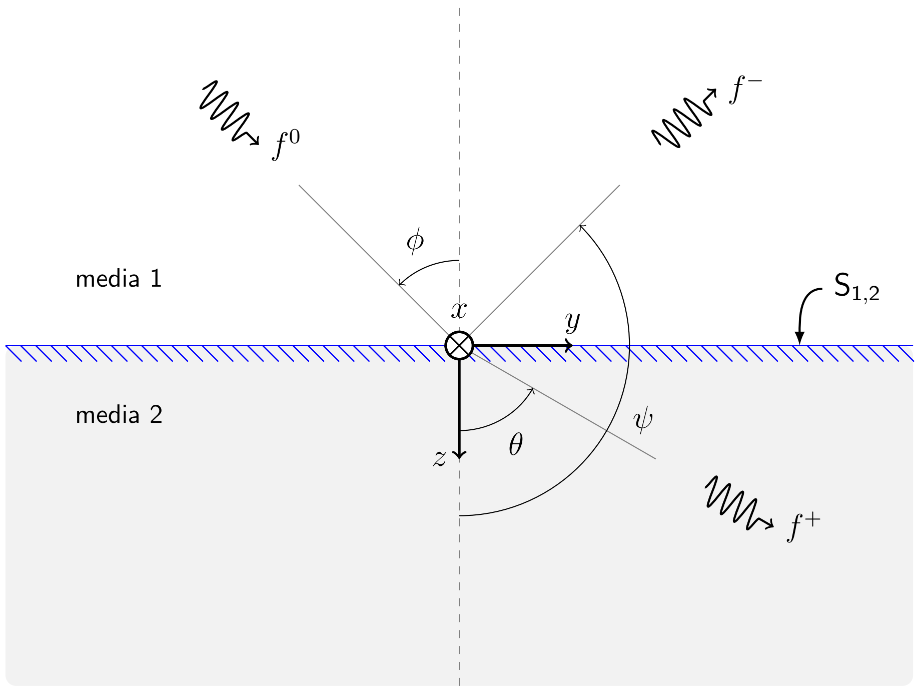

Figure 2.5: Incidence, transmission and reflection

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Chapter 3

Hemodynamic Flow Vortex Identification

The assessment of disturbed hemodynamic patterns is known to have an impact on the origin and national history of IAs [30, 255]. From a clinical perspective, phase-contrast magnetic resonance imaging (PC-MRI) or Phase-contrast magnetic resonance angiography (PC-MRA) has been used to assess flow characteristics in the vasculature *in-vivo* [22, 168]. Yet, determining flow details in and around IAs has proven difficult with PC-MRI/PC-MRA. The complex, disturbed patterns of aneurysmal flow results in incoherent velocites (at the sub-grid level) and these specific characteristics cannot be resolved by a typical "averaged" velocity measurement at the relatively large resolution in medical imaging (*at 1-mm scale*). The consequence of this sub-grid limitation, clinical hemodynamic flow measurements may be impacted by errors and potential flow artifacts which adversely affect the accuracy of PC-MRI/PC-MRA results [168, 224]. In parallel to research efforts of assessing MR flow imaging in and around IAs to determine rupture characteristics, blood flow simulated from "patient-specific" computational fluid dynamics (CFD) simulations [222] have also garnered interest by the clinical and research community [40, 263]. CFD simulation data has an initial advantage over MRI derived data in that a high degree of control of both desired data resolution, as well as data quality (lack of imaging errors and flow artifacts) can be maintained. The development of novel flow measurement techniques can be initially tested and refined on data free from errors which may confound initial findings.

Assessment of hemodynamic conditions from CFD simulations has brought about a number of potential parameters that correlate to IA rupture risk: wall shear stress [13], oscillatory shear index [223], flow impingement [40], and flow stability [32]. As mentioned in the focus of flow stability, Bryne et al. [32] found that aneurysmal flow flow spatial complexity and temporal stability is closely correlated with IA rupture. Yet this methodology only relies on qualitative assessment of the centroid-most region of vortex patterns (vortex core), giving less insight on the broader structural changes to vortices. Expanding upon the idea of vortex analysis to identify, and quantify changes to, the broader structural characteristics of vortices may give additional insight into conditions that can be linked to IA rupture potential.

In an initial study, the development of an alternative technique for vortex analysis (expanding upon vortex core analysis) was investigated to assess the presence, destruction and spatialtemporal characteristics of vortices within an IA [228]. This image processing algorithm expanded upon two established vortex identification methods, the Q-criterion [116] and λ_2 [120] methods, to identify and assess the broader aspects of vortices as opposed to solely identifying the vortex core. Subsequent studies investigated the use of a vortex identification methodology based on the Shannon's entropy (CITATION NEEDED) as an alternative identification metric not wholly reliant upon vortex core identification.

Due to the differences between traditional CFD data resolution (high) and MRI derived flow data (lower resolution), the identification methods were based on velocity data on a rectilinear grid, and the susceptibility of changes to vortex identification outcomes were tested under a range of grid resolutions. Additionally, variations to methodological outputs were tested under a range of chosen threshold values (dependent on the method and will be explained in Section NEED THE SECTION). The vulnerability of a methodology to significant changes in outcomes with minimal changes to methodological threshold values would have the potential to result in broad variations to research findings if thresholds were to be applied to a wide array of studies. Toward this end, the primary focus of this work was to explore analyzing the spatiotemporal characteristics of hemodynamic vortex structures as a possible means to compliment future assessment of IA rupture potential.

Materials and Methods

Modeling of "Patient-specific" Vasculature

Medical imaging scans of ten (10) patient's vasculature structure were arbitrarily selected from an internal database: five patients with a single bifurcation aneurysm, and five with a single sidewall aneurysm. Models were located within the internal carotid artery or the basilar artery. A commercially available image segmentation package (Mimics Innovation Suite, version 17, Materialise Inc. Leuven, Belgium) was used to reconstruct the vascular surface from digital subtraction angiography (DSA) scans, resulting in 'patient-specific' vascular structures. For all cases, the longest available upstream vessel section proximal to the aneurysm was left intact to maintain as much of the patient vessel geometry as possible. Surface irregularities were manually removed using the localized smoothing capability in the commercially available computer aided design (CAD) 3-matic software (Version 9, Materialize Inc., Leuven, Belgium). Additionally, a 1^{st} order Laplacian smoothing filter was used to perform a global smoothing to the vessel structure, reducing surface irregularities while preserving the vascular geometry. Cylindrical flow extensions (6 times the inlet vessel diameter) were added to each model using the open-source Vascular Modeling Toolkit (VMTK) software (version 1.2). The addition of vessel extensions help reduce the effects of inlet, plug-flow flow on hemodynamic characteristics [191].

Mesh Generation

Processed vascular surface structures were converted into an unstructured, 3D, tetrahedralized volumetric mesh using an open-source mesh generator, Tetgen (version 1.4.2) [214]. The mesh generation process was done by an in-house Python script derived from the VMTK program. Approximately, 1 million computing cells were used per case, with the average mesh size as 0.0022-mm^3.

CFD Simulation

To compute fluid velocity data in and around the IA, the time-dependent incompressible, 3D Navier-Stokes equations was (THERE NEEDS TO BE AN ADJECTIVE HERE....CHECK RECENT PAPERS) solved using two CFD solvers: a commercial CFD solved (version 14.0, ANSYS-FLUENT Inc., Lebanon, NH) and a research prototype CFD solver (version 4.0, Siemens Medical Solution Inc., IL). Details on the Navier-Stokes equation can be found in the Appendix. In the ANSYS-FLUENT solver, the pressure-velocity coupling was obtained using the SIMPLEC algorithm [246]. The explicit time-marching second-order scheme with a time step of $1 \times 10^{-}3$

second (1000 steps per cardiac cycle) was used for computations.

As the Siemens research CFD solver is still under development, limited information on its hosen methods for solving the Navier-Stokes equation will be discussed in this thesis. IA models were defined by water-tight 3D surface triangles were automatically discretized with cubical voxels. A Lattice-Boltzmann Mehod (LBM) solver was then used to chose adaptievly choose the solver time-step, and varied from $1 \times 10^{-}3$ to $2 \times 10^{-}3$. A Siemens Leonardo workstation equipped with a dual quad-core CPU and 8 GB of memory was used to perform CFD simulations. Of note, the exact same vessel geometries (STL files) were used to generate the volumetric meshes (for the ANSYS-FLUENT solver) and voxel discretization (for the LBM solver). The final velocity results obtained from the LBM method were re-sampled to form velocity data onto a rectilinear grid whose voxel size varied from 0.18 to 0.25 mm.

In both solvers, vessels walls were assumed rigid with a no-slip boundary. While blood vessels are not rigid, the assumed rigidity for simulation has been shown to cause a rise in overall wall shear stress values, but have minimal impact on flow pattern characteristics in the arterial system unless the microvasculature is being simulated. Blood was considered an incompressible and Newtonian fluid with a dynamic viscosity of 0.004 kg/m-s and a mass density of 1050 kg/m^3. A zero-pressure condition was used for all vessel outlets. For inlet flow rates, two pulsatile waveforms at a rate of 60 bpm were derived from magnetic resonance measurements and were taken from Gwilliam et al. [99] as patient-specific flow waveforms were not available. Each case had its inlet waveform scaled according to their inlet cross-sectional area, standardizing their mean volumetric flow rate to either 280mL/min for ICA cases or 180mL/min for BAs. This choice of volumetric flow rate(s) were based on measured physiological flow rates available in MR literature [82, 275]. Four (4) cardiac cycles were simulated per case at 20 data points per cardiac cycle with only the final cardiac cycle saves as a means to reduce initial transient flow conditions.

Aneurysm Extraction and Voxelization of Aneurismal Velocity Data

A published method [121] was used to semi-automatically isolate and extract the IA sac. The isolated IA sac was sealed at the IA opening (ostium) and converted to a binary mask that is spatially-registered with the volumetric velocity data. The mask allows the analysis of only the intra-aneurysmal velocity data. TO verify intra-rater reliability of proper sectioning of IA masks, 2 separate users sectioned the IAS and Bland-Altman plots were performed on the resultant mask volumes and ostium areas to determine the similarity between chosen masks. Once no significant differences were ensured between sectioned masks, one user was chosen at random and all resultant masks from that user were implemented in the rest of the study.

Vortex Core Extraction and Analysis

All computational methods for identification and extraction of vortices and spatio-temporal analysis of said vortices were performed using in-house scipts (C++ and Python) that were derived form the open-source VTK/VMTK software package.

This flow assessment was performed through the analysis of vortex critical point (core) lines and it was concluded that ”ruptured aneurysms had more complex and more unstable flow patterns than un-ruptured aneurysms.” In their work, Bryne used proper orthogonal decomposition [] of time-resolved velocities were used to characterize temporal flow stability. As a brief explanation, vortex core lines, identifying the center-most region of vortex pattern, were identified by the use of a co-linearlity conditions between flow instantaneous vorticity $\vec{\omega}$ and velocity $\vec{v}$ vectors.

$$\begin{aligned} \vec{\omega} \times \vec{v} &= 0 \\ \vec{\omega} &= \nabla \times \vec{v} \end{aligned} \tag{3.1}$$

To identify the centroid region of vortices, the eigenvalues of the velocity gradient tensor was calculated. In the event of a pair of complex conjugate eigenvalues was identified, the vorticity vector $\vec{\omega}$ as calculated and tested against the velocity vector to assess whether Equation 3.1 was satisfied. From an identified element, the velocity component in the direction of vorticity vector was subtracted form the velocity vector (reduced velocities). Element faces that had a point where the reduced velocity is zero was marked, and if two or more faces of an element had a zero reduced velocity, a vortex core line passes through the element.

Vortex core line analysis

$$\begin{aligned}\nabla\vec{u} &= S + \Omega \\ S &= \frac{1}{2}\left[(\nabla\vec{u}) + (\nabla\vec{u})^T\right] \\ \Omega &= \left[\frac{1}{2}(\nabla\vec{u}) - (\nabla\vec{u}^T)\right]\end{aligned} \tag{3.2}$$

Where $\nabla\vec{u}$ is the calculation of the velocity gradient: S as the rate-of-strain tensor and Ω as the vorticity tensor.

Hunt, Wray and Moin [116] defined a vortex as the spatial region of flow where the Euclidean norm of the vorticity tensor dominates.

$$Q = \frac{1}{2}\left[|\Omega|^2 - |S|^2\right] > 0 \tag{3.3}$$

Jeong and Hussain identified the vortices as:

$$\lambda_2 = (S^2 + \Omega^2) < 0 \tag{3.4}$$

where $\lambda_2 A$ identifies a vortex when the second intermediate eigenvalue of the 3 x 3 tensor A is symmetric (all three eigenvalues are real).

In our original study, the normalized Q and λ_2 values were tested to identify vortices within IAs.

$$\begin{aligned} Q(x,t) &= \frac{Q(x,t)}{|\vec{u}(x,t)|^2} \\ \lambda_2(x,t) &= \frac{\lambda_2(x,t)}{|\vec{u}(x,t)|^2} \end{aligned} \tag{3.5}$$

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$$d\nu_\theta \;=\; \frac{N}{V}\,\left(\frac{m}{2\pi\,kT}\right)^{3/2}\,\left[\int_0^{2\pi}\,\int_0^{\infty}\,v^3\,e^{-mv^2/2kT}\,dv\,d\phi\right]\,\sin\theta\,\,\cos\theta\,\,d\theta$$

$$=\;2\,\pi\,\frac{N}{V}\,\left(\frac{m}{2\pi\,kT}\right)^{3/2}\,\left[\int_0^{\infty}\,v^3\,e^{-mv^2/2kT}\,dv\,\right]\,\sin\theta\,\,\cos\theta\,\,d\theta$$

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$$d\nu_\theta \;=\; \frac{N}{V}\,\left(\frac{2\,k\,T}{m\,\pi}\right)^{1/2}\,\sin\theta\,\,\cos\theta\,\,d\theta \tag{3.6}$$

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$$
\begin{aligned}
d\nu_v &= \frac{N}{V} \left(\frac{m}{2\pi\, kT}\right)^{3/2} \left[\int_0^{2\pi} \int_0^{\pi/2} \sin\theta \ \cos\theta \ d\theta \ d\phi\right] v^3 \ e^{-mv^2/2kT} \ dv \\
&= 2\,\pi \ \frac{N}{V} \left(\frac{m}{2\pi\, kT}\right)^{3/2} \left[\int_0^{\pi/2} \sin\theta \ \cos\theta \ d\theta\right] v^3 \ e^{-mv^2/2kT} \ dv
\end{aligned}
$$

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$$
d\nu_v = \frac{N}{V} \, \pi \left(\frac{m}{2\pi\, kT}\right)^{3/2} v^3 \ e^{-mv^2/2kT} \ dv \tag{3.7}
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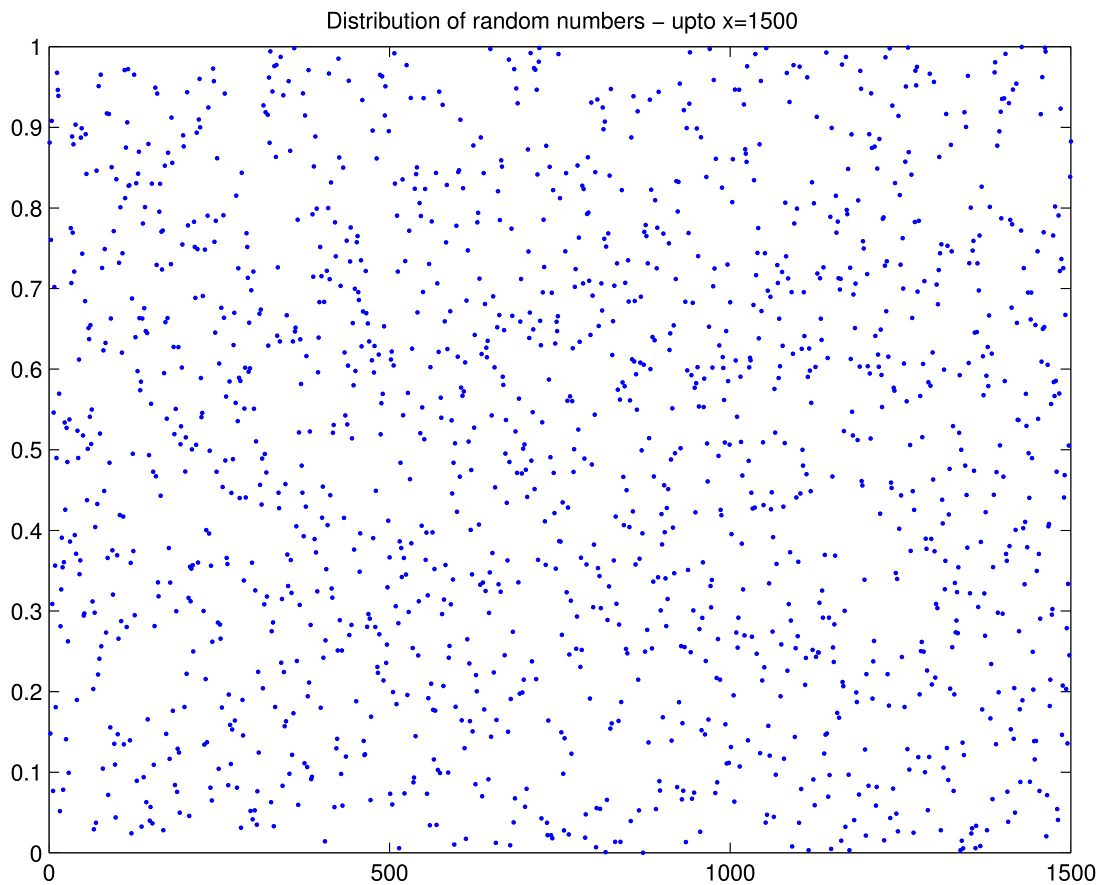

Figure 3.1: Distribution of random numbers

Table 3.1
Measured data points representing the relationship between x and y

x	0	1	2	3	4	5	6	7	8	9	10
y	0	0.94	0.99	-0.52	-1.82	-0.44	3.54	6.69	5.38	0.00	-4.42

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Table 3.2

A landscape table: first column represents the year in which the Nobel prize in physics was awarded; second column indicates the name of the scientist and the third column is an *as is* Nobel citation

Year	Scientist(s)	Nobel Work
1901	W. C. Röntgen	in recognition of the extraordinary services he has rendered by the discovery of the remarkable rays subsequently named after him
1902	H. A. Lorentz and P. Zeeman	in recognition of the extraordinary service they rendered by their researches into the influence of magnetism upon radiation phenomena
1903	A. H. Becquerel	in recognition of the extraordinary services he has rendered by his discovery of spontaneous radioactivity
	M. Curie and P. Curie	in recognition of the extraordinary services they have rendered by their joint researches on the radiation phenomena discovered by Prof. Henri Becquerel
1904	J. W. Strutt	for his investigations of the densities of the most important gases and for his discover argon in connection with these studies
1905	P. E. A. von Lenard	Cathode rays
1906	J. J. Thomson	Electrical conductivity of gases
1907	A. A. Michelson	Spectroscopic and metrological investigations
1908	G. Lippmann	Photographic reproduction of colours
1909	K. F. Braun and G. Marconi	Wireless telegraphy
1910	J. D. van der Waals	Equation of state of gases and liquids
1911	W. Wien	Laws governing heat radiation
1912	N. G. Dalèn	Automatic regulators for lighting coastal beacons and light buoys

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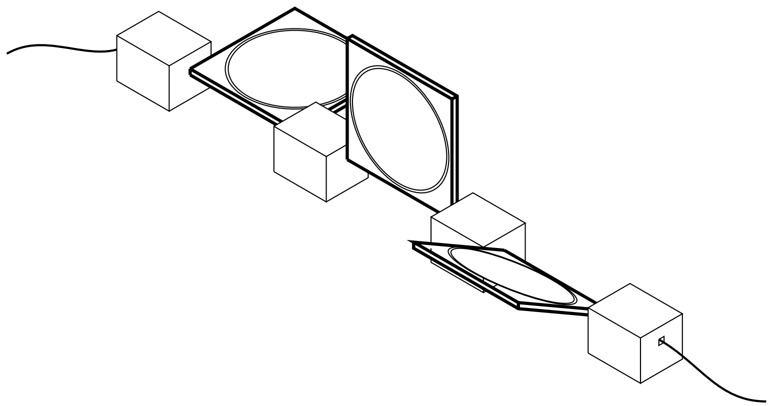

Figure 3.2: Fibre optics

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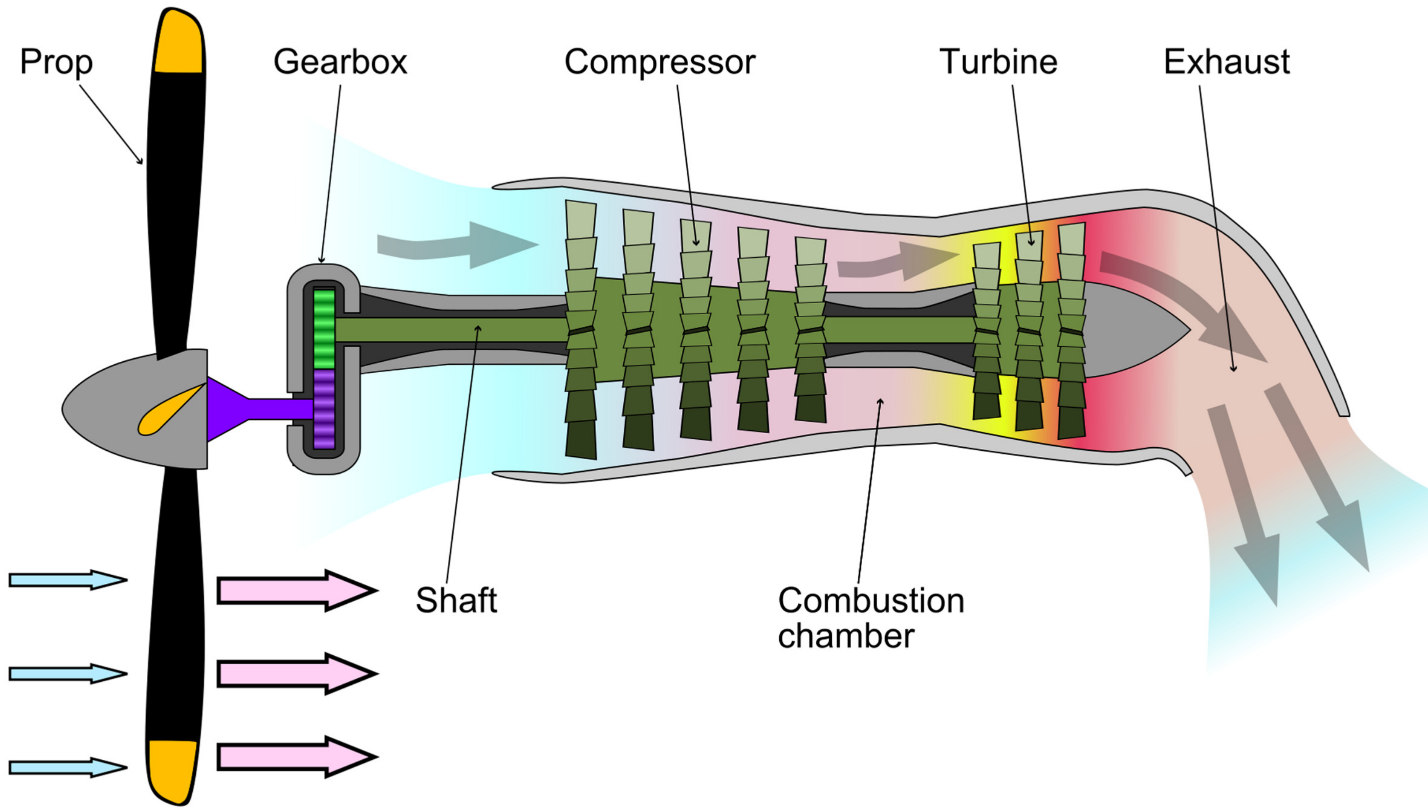

Figure 3.3: A landscape view of a Turboprop engine - these are jet engine derivatives, still gas turbines, that extract work from the hot-exhaust jet to turn a rotating shaft, which is then used to produce thrust by some other means

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Chapter 4

Vortex Analysis to predict IA Initiation

The tangential, frictional stress caused by blood flowing along the vessel wall is known as WSS. The ANSYS-FLUENT software calculates WSS by the normal velocity gradient at the vessel wall:

$$\tau_{\mathrm{w}} = \mu \frac{\partial v}{\partial n} \tag{4.1}$$

where μ is the dynamic viscosity. In this work, areas of high WSS were of interest as it is thought to play a role in the IA initiation [169]. High WSS was defined as values ≥ 20 Pa during peak systole of the MRI waveform.

The WSSG was calculated using in-house VMTK scripts and is derived from three spatial derivatives of the WSS as follows:

$$WSSG = \sqrt{(\frac{\partial \tau_w}{\partial x})^2 + (\frac{\partial \tau_w}{\partial y})^2 + (\frac{\partial \tau_w}{\partial z})^2} \tag{4.2}$$

with the time-averaged WSSG calculated as

$$WSSG_{\text{av}} = \frac{1}{T}\int_0^T |WSSG|dt \tag{4.3}$$

OSI is a nondimentional parameter, computing oscillations in the direction of the WSS vectors over the course of a cardiac cycle:

$$OSI = \frac{1}{2}\left\{1 - \frac{|\int_0^T \tau_i dt|}{\int_0^T |\tau_i| dt}\right\} \tag{4.4}$$

were τ_{i} represents the WSS vector at a given time step across the duration of the cardiac cycle (T). The OSI describes the changes of a WSS vector's alignment with the cardiac cycle's temporally-averaged WSS vector. An OSI of 0 indicates no change in directionality and 0.5 being a complete direction reversal.

The AFI [**?**] quantifies the variation in angle between the instantaneous WSS vector and time-averaged WSS vector:

$$AFI = cos(\theta) = \frac{\tau_i \cdot \tau_{av}}{|\tau_i| * |\tau_{av}|} \tag{4.5}$$

For each point along the vessel wall, the minimum AFI calculated during the cardiac cycle was used to indicate the greatest deviation of the WSS vector from its mean direction. A minimum AFI of -1, 0, and 1 indicate deviations of 180°, 90°, and 0°respectively.

The GON index [?] quantifies fluctuations in WSSG directionality over the cardiac cycle.

$$GON = 1 - \frac{|\int_0^T G dt|}{\int_0^T |G| dt} \tag{4.6}$$

T is the period of the cardiac cycle and G is the spatial wall shear stress gradient vector

The human vasculature is a system aimed at carrying blood and lymph through the body. The arteries of the vascular system help deliver oxygenated blood, nutrients, and components such as inflammatory markers and hormones through the body, while the veinous structures help take cellular and tissue waste matter to organs such as the lungs, liver and kidneys for removal as well as carrying deoxygentaed blood back to the heart. The lympahtic vessels carry lymph (fluid containing water and blood cells) to help maintain hemodynamic pressures withing the body. Intracranial aneurysms have only been shown to develop within the arterial portion of the vasculature, and as such will be the the part of the vascualture fouced upon in this work. The innermost layer of arteries, known as the tunica intima, is made of a monolayer of endothelial cells (EC) supported by a layer of collagen and elastin. This intimal layer comes into direct contact with the hemodynamic flow environment of the lumen (hollow cavity of the arterial system in which blood flows). This endothelial layer serves multiple purposes: hemodynamic mechanosensors, reacting to fluid forces and transducing said force into biochemical signals triggering cellular cascades controlling vascular tone and homeostasis[35, 51, 75], and acting as a selective permeability layer for macromolecules[21, 95, 177]. Underlying the intima layer is the the tunica media, or media, a layer composed of smooth muscle cells, elastic connective tissue and collagen fibers. The main purpose of the media layer is to contract or dilate the arterial vasculature in response to (as signaled by the intima layer) differing hemodynamic conditions as a means to regulate circulation within the body[26, 170]. The outermost layer of arteries, the tunica externa/adventitia, is composed of collagen fibers and elastic tissue helping to maintain the mechanical properties of the vasculature, while the collagen having a secondary purpose of anchoring the vessel to surrounding tissues (improving vessel stability)(Fig. 4.1).

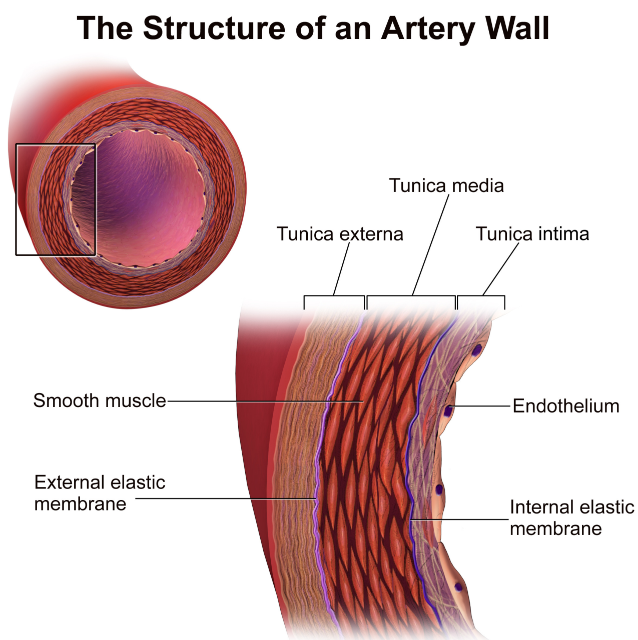

Figure 4.1: Layers of the vasculature. Original image from [218]

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Appendix A

Statistics

In this type of predictive modeling, there exists an input-output dataset (X,Y) $\in XxY$ w with an unknown probability distribution P. The goal of predictive modeling is to find a function $f_n : X \to Y$, that is determined using a training set (X_1, Y_1,...,(X_n, Y_n) of n random pairs distributed as (X,Y). A desirable solution of f_{n} is one that, given a new data-point $x \in X$, the resultant f_{n}(x) is an accurate prediction of the true output $y \in Y$. This desired outcomes not only relies on the chosen function's predictive accuracy, but also of the selecting of relevant variables that are capable of achieving desired predictions. For desired models, it is often preferred to find the prediction function that achieves the desired accuracy while using the minimal amount of variables required: i.e a *parsimonious* model. Brute-force methods of testing all variable combinations becomes increasingly unviable, especially when the number of variables in a dataset is larger than the number of n data points (cases) available for analysis: often refereed to the "large p, small n paradigm". One type of methodology to determine a desired model is through the use of sparsity-based regularization methods [118, 232, 233, 278]

Section 1

Multiple logistic regression (MLR) analysis looks both to estimate the odds of a dichotomous outcome occurring, and to determine the impact of an individual variable (covariate) in relation to the other covariates in a model. The probability of an outcome occurring in MLR can be calculated as such:

$$\hat{p} = \frac{exp(b_0 + b_1X_1 + b_2X_2 + ... + b_pX_p)}{1 + exp(b_0 + b_1X_1 + b_2X_2 + ... + b_pX_p)} \tag{A.1}$$

$\hat{\mathrm{p}}$ being the probability of the desired outcome, $\mathrm{X_1}$ through $\mathrm{X_p}$ as the individual dependent variables applied to the model, and $\mathrm{b_1}$ to $\mathrm{b_p}$ being each variable's (respective) regression coefficients. To determine the expected log odds ratios of the model's variables, the *logit* function of the above equation can be calculated:

$$
\begin{aligned}
logit[\hat{p}] &= ln[\frac{\hat{p}}{1-\hat{p}}] \\
&= ln\left[\frac{\frac{exp(b_0+b_1X_1+b_2X_2+...+b_pX_p)}{1+exp(b_0+b_1X_1+b_2X_2+...+b_pX_p)}}{1-\frac{exp(b_0+b_1X_1+b_2X_2+...+b_pX_p)}{1+exp(b_0+b_1X_1+b_2X_2+...+b_pX_p)}}\right] \\
&= ln\left[\frac{\frac{exp(b_0+b_1X_1+b_2X_2+...+b_pX_p}{1+exp(b_0+b_1X_1+b_2X_2+...+b_pX_p}}{\frac{1}{1+exp(b_0+b_1X_1+b_2X_2+...+b_pX_p}}\right] \\
&= ln[exp(b_0+b_1X_1+b_2X_2+...+b_pX_p] \\
&= b_0+b_1X_1+b_2X_2+...+b_pX_p
\end{aligned}
\tag{A.2}
$$

Taking the *logit* of the desired outcome's probability, transforms the occurrence of the event given Xs into a simplified linear function.

For each variable added to a regression model, the resultant R^2 (coefficient of multiple determination) may increase, indicating an improved fit of the data. However applying a large number of variables to a predictive model may result in over-fitting without a significantly large dataset: large p, small n paradigm. In such an event, the R^2 values, regression coefficients, and any statistical significance (p-values) determined may be misleading. To reduce the initial choices of variables in assessed predictive models, the correlation between variables were determined. The correlation of data can be determine by:

$$
r_{jk} = \frac{s_{jk}}{s_j s_k} = \frac{\sum_{i=1}^{n}(x_{ij}-\overline{x}_j)(x_{ik}-\overline{x}_k)}{\sqrt{\sum_{i=1}^{n}(x_{ij}-\overline{x}_j)^2}\sqrt{\sum_{i=1}^{n}(x_{ik}-\overline{x}_k)^2}}
\tag{A.3}
$$

with r as the Pearson correlation coefficient between variables x_j and x_k, n as the sample size, and $\overline{x}$ is a variable sample mean. Correlations between the variables are often displayed via a correlation table:

$$R = \begin{bmatrix} 1 & r_{12} & r_{13} & \dots & r_{1p} \\ r_{21} & 1 & r_{23} & \dots & r_{2p} \\ r_31 & r_{32} & 1 & \dots & r_{3p} \\ \vdots & \vdots & \vdots & \ddots & \dots \\ r_{p1} & r_{p2} & r_{p3} & \dots & 1 \end{bmatrix}$$

Initial correlation analysis of all available geometric and hemodynamic variables was performed to eliminate highly correlated variables from analysis: i.e aneurysm volume and surface area are highly correlated so surface area was removed from analysis.

From the remaining variables, stepwise MLR was implemented to determine the parsimonious model. In stepwise regression, a linear regression is first performed for each variable X one at a time, and the variable with the highest R^2 is kept for the model. Next, a multiple regression step is performed with the kept variable and each remaining variable. The variable with the largest increase in R_2, if the p value of the R^2 is below a desired cuttoff (<0.05), is added to the model. The calculation of the p value of an increase in R^2 resulting from the increasing of X variable(s) from a to

b is as follows:

$$p_{ab} = \frac{(R_b^2 - R_a^2)/(b-a)}{(1-R_b^2)/(n-b-1)} \tag{A.4}$$

with the total sample size n.

Each time a new variable is added to the model, the impact of removing any of the other variables (already added to the model) on outcomes is tested. The chosen (removed) variable is excluded from the model if it does not make R^2 significantly worse. This process is continued till adding any new variables does not increase R^2 and removing any X variables does not significantly decrease R^2

In the event that all of the independent variables in the model are completely uncorrelated with each other, the interpretation of coefficients are as such:

$$OR = exp(b_1)^z \tag{A.5}$$

Where z is the number of unit changes for a variable X, and OR is the odds ratio resultant from said change. When the variables are not uncorrelated, the $OR = exp^z b_1$ is expressed as the change of unit z for a variable *adjusted in relation to the impacts of the other variables in the model.* This stresses the need to assess collinearity between variables prior to model assessment.

in Section **??**.

Section 2

Limitations may arise in applying multiple logistic regression analysis to data sets with a large number of variables in relation to the number of samples.

The Nearest Shrunken Centroid (NSC) method is a statistical classification methodology developed by Tibshirani et al.[233] as a means to improve predictions in high-dimensional data. Additionally, a 2014 study by Finch [86] compared a number of methods for statistical group prediction. The NSC method was found to be robust in terms of accuracy and identification of predictor variables over other methods when dealing with high-dimensional datasets. Initially developed for predictions within genetic data, NSC aims to shrink class (e.g. rupture status) centroids towards the *overall* centroid of they entire data-set after standardizing by the within-class deviation for each variable. This standardization of the resultant centroid gives higher impact to variables whose expression is more stable withing samples of the same class. Due to the number of geometric and hemodynamic variables that may impact IA rupture, the NSC method was investigated as a useful methodology to help predict rupture potential within our data-set.

In our study, x_{ijk} represents the value of quantified variables $i = 1, ..., p$ in aneurysm $j = 1, ..., n_k$ of class k. The mean for variable i in each class k is calculated

$$\overline{x}_{ik} = \sum_{j=1}^{n_k} \frac{x_{ijk}}{n_k} \tag{A.6}$$

and the *overall* mean for variable i is calculated.

$$\overline{x}_i = \sum_{k=1}^{K} \sum_{j=1}^{n_k} \frac{x_{ijk}}{n_k} \tag{A.7}$$

A t-statistic value, d_{ik}, for each variable is calculated, comparing class k to the overall mean:

$$d_{ik} = \frac{\overline{x}_{ik} - \overline{x}_i}{m_k \cdot (s_i + s_0)} \tag{A.8}$$

with

$$s_i^2 = \frac{1}{n-K} \sum_{k=1}^{K} \sum_{j=1}^{n_k} (x_{ij} - \overline{x}_{ik})^2 \tag{A.9}$$

and

$$m_k = \sqrt{1/n_k + 1/n}, \text{ where } n = \sum_{k=1}^{K} n_k \tag{A.10}$$

The s_0 in (A.8) is used as a *regularization parameter* to help protect form large d_{ik} values occurring from variables at low expression levels. The value of s_0 is determined as the median value of s_i over the set of variables.

With the inclusion of d_{ik}, the class centroid can be rewritten as:

$$\overline{x}_{ik} = \overline{x}_i + m_k(s_i + S_0)d_{ik} \tag{A.11}$$

The NSC method shrinks each d_{ik} toward zero, creating a new value $'_{ik}$ which generates a new shrunken centroid value

$$\overline{x}'_{ik} = \overline{x}_i + m_k(s_i + s_0)d'_{ik} \tag{A.12}$$

The value of shrinkage is determined through *soft thresholding*, where the absolute value of d_{ik} is reduced by Δ and is given the value of 0 if the result is < 0, with

$$d'_{ik} = sign(d_{ik})(|d_{ik}| - \Delta)_+ \tag{A.13}$$

This combination of shrinkage and *soft thresholding* can result in many 'noisy' $\overline{x}_{ik}$ being close to the value of the overall mean $\overline{x}_i$. If Δ shrinks d_{ik} to zero for all k, then the centroid for variable i is the same for all classes and is excluded from prediction analysis.

The value of Δ is determined through cross-validation: fitting the model for many values of Δ and determining the level of error per chosen Δ. The Δ resulting in the smallest error was chosen for our prediction.

Section 3

Elastic Net Regularization (ENR) overcomes some of the limitations of the LASSO selection method, primarily being able to accurately handle data sets with a high number of variables in relation to the sample size [80, 232]. Additionally, the ENR method is able to handle data sets with groups of highly correlated variables.

ENR solves two optimization problems:

$$\begin{aligned} \tilde{\beta} &= \text{arg } \min_{\beta} \sum_{i=1}^{N} (y_i - (X\beta)_i)^2 \\ \text{subject to} &\sum_{j=1}^{p} |\beta_j| \leq t_1 \text{ and } \sum_{j=1}^{p} \beta_j^2 \leq t_2 \end{aligned} \tag{A.14}$$

were a penalty is places on the L_i norm($\sum_{j=1}^{p} |\beta_j|$) and the L_2 norm ($\sum_{j=1}^{p} \beta_j^2 \leq t_2$ of the regression coefficients. The purpose of these penalties are as follows: L_1 performs variable selection by setting some coefficients to 0, and L_2 works toward group selection by shrinking the coefficients of correlated variables toward each other. Re-writing equation A.14 in the Lagrangian form using two tuning parameters ($lambda_1$ and λ_2)

is as follows:

$$\tilde{\beta} = \arg\ \min_{\beta} \left(\sum_{i=1}^{N} (y_i - (X\beta)_i)^2 + \lambda_1 \sum_{j=1}^{p} |\beta_j| + \lambda_2 \sum_{j=1}^{p} \beta_j^2 \right) \tag{A.15}$$

The choice of tuning parameter values is performed by analyzing an array of λ_2 values (0, 0.01, 0.1, 1, 10, and 100). For each value in the array, the LARS-EN algorithm calculates the resultant λ_1 value. The $lambda_1$ value that yields the smallest k-fold cross validation error, and its $lambda_2$ value used to generate it, are used as the tunning parameters for the ENR method.

Section 4

Section 5

To assess the diagnostic ability of predictive model(s), a receiver operating characteristic curve (ROC) is often deployed (REFERENCES). The ROC curve assesses a model's predictive true positive rate (TPR) against its false positive rate (FPR) as a means to determine overall predictive strength (HANLEY). From a statistical perspective, ROC analysis can be considered as a plot of the power (probability of a test correctly rejecting the null hypothesis when an alternative hypothesis is true)

$$
\begin{aligned}
TPR &= \frac{\Sigma TruePositive}{\Sigma ConditionPositive} \\
FPR &= \frac{\Sigma FalsePpositive}{\Sigma ConditionNegative} \\
FNR &= \frac{\Sigma FalseNegative}{\Sigma ConditionPositive} \\
Specificity &= \frac{\Sigma TrueNegative}{\Sigma ConditionNegative}
\end{aligned} \tag{A.16}
$$

When dealing with a binary classification, as per this study, the predictive test measure for each instance is denoted by a continuous random variable (x). Given a desired threshold (T), each instance is positive if x>T and negative if x<T. Setting the probability distribution functions of the positive and negative values of x to $f_p(x)$ and $f_n(x)$ respectively, the . Given this, TPR is calculated as:

$$TPR(T) = \int_T^\infty f_p(x)dx \tag{A.17}$$

and the FNR as:

$$FPR(T) = 1 - \int_T^\infty f_n(x)dx \tag{A.18}$$

The ROC curve is generated by plotting TPR(T) against FPR(T) parametrically, varying across T, or as a plot of:

$$ROC(T) = 1 - f_p(f_n^{-1}(1-T)) \tag{A.19}$$

over T from [0,1] where $f_p{}^{-1}$(1-T) = inf

Comparing the resultant ROC curves across multiple models provides the selection of the desired model based off of varying predictive accuracies. To quantify the predictive accuracy, the area under the curve (AUC) of the ROC curve is calculated, as it equals the probability of a classifier ranking a positive instance higher than a negative instance (both chosen at random).

$$\begin{aligned} A &= \int_{\infty}^{-\infty} TPR(T)FPR'(T)dT \\ &= \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} I(T' > T)f_1(T')f_0(T)dT'dT = P(X_1 > X_0) \end{aligned} \tag{A.20}$$

The initial integral has reversed boundaries due to larger T values having a lower value on the x-axis.

Section 5

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Appendix B

Sample Code

The method for vortex identification for this study is a modification from previous work[228]. The calculation of vortex cores was based on in-house C++/Python codes derived from the open-source Vascular Modelling ToolKit (VMTK) [6]. Prior to any calculations, velocity data is first re-sampled onto a rectilinear grid whose voxel size is 0.2mm.

In the first step, the classic λ_2 method by Jeong and Hussain [120] was used to define the negative λ_2 region (*i.e* $\lambda_2 < 0$). Then, in the second step, vortex core lines were estimated by the method proposed by Sujudi and Haimes [225]. In essence, in the negative λ_2 region, a local velocity vector $\overline{v}$ lies along a vortex core line if the following two conditions hold: (1) the 3×3 spatial gradient matrix of $\overline{v}$ has two complex eigenvalues and one real eigenvalue and (2) the 3×3 spatial gradient matrix of $\overline{v}$ has

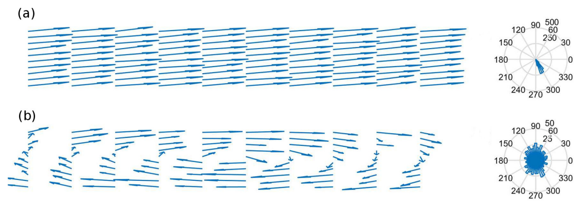

Figure B.1: Two examples illustrating the relationship between the angular histogram and NE: (a) a simple laminar flow case and (b) a rotational flow (eddy) case. In both cases, the right and left plots are the vector flow field and the histogram of angular vector direction, respectively. Vector fields were decimated by a factor of 3 for better visualization.

an eigenvector $\vec{\alpha}$ corresponding to the above-mentioned real eigenvalue. Now, if we define a new scalar value K as follows,

$$K(x,y,z) = \begin{cases} |dot(\overline{v},\vec{\alpha})|, & if\ \lambda_2 < 0 \\ 0, & Otherwise \end{cases} \tag{B.1}$$

where $|\cdot|$ is an absolute operator. Of note, in Eqn. 2, both the $\overline{v}$ and $\vec{\alpha}$ are normalized and therefore, the scalar field K defined above is bounded between 0 and 1. If the *K(x,y,z)* is close to 1 then the location *(x,y,z)* is within the proximity of the vortex core line as suggested by Sujudi and Haimes [225].

In the third step, we calculated local normalized entropy (NE) of velocity directions [210] following work in the flow visualization literature (e.g. [159, 269]). The *NE* is close to 0 if the velocity direction closely concentrates one value out of N possible values (see Fig. B.1(a); NE=0.05). In contrast, the entropy measure *NE* becomes

0.95 if the probability of velocity directions is almost equally likely, as shown in Fig B.1(b). Given an arbitrary vexel located at *(x,y,z)* within the dome of an IA, we selected a fixed volume of interest (VOI; N_x x N_y x N_z; $N_x = N_y = N_z = 11$ in this study) centered at the voxel. One additional metric *H(x,y,z)* can be obtained by combining *K(x,y,z)* together with the *NE(x,y,z)* as follows,

$$H(x, y, z) = K(x, y, z) * NE(x, y, z) \tag{B.2}$$

H(x,y,z) is a scalar field representing the likelihood of residing within a vortex core region for a location (x,y,z). H also has a normalized range between 0 and 1. Thus, based on a fixed threshold, the vortex core region in this study can be obtained using the classic Marching-cube method [157]. in this study, 0.30 was used as the threshold for all data sets.

HelloWorld.c

```
// HelloWorld.c
// C program to display 'Hello, World!' in the terminal.
//
// Compilation:
// gcc -g -Wall HelloWorld.c -o HelloWorld.x
//
// Execution:
```

```c
// ./HelloWorld.x

// Standard headers
#include <stdio.h>

// main() begins
int main() {

  // Print the message
  printf("\n  Hello, World!\n\n");

  // Indicate the termination of main()
  return 0;
}
// main() ends
```

Appendix C

Letters of Permission

Include letters of permission from journal editors and/or other sources from which you may have used materials (images, information, etc.) in this this work.

These materials may also be submitted separately to the Graduate School as a single, well-organized PDF file.

Appendix D

Cellular Biology

TUNEL-assay

Terminal deoxynucleotidyl transferase dUTP-biotin nick end labeling (TUNEL)is an assay for detecting DNA fragementation: an aspect of cellular damage and apoptosis. TUNEL uses the enzynme terminal deoxynucleotidyl transferace (TdT) to attach labeled deoxyuridine triphosphate (dUTP) onto the 3'-hydroxyl termini of internucleosomal DNA fragmentation. Modification of dUTP through the addition of fluorphores or haptens, such as biotin, allow for DNA fragments to be detected directly using a fluorescently-modified nucleotide and fluorescence microscopy or flow cytometry.

VCAM-1

VCAM-1 is a member of the immunoglobulin superfamily (cell surface and soluble proteins involved in the recognition and/or binding of cells) and encodes a cell surface sialoglycoprotein (sialic acid and glycoprotein combination) expressed by cytokine-activated endothelium. This membrane protein acts as a ligand for leukocyte-endothelial cell adhesion, signal transduction, and may play a role in the development of artherosclerotic and/or inflammatory based pathologies. Molecules containing VCAM-1 counterreceptors (VLA-4 on monocytes and lymphocytes) can adhere to VCAM-1 activated cells[131]. Bound leukocytes may undergo polarized motility into the vascular wall, disrupting the cellular and matrix components of the vasuclautre, and degrading endothelial cell permeability.