Gas Embolic Factors in Cardiovascular Health

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GAS EMBOLIC FACTORS IN CARDIOVASCULAR HEALTH

by

Eric L. Cutler

A THESIS

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Gas emboli are a clinical issue often encountered in the space exploration, marine construction, and medical fields. Individuals involved with these disciplines will often have asymptomatic gas emboli circulating throughout their bodies on a regular basis. A study into the impact of chronic asymptomatic gas emboli on the risk of atherosclerosis in humans is thus presented. This research utilized a custom built cardiovascular flow simulator and accompanying data acquisition system to experimentally determine the influence of gas emboli under varied circumstances on the mean wall shear stress of a tube in-vitro. A directly inverse relationship between gas embolus presence in the blood stream and mean wall shear stress, particularly at low vascular flow and pulse rates was subsequently found. While the study was unable to support or refute any significant effect of gas emboli on the onset of atherosclerosis it did establish a positive causal link between bubbles in the bloodstream and diminished mean wall shear stress.
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"There is no such thing as a 'self-made' man. We are made up of thousands of others. Everyone who has ever done a kind deed for us, or spoken one word of encouragement to us, has entered into the make-up of our character and of our thoughts, as well as our success."

- George Matthew Adams

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Finally, the love and tolerance of both family and friends has been an unprecedented source of strength. In particular I would like to thank my parents, Dennis and Eileen, for always unquestioningly believing in me. In the same vein I would be remiss without acknowledging my god parents Nelson and Sharon Newman, who have done far too much for me to be mentioned here. I also want to thank Stephanie Wawers for gracing my life with her presence, and ultimately giving it a greater purpose. Finally, Ryan McCormick has been a consistent companion and source of scholarly motivation for the entirety of my postsecondary education.
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Glossary of Terms

Atherosclerosis - a process of progressive thickening and hardening of the walls of medium-sized and large arteries as a result of fat deposits on their inner lining.

Carotid artery - An artery located in the front of the neck through which blood from the heart travels to the brain.

Coronary branch/arteries - The vessels that encircle and supply the heart muscle with oxygenated blood.

Data acquisition system - an electronic system used to sample digital and analog signals.

Decompression sickness - the bends, a complex resulting from changed barometric pressure that consists of a rapid buildup of nitrogen bubbles in the bloodstream and tissues.

Diastole - the time period when the heart is in a state of relaxation and dilatation (expansion).

Embolism - an embolus lodged in the vasculature creating a partial or complete occlusion of blood flow.

Embolus - an abnormal particle entrained within the bloodstream.

Endothelium - A layer of flat cells lining the closed internal spaces of the body such as the inside of blood vessels and lymphatic vessels.

Extra vehicular activity - a spacewalk, tasks performed external to a pressurized space craft.

Hemodynamics - the physical flow of blood or the study thereof.

Hypotension - blood pressure insufficient of normal or healthy levels.

Hypovolemia – an abnormal decrease in the blood volume within an organism, particularly in the volume of plasma.

Hypoxia - a subnormal concentration of oxygen within a tissue.

Ischemia - inadequate blood circulation to a localized area due to the blockage of a blood vessel.

Lumen - the cavity or channel within a tubular structure such as a blood vessel or the intestines.
Myocardial infarction - a heart attack, the death of heart muscle arising from a lack of oxygenated blood supply

Pneumatic caisson - a structure used in submarine construction consisting of an open bottomed chamber

Shear stress - the force applied by, and parallel to, a flow stream to its adjacent matter as they move relative to one another

Standard temperature and pressure - STP, 1 atm and 20˚C. Default conditions for stated material properties

Stroke - the death of brain tissue arising from a lack of oxygenated blood supply

Systole - The time period when the heart is contracting, the period specifically during which the left ventricle of the heart contracts
Chapter 1 Background

1.1 Embolus Introduction

An embolus is best described as an abnormal particle entrained within the bloodstream. As an embolus is carried along with the surrounding circulatory flow it may become lodged within the narrowing vasculature. In the event that a particle is wedged within a blood vessel it is redefined as an embolism [1].

Subsequently, pressure builds upstream of the blockage and the blood flow is partially or completely occluded. If such an obstruction occurs in an artery or enough capillaries within a region the result is ischemia and tissue death. This clinical occurrence often has sudden and severe consequences including, but not limited to, ischemic stroke and myocardial infarction [2]. Because of the apparent and potentially fatal nature of large embolisms this phenomenon has been thoroughly studied.

These emboli may occur in a variety of forms (thrombotic, septic, foreign body, cholesterol, fat, amniotic, etc.) and for a variety of physiological reasons. For example, amniotic emboli (composed of amniotic fluid) leak into the circulatory system as a side effect of childbirth and septic emboli (consisting of pus) are created by cardiovascular infections [2]. Gas emboli, however, are one of the most clinically common and problematic forms. As gas becomes entrained in the bloodstream it may take on a variety of forms, from large elongated bubbles spanning the lumen to bubbly foams [3].

1.2 The Spencer Scale

In order to gain a more complete understanding of the proceeding research it is first imperative to become familiar with the relevant preceding work. One of the earliest and most prolific researchers into gas emboli and their effects was M.P. Spencer [4]. Most notably, he
developed a scale to describe the rate and size of gas emboli. This rating system is still in use today for clinical and research applications and is commonly referred to by its originator’s name. This scale rates embolus presence in the bloodstream from 0 – 5 with 0 indicating a complete lack of bubbles and 5 being almost entirely gas [5]. While not a particularly exact quantification this approach is a proven way to approximate the type of flow. The breakdown of the Spencer gas embolus scale is illustrated in Table 1. It should be noted that the Spencer rating of 5, or “white out” state, has never been documented in human subjects [6]. However, if such a phenomenon were to occur it would almost certainly be fatal.

### Table 1: Spencer gas embolus scale

<table>
<thead>
<tr>
<th>Rating</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>No bubbles</td>
</tr>
<tr>
<td>I</td>
<td>Occasional bubbles</td>
</tr>
<tr>
<td>II</td>
<td>At least one bubble per fourth cycle</td>
</tr>
<tr>
<td>III</td>
<td>At least one bubble per cycle</td>
</tr>
<tr>
<td>IV</td>
<td>Continuous bubbling, at least on bubble per cm² in all frames</td>
</tr>
<tr>
<td>V</td>
<td>“White Out”, bubbles can no longer be seen</td>
</tr>
</tbody>
</table>

#### 1.3 Entrance Mechanisms

In order to fully understand the medical implications of gas emboli it is first imperative to discuss the mechanisms through which they enter the cardiovascular system. The two most prominent entrance mechanisms for gas emboli are as a result of a medical procedure or through environmental decompression [1]. Other causes do exist, for instance child birth as previously mentioned. However, these only represent a small minority of documented gas embolus cases.
1.3.1 Surgical Procedures

All surgical procedures carry an inherent risk of inducing gas emboli and are therefore relevant in almost all medical fields [3]. However, the likelihood of this clinical event is highly dependent upon several risk factors. Whenever the bloodstream is exposed to a gaseous body emboli will occur when the atmospheric pressure exceeds the instantaneous blood pressure.

This severity of a gas embolus or embolism is commonly considered a function of both patient position and hemodynamic condition [1] [2] [7]. Dependent on how the subject is situated the gas bubble will travel to different parts of the body resulting in different clinical outcomes. For instance, when a patient is operated on while in the left lateral position it increases the likelihood of ensuing emboli to become trapped in the left ventricle of the heart [7]. It was formerly assumed that the buoyancy of the bubbles caused these effects, however, this has been recently refuted. Buoyancy is not sufficient to overcome the propulsion of surrounding blood flow [1] [7]. Hypotension (abnormally low blood pressures) may also have an aggravating effect on the ischemia caused by obstructive embolisms [2]. Hypovolemia (a decrease in plasma volume) will conversely increase the volume of emboli thus increasing the probability of forming an embolism.

However, the type of operation being performed can be even more important than patient condition. When incisions are made into the venous circulation or any blood vessels above the heart there is a particularly high risk of embolus entrainment (both are cases where the bloodstream undergoes transient negative pressures) [3]. Several studies have been conducted to quantify the rate of embolism formation for various specific medical procedures using several forms of detection. A table displaying a sample of these findings is shown in Table 2 [1] [2] [7] [8].
Table 2: Incidence rates of gas emboli for surgical procedures

<table>
<thead>
<tr>
<th>Procedure</th>
<th>% with Gas Emboli</th>
</tr>
</thead>
<tbody>
<tr>
<td>Craniotomy</td>
<td>58</td>
</tr>
<tr>
<td>Caesarian Section</td>
<td>65</td>
</tr>
<tr>
<td>Venous Catheterization</td>
<td>24.3</td>
</tr>
<tr>
<td>Total Hip Replacement</td>
<td>30</td>
</tr>
<tr>
<td>Some Neurosurgeries</td>
<td>up to 80</td>
</tr>
</tbody>
</table>

1.3.2 Atmospheric Decompression

Atmospheric depressurization creates emboli through the phenomenon of Boyle’s law, first proposed by physicist Robert Boyle in 1662 [2]. This theory states that the pressure and volume of an ideal gas at constant temperature within a closed system are inversely proportional [9]. Thus, when the pressure of a subject’s environment decreases (and the absolute blood pressure concurrently decreases) the volume of the gases in the bloodstream conversely increases. There are constant inert gases, such as nitrogen, in solution in the circulatory system but as gas volume increases these gases come out of solution and coalesce to form bubbles. Boyle’s law is illustrated mathematically below where $p$ is pressure, $V$ is volume, and $k$ is Boyle’s constant (a gas material property) [10].

$$pV = k$$

In addition to gas emboli some examples of this phenomenon include ears popping at higher altitudes, and hydraulic braking systems. If the atmospheric pressure drop is large enough, or rapid enough, the subject will experience decompression sickness (or more colloquially “the bends”) [5]. This condition involves not only bubble formation in the circulatory system but also in the joints of the extremities. The symptoms include acute abdominal and joint pain, skin rash,
vertigo, and nausea, and this condition can even be fatal [10]. A clinical case of severe
decompression sickness is displayed below in Figure 1. Fortunately, this clinical event occurs
only for a minority of gas embolus cases and has, due to its debilitating nature, been thoroughly
studied.

![An example of decompression sickness symptoms](image)

**Figure 1: An example of decompression sickness symptoms**

The circumstances for environmental depressurization are associated with a few
occupations and leisure activities. These include, but are not limited to, scuba diving, marine
construction, aviation, and space exploration [1]. For this reason several professional
organizations have conducted long term studies into the incidence and implications of gas emboli
for their respective members.

### 1.3.2.1 Marine Submersion

Commercial diving is one of the most common fields in which the effects of
environmental decompression are observed and documented. As divers descend from the surface
the hydrostatic pressure exerted upon them increases linearly with depth according to the
equation \( p = \rho gh \) where \( p \) is pressure, \( \rho \) is liquid density, \( g \) is the gravitational constant, and \( h \)
denotes depth below the surface [5]. As the diver subsequently begins his/her ascent the resulting
pressure decrease creates gas emboli [11]. Although protocols in rate of ascent have been
introduced in order to mitigate this effect there is no known way to eliminate it. In fact, in a study conducted by Spencer as many as 36% of dives produced detectable gas emboli [12]. Another survey by the Divers Alert Network of Europe found that 37.4% of dives created bloodstream bubbles [13]. Emboli categorized as I-II on the Spencer scale accounted for 25.4% of cases and III-IV accounted for 12% of cases.

This effect is also seen in the closely related field of marine construction. A common example would be the erection of caissons, a type of water tight retaining structure. The pneumatic caisson was first patented in 1889 by William M. Patton in Philadelphia to be used in excavating and pouring the foundations of bridges, piers, and dams [14]. These concrete shells are filled with compressed air in order to keep soil and water from flowing in at the base. As workers exit the interior of the caisson they undergo decompression that can be quite rapid [10]. In fact, decompression sickness was first documented as a result of this construction process and was originally known as caisson disease. An early caisson design is depicted in Figure 2 in order to provide an example profile (each airlock being labeled D) [14].

---

1.3.2.2 Space Exploration

Another important field for the study of gas emboli is that of space exploration. Often astronauts are required to perform tasks outside of the confines of their spacecraft [15]. Extra
vehicular activity (EVA), more commonly referred to as a spacewalk, is conducted within a sealed low pressure suit [16]. The subject must undergo barometric decompression when transitioning from their pressurized craft to their EVA suit. As an example, the pressure drop of going from the space shuttle (23.4 psi) to a standard National Aeronautics and Space Administration EVA suit (4.3 psi) is 19.1 psi, 1.3 times atmospheric pressure at sea level [17]. This, of course, leads to a nontrivial incidence of gas emboli.

Fortunately, NASA and several other space agencies have been proactive about documenting and researching this phenomenon. In fact, since 1982 NASA has operated the Hypobaric Decompression Sickness Database as an ongoing project of the Environmental Physiology Laboratory at Johnson Space Center [18]. This is a depository for all recorded cases of decompression sickness, and furthermore gas emboli, that occur during training, experimentation, and in the field. The longitudinal time frame and varied test parameters provide a rich and useful source of data. As of 2007 the Environmental Physiology Lab had performed a total of 783 controlled decompression exposures with 326 exhibiting detectable gas emboli, an overall rate of 41.6%. Of all decompressions 5.2% were categorized as I on the Spencer scale, 6.8% were II, 10.7% were III, and 18.9% were IV [17].

Concern about the cardiovascular health of astronauts has increased even further as a result of recent events. In 2004, subsequent to the Space Shuttle Columbia disaster, then-President George W. Bush outlined a new agenda for American space exploration. This plan detailed several high priority goals including a permanent or long duration human presence on the moon by 2020 and subsequent human exploration of Mars [19]. Although the funding of this plan has been somewhat diminished by the subsequent administration its ambitious nature has remained intact.

The effects of the long duration missions necessary to meet these goals remain largely unknown [15]. This has considerably raised the concern amongst the space exploration community as to the dangers posed to astronauts. In fact, as a result of this new direction of
extra-planetary efforts NASA ordered a study into potential medical issues, entitled the 
Bioastronautics Roadmap: A Risk Reduction Strategy for Human Space Exploration. This 
investigation identified alterations of cardiovascular function and structure to be major risks [20]. 
Indeed, the occurrence of several circulatory disorders, such as orthostatic intolerance, has been 
well documented as a side-effect of short term extraterrestrial travel [21] [22]. The grave nature 
of a possible space related health incident was echoed in a recent study, “due to the limited 
treatment and return capabilities of most space vehicles, an in-flight cardiac event would result in 
mission failure” [23].

1.4 Wall Shear Stress and Endothelial Function

Another important consideration to introduce is the causal link between deficient wall 
shear stress and the onset of atherosclerosis. Shear stress is the force applied by and parallel to a 
flow stream to its adjacent matter as they move relative to one another. This can be thought of as 
the liquid equivalent of friction and when summed over a surface is referred to as drag [24]. A 
typical formulation for shear stress in a two dimensional plane is shown below where $\tau_{xy}$ denotes 
shear stress, $\mu$ denotes dynamic viscosity, $u$ denotes stream velocity, and $y$ denotes distance 
normal to the flow stream [9].

$$\tau_{xy} = \mu \frac{du}{dy}$$

Atherosclerosis is an increasingly common and potentially fatal cardiovascular condition. It is characterized by the hardening and narrowing of the arteries [25]. This plaque buildup 
slowly occludes blood flow until it either completely blocks the vessel or ruptures spilling large 
emboli downstream. In either case the result is often an ischemic stroke or myocardial infarction. 
This disorder even has become a leading cause of death amongst industrialized nations [26].

Since the early 1970s evidence has been mounting in support of the somewhat 
counterintuitive theory that low shear stress leads to increased risk of atherogenesis. It was first
suggested when cardiologists noticed a pronounced concentration of atherosclerotic lesions near curvatures and bifurcations, points where blood vessel geometry becomes more complex. Specific examples of notable locations for the onset of atherosclerosis are the aortic bifurcation, carotid bifurcation, coronary arteries, and in the distal abdominal aorta [27]. In these regions the flow separates and becomes more turbulent which contributes to a drop in wall shear stress [28]. The hypothesis that local hemodynamics governed the onset of atherosclerosis was initially one of several competing ideas. However, experiments using endothelial cell cultures and animal subjects soon confirmed that low shear stress was a primary culprit [29].

It should be noted that insufficient wall shear stress contributes to the onset of lesion formation but not to later stages. This adds to the complexity involved in quantifying the influences responsible for atherosclerotic formation. The endothelium is extremely sensitive to its environment and will adapt accordingly as that environment changes. As wall shear stress decreases this cellular lining will begin to express a different phenotype and produce different chemical signals [30]. At physiological shear stress levels endothelial cells are fusiform and aligned with flow direction which allows them to fit closely together and create an effective barrier [26]. However, at lower shear stress regimes these cells assume a polygonal shape and random orientation thus leaving gaps for possible fluid flow. Also, cellular turnover is much higher at normally occurring shear stresses, further serving to increase these gaps [31]. An increased production of adhesion molecules is observed with reduced wall shear as well.

The shape change and altered cellular production rates ultimately result in the increased permeability of the blood vessel wall [31]. This permeability allows particles such as cholesterol, calcium, red blood cells, monocytes, and foam cells to accumulate in the vascular intima thus altering the geometry of the blood vessel. This buildup is accelerated by the aforementioned presence of adhesion molecules, which attract these molecules to the arterial wall [32]. The buildup of these substances injures surrounding tissues which leads to an additional inflammatory
response, furthering the occlusion of blood flow [29]. Of course, a therogenesis can also be exacerbated by other conditions such as diabetes, hyperlipidemia, and tobacco smoking [25]. This process is illustrated below in Figure 3.

![Figure 3: Formation of atherosclerotic lesions](image)

Thus, if wall shear stress were to be chronically lowered in a region of the arterial tree it would subsequently increase the risk of plaque formation. While this trend was originally discovered through geometric observations differences in vessel shape are not necessary for a decrease in wall shear stress. Although this theory was originally rather controversial it has been confirmed by literally dozens of studies over the past three decades [33] [30] [31] [25] [34] [35] [26] [29] [32] [28] [36] [27] [37].

Of course, the important task is quantifying what exactly denotes low shear stress. Generally, the abovementioned studies placed an approximate lower wall shear stress threshold for the onset of atherosclerosis. Differing methodologies and objectives had to be taken into consideration when summarizing the results of this body of academic work. Various studies investigated multiple arteries while utilizing multiple techniques to achieve multiple goals. Early on in the exploration of this phenomenon there was less agreement between conclusions drawn by different researchers (ranging from 3 to 12 dynes/cm²) [27] [35] [31]. However, over time those
studying this issue came to a much better consensus. Currently a mean wall shear stress level at or below 6 dynes/cm² is considered to be a very accurate predictor of atherogenic behavior [37] [32] [30] [33] [26]. It should be noted that this is the time averaged shear stress imparted on the vessel wall. There is some evidence that wall shear stress maxima and amplitude also contribute to the onset of atherosclerosis [29] [28]. However, this is a more controversial and far less established hypothesis. Thus, mean wall shear stress will be the focus of this work.
Chapter 2. Objective

Certain occupations (commercial diving, maritime construction, aviation, and space exploration) and medical procedures (hemodialysis and catheterization) expose their practitioners and patients to frequent bloodstream gas bubbles. The presence of gas emboli will lower the density of the vascular flow and thus potentially reduce the shear stress acting on the artery wall. It has been well established that a reduction of wall shear stress contributes significantly to the onset of atherosclerosis (particularly below 6 dynes/cm$^2$). Thus, the objective of this study was to quantify the relationship between the presence of bubbles and mean wall shear stress in order to evaluate the potential risk of atherosclerosis posed by chronic gas emboli.

In order to test this hypothesis a cardiovascular flow simulator was designed and built in order to reasonably replicate the physiological conditions for experimentation. The embolism injection rate, pulse rate, and time-averaged flow rate were varied in order to test the effect on mean wall shear stress at the most atherosclerosis prone arteries at resting and elevated heart rates. By recording the pressure drop across and volumetric flow through a section of straight tube the subsequent mean wall shear stress under these circumstances was calculated and compared to data in the relevant literature.
Chapter 3. Experimental Apparatus

3.1 Flow Simulator

The testing apparatus was designed, built, and refined over two semesters in order to conduct experiments on the wall shear stress effects of gas emboli. The primary functions of this flow simulator were to produce fluid flow, measure relevant parameters, and record them. It was also necessary for it to be capable of creating both static and pulsatile flow. Due to limited funding it was also necessary to constrain the costs of construction. A picture of the finished flow simulator design (subsequent to testing) is shown in Figure 4.

![Figure 4: Physiological flow simulator profile](image)

As with all hydraulic circuits this system contained a reservoir (labeled A) to contain the working fluid. This vessel is necessary to hold excess fluid, provide an outlet for gas bubbles to escape, and provide hydrostatic pressure. By elevating the reservoir above the rest of the plumbing a pressure is created which feeds the pump for pulsatile flow and propels static flow, depending on the configuration. The reservoir was created from a Coleman 6278-718G commercial cooler, measuring 19.7” x 12” x 13.7”. Exit and entrance holes were added and the subsequent piping was sealed using generic plumber’s putty.
A pulsatile pump (labeled B) was also included in order to provide the time dependent flow characteristic of the circulatory system. This pump was designed for experimental work on medium to large sized animal test subjects with the use of blood or blood stimulants and was somewhat ideal the purposes of this work. This Harvard Apparatus (no affiliation with Ivy League institution) Model 1421 Pulsatile Blood Pump (part # 55-3321) had variable settings for stroke volume, stroke rate, and systole/diastole ratio (the ratio of piston compression time to piston expansion time). These adjustable parameters ranged from 4-30 mL for stroke volume, 20-200 strokes/min for stroke rate, and 25/75 to 50/50 for systole diastole ratio.

In order to create a sufficient steady stream of gas emboli a syringe pump (labeled C) was purchased and installed. This type of pump creates a static flow and is little more than a ball screw with a controlling microprocessor. Attaching a syringe to the top of the ball screw creates a cylinder and piston sufficient to mechanically induce flow. This arrangement also allows for the syringe to by changed in order to suit the needs of the experiment. By using a 60 mL syringe a flow rate of up to 30 mL/min was possible. The air flow created by this pump entered the primary circuit just prior to the test section. A more detailed view of the various components is shown in Figure 5.

Figure 5: Close ups of various apparatus components
The flow exiting the pump passes through an ultrasonic flow meter (labeled D) in order to measure and record the volumetric flow of the working fluid. This type of flow meter relies on the Doppler effect to accurately and non-invasively calculate the average flow velocity across a section of known diameter. A sonic emitter/sensor pair is installed in each end of the flow sensor in order to send alternating bursts of noise back and forth, an arrangement known as the pi method. By measuring the subsequent phase shift in the signal from emission to receiving the average flow velocity may be measured. The Shenitech STUF-300F1G ultrasonic flow meter consisted of not only the flow vessel but also a self contained data processor operated on 20 Vdc. The inner diameter of the flow cell was 3 mm and the meter was capable of measuring flow velocities up to 18 m/s in either direction with an accuracy of ±0.1% of full range. A schematic of a similar vessel design is shown in Figure 6.

![Figure 6: Pi method ultrasonic flow meter](image)

The test section (labeled E) across which the bubbly flow passes and data are recorded was a ¼” inside diameter rubberized latex tube. This tube was fixed on barbed fittings at each end and held straight. Unfortunately, it appears skewed in Figure 4 and Figure 5 due to an accident subsequent to data collection. Using c-clamps the tube length was maintained at a distance of 6”. The main plumbing tubes for the flow simulator were ½” ID clear vinyl tubes and the tubes leading to the pressure sensors were .17” ID and of the same material.

Pressure sensors (labeled F) were situated in a breadboard with fluid filled lines leading to locations upstream and downstream of the test section. One of the two differential pressure
transducers was connected to the upstream location while the other outlet was open to atmosphere. This sensor measured the gage pressure at the beginning of the test section. The other transducer was connected to both the up and downstream terminals, thus recording the pressure drop across the experimental tube. Each of these was an All Sensors 5 PSI-D-4V-ASCX differential pressure transducer. This model has an operating range of 0-5 psi, a proof pressure of 15 psi, and a burst pressure of 30 psi. Each sensor was powered by 5 Vdc with a full output range of 0-4 Vdc and an accuracy of ± .1% of full range.

The data from both of the pressure sensors and the flow meter were subsequently recorded by the data acquisition system (or DAQ, labeled G). The National Instruments USB-6009 DAQ was used to record test data in real time. This inexpensive system contains 8 analog inputs (14 bit, 48 kS/s), 2 analog outputs (12 bit, 150 S/s), 12 digital I/O pins, and a 32 bit counter, which were more than sufficient for the purposes of this work. The data from the DAQ were fed to a PC where they were processed and stored by the National Instruments LabVIEW SignalExpress software. This program was later used to export test data into easily analyzed ASCII files. It should be noted that all experiments were sampled for a duration of ten seconds at a rate of 30 Hz. No aliasing or other unwanted phenomena were ever observed as a result of the sampling rate. Three of the available analog inputs (for the two pressure transducers and ultrasonic flow meter) and one of the analog outputs (to provide excitation voltage for both pressure sensors) were used.
By manipulating various globe valves it was also possible to toggle the flow simulator between pulsatile and static test arrangements. The essential difference between these two configurations is that the static setup bypassed the pulsatile pump. Additionally, the pulsatile mode is a true circuit, with the working fluid returning to the reservoir. However, for the static setup the flow is simply released into a lower drain. Each of these setups is illustrated in Figure 7 and Figure 8, with red lines indicating main paths of active flow and black lines indicating pressure sensor tubes.

### 3.2 Physiological Validation

In order for the results of this study to be valid it is vital that physiological conditions be accurately replicated. The goal was to test the effect of gas emboli in the carotid and coronary arteries under resting and elevated heart rates. These blood vessels are particularly prone to the onset of atherosclerosis and the consequences of occlusion are very grave [27]. The major considerations of this accuracy were the viscosity and density of the blood simulant, the
dimensions of the test tube, the volumetric fluid flow, the rate of embolism injection, and the pulsatile pump rate.

### 3.2.1 Blood Simulant

Due to the practical complications of using actual blood a simulant was used instead for this study. This mixture consisted of glycerin, water, and cornstarch. Aqueous glycerol solutions have been reliably used by researchers for decades to replicate the fluid mechanical properties of blood [38] [39]. The glycerin was initially mixed in the water solvent to a concentration of 40% by volume, the most commonly used ratio for a study of this nature [38] [39] [40] [41]. This solution is well documented as approximating both the viscosity and density of blood for low frequencies (on the order of 100 bpm), moderate temperature, and macroscopic scales [41]. 1 g of cornstarch was subsequently added to the mixture for each liter of liquid. This step was not taken for mechanical considerations but instead to increase sonic reflectivity, the rate at which the medium reflects rather than transmits sound. The increased reflectivity of a medium makes the ultrasound flow meter measurements more reliable and stable.

### 3.2.2 Vascular Dimensions

It was also imperative for the cross-sectional shape and diameter of the test section to approximate those of the carotid and coronary arteries. While there is some biological variability to these parameters, as is expected with any tissue property, several studies have found population averages for several different genders, ages, etc. The average inside diameter of the common carotid artery (which leads from the heart, up the neck, and to the brain as shown in Figure 9) was found to be 6.11±.80 mm for women and 6.52±.98 mm for men while the average inside diameter of the internal carotid artery was 4.66±.78 mm for women and 5.11±.87 mm for men in a 2006 study [42]. The average inside diameter for the common carotid artery was found to be 5.55 mm on the right side and 5.38 mm on the left side for women while it was 6.21 mm on the right side and 6.15 mm on the left side for men according to a 2000 study [43]. The slight discrepancy
between these results likely stems from different methodologies. In any case, the carotid artery diameter appears to remain on the order of 5-6 mm in the general populace.

![Anatomical diagram of the carotid artery](image)

**Figure 9: Anatomical diagram of the carotid artery**

The coronary artery supplies blood to the heart, as shown in Figure 10, and is slightly smaller than the carotid artery. According to one 2002 publication the diameter of the various branches of the coronary system vary from $3.30 \pm 0.85$ mm to $5.07 \pm 0.75$ mm, however the majority of these arteries remain in the 3.5 mm range [44]. Another 1970 study found that the majority of the coronary branch had a diameter on the order of 3 mm [45]. Finally, a 2000 study found an average coronary diameter of 3.63 mm [46]. With the two arterial trees (carotid and coronary) staying typically within the range of 3.5-6.5 mm the nearest available standard tube size was $\frac{1}{4}$" (6.35 mm). In general this size is an excellent approximation for the mid-sized arteries of the human circulatory system.
3.2.3 Vascular Volumetric Flow

The volumetric flow of blood was also an important consideration for experimental design. While the same tube was used for both the carotid and coronary settings separate flow settings had to be used for each of the two test configurations. The amount of blood pumped through the carotid area is actually much larger than that of the coronary branch. This most likely arises from the larger amount of tissue nourished by the carotid branch [47]. A number of studies have been undertaken to ascertain an average flow for the carotid artery for a variety of purposes. The results of a selection of 4 studies is shown below in Table 3 [47] [48] [49] [50]. Based on the agreement of these results a carotid time averaged volumetric flow rate of 280 mL/min was assumed.

<table>
<thead>
<tr>
<th>Author</th>
<th>Year</th>
<th>Method</th>
<th>Cycle-Averaged Flow (mL/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bendel et al.</td>
<td>1989</td>
<td>MR Imaging</td>
<td>250-580</td>
</tr>
<tr>
<td>Ford et al.</td>
<td>2005</td>
<td>MR Imaging</td>
<td>275 ± 52</td>
</tr>
<tr>
<td>Ho et al.</td>
<td>2002</td>
<td>Doppler Imaging</td>
<td>312.6</td>
</tr>
<tr>
<td>Schoning et al.</td>
<td>1994</td>
<td>Sonography</td>
<td>265 ± 62</td>
</tr>
</tbody>
</table>
Inquiry into the volume carried by the coronary artery has also been quite prevalent, although not as common as for the carotid branch. However, it appears that thermodilution was the preferred method of measurement for cardiac studies. A pair of studies (one old and one recent) are presented in Table 4 [51] [52]. The flow of blood feeding the heart has indeed become rather well established [52]. Based on this literature review a coronary volumetric flow of 130 mL/min was assumed.

<table>
<thead>
<tr>
<th>Author</th>
<th>Year</th>
<th>Method</th>
<th>Cycle Averaged Flow (mL/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ganz et al.</td>
<td>1971</td>
<td>Thermodilution</td>
<td>122 ± 25</td>
</tr>
<tr>
<td>Aarnoudse et al.</td>
<td>2007</td>
<td>Thermodilution</td>
<td>141 ± 55</td>
</tr>
</tbody>
</table>

### 3.2.4 Embolism Rate

Information on naturally occurring asymptomatic gas embolism rates unfortunately was somewhat difficult to locate. This is due to the relative hardship encountered in quantifying embolus size in real time as well as the risk of purposely inducing an embolism in a patient (the reason most relevant experiments were conducted with dogs). It may have also been a consequence of the relative obscurity of the subject. However, with some intrepid research a variety of sources were uncovered.

One review based on available case data found that a gas embolus of 300-500 mL injected at 100 mL/sec is fatal for humans [53]. Another review of available clinical cases found that a gas infusion rate of 1 L/min is lethal for humans in less than 2 minutes and that a gas embolus of 2 mL/kg of body weight would cause tachycardia, hypotension, and raised central venous pressure [54]. An experiment conducted on dogs found that gas emboli of a volume of 1-1¼ mL/kg of body weight were lethal for 50% of subjects and ½ mL/kg of body weight resulted in permanent detectable brain damage [55]. An earlier canine study found that embolus
entrainment rates less than or equal to 1 mL/kg of body weight/min were non-lethal in test subjects [56].

However, one of the most comprehensive works quantifying gas emboli was a gynecological article. This review found that gas emboli resulting from surgical procedures generally occur at rates lower than 1 mL/kg/min [57]. It also concluded that a bubble entrainment rate of .5 mL/kg/min caused irreversible brain damage. However, most notably this review stated that a gas embolus rate of .1-.4 mL/kg/min administered intravenously was non-fatal in humans [57]. However, the upper end of this range would not be asymptomatic but still caused decompression sickness like expression. Indeed asymptomatic gas emboli were found to not exceed .2 mL/kg /min [57].

Of course, this value is still in units scaling with body weight, which is not particularly useful for this in-vitro work. Thus, data regarding average American provided by the Center for Disease Control was utilized. The most recent available study, from 2004, found that the average body weight was 78.2 kg for men and 67.9 kg for women [58]. Since the study also reflected an almost even number of men and women in the United States these two values were averaged to result in an assumed average of 73.05 kg. Multiplying this mass by the abovementioned embolism rate resulted in an experimental gas embolus rate of 0-15 mL/min (rounded up from 14.61). Under the Spencer gas embolus scale these would qualify as I-II in classification [17].

3.2.5 Pulse Rate

The final test setting validation is that of pulse rate. Considering that for many of the aforementioned cases the subject would be undergoing the presence of gas emboli at both resting and elevated heart rates it was thought necessary to perform tests under both circumstances. These two settings were simply based on information acquired from the American Heart Association. The heart should beat approximately 70 beats per minute when at rest for a healthy
adult [59]. However, during or shortly after rigorous activity the pulse may range from 100-170 bpm [60]. The median value of this range, 135 bpm, was used as an approximation for the elevated heart rate of a healthy adult.
Chapter 4. Experimental Methodology

4.1 Testing Configurations

As with any experiment, it is imperative to provide a thorough explanation of the methodology used. The object of this study was to investigate the effect of gas emboli on wall shear stress under varying parameters (fluid flow, pump frequency, and gas embolism rate). A series of static flow tests were conducted in order to acquire a baseline of data with minimal transient effects. Thus, after the results from this round of tests were found to be satisfactory the pulsatile set of tests would then be conducted.

Both the static and pulsatile experiments were also conducted at different volumetric flow rates to simulate the carotid and coronary arteries. This flow level was controlled by the pump stroke rate and stroke volume settings \( \left( \frac{\text{strokes}}{\text{min}} \times \frac{\text{mL}}{\text{stroke}} = \frac{\text{mL}}{\text{min}} \right) \) or reservoir liquid level and circuit needle valve, depending on the configuration. The time averaged volumetric fluid flow value corresponding with carotid simulation was 280 mL/min while the flow corresponding with coronary simulation was 130 mL/min.

For pulsatile experiments the tests were further subdivided into resting and elevated experimental conditions. A resting heart beat was defined as 70 bpm while an elevated heart rate was defined as 135 bpm. This adjustment was easily made manipulating the pump controls. Of course, such adjustments were unnecessary for the static tests since the piston pump was not used.

Finally, at each particular flow and pulse setting tests were performed under four progressive gas embolus rates. These four values were 0 (baseline), 5, 10, and 15 mL/min spanning the range of asymptomatic gas emboli injection rates. In terms of Spencer Scale these emboli would be characterized on the order of I-II [17]. This hierarchical testing setup is somewhat complex (a parameter within a parameter with a parameter), thus a matrix form is provided in Table 5 to better explain the manner in which these experiments were conducted.
Table 5: Test configuration matrix

<table>
<thead>
<tr>
<th>Characteristic flow</th>
<th>Static</th>
<th>Pulsatile</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volumetric fluid flow</td>
<td>Carotid</td>
<td>Coronary</td>
</tr>
<tr>
<td></td>
<td>280 mL/min</td>
<td>130 mL/min</td>
</tr>
<tr>
<td>Frequency</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td>70 bpm</td>
<td>135 bpm</td>
</tr>
<tr>
<td>Embolism injection rate (mL/min)</td>
<td>0 5 10 15</td>
<td>0 5 10 15</td>
</tr>
</tbody>
</table>

Adding up all subcategories in the bottom row of Table 5 it can be seen that there are 24 different experimental settings. Adding to this the fact that each test setup was repeated four times to increase reliability and identify outliers and there was quite a bit of data generated. The results of each repetition were averaged in order to gain a more meaningful sample. However, the mathematical approach will be further discussed in the next chapter.

4.2 Flow Simulator Operation

Before conducting either type of test the apparatus would first have to be cleaned and prepared. If it had been over a day since the last use of the system an aqueous vinegar solution was poured into the reservoir and circulated using the pulsatile pump. This mildly acidic mixture provided a gentle non-caustic cleaning agent to remove any debris or residues from previous experiments. The system was subsequently emptied and refilled with water. This was then pumped through the flow simulator and drained. This second step was to remove any residual vinegar, thus helping to ensure that the test fluid is uncontaminated.

The less controllable nature of the static tests made for some additional setup. First, the liquid level of the reservoir was adjusted to create the necessary hydrostatic pressure. By
adjusting a needle a valve the flow caused by the hydrostatic pressure could be fine tuned to the intended volumetric rate (130 or 280 mL/min). After achieving the proper liquid level height all of the relevant globe valves were checked to insure proper alignment.

To begin each individual static test the syringe pump (with the intended embolisation rate) was activated and the drain valve was opened. The SignalExpress program was shortly thereafter prompted to begin recording data. After the passing of the prescribed test duration and cessation of data acquisition the drain valve was closed and subsequently the syringe pump was deactivated. Before the next test the reservoir level, syringe pump plunger, and miscellaneous valving would have to be restored to their initial states. The cross sectional area of the reservoir was much larger than the volumetric flow regimes ensuring a quasistatic pressure head for the duration of a given test and resetting the water level was not always necessary between tests. Obviously, this process required several hands and assistance was graciously provided by fellow graduate students.

Fortunately, conducting pulsatile tests was much simpler than static tests. By closing the pump bypass valve, opening the reservoir return valve, and adjusting the needle valve, the flow simulator was easily reconfigured into a closed hydraulic circuit. Before every test the settings of the pulsatile and syringe pumps would be adjusted to the desired values. Both pumps would then be activated and allowed a brief period to settle into steady state operation. This configuration allowed for multiple samplings to be conducted until the syringe pump was reset, unlike static tests. Of course, no matter the experimental arrangement the flow simulator was also closely monitored to ensure that all components were functioning properly.
Chapter 5. Mathematical Modeling

5.1 Calculation of Wall Shear Stress

The measurement of wall shear stress is an experimentally challenging proposition. Several non or partially invasive techniques are capable of accurately measuring wall shear stress such as laser particle velocimetry, video microscopy, and Doppler techniques [61] [62]. However, these approaches were largely unavailable due to budgetary limitations. It is, however, possible to calculate the wall shear stress based on measurements of related variables and parameters.

![Cylindrical Coordinate System](image)

A derivation was conducted in order to find the wall shear stress as a function of measurable properties (pressure drop and mean fluid velocity). The relevant starting point for a duct of this nature would be the conservation of momentum in the z-direction for a cylindrical coordinate system, an example of which is shown in Figure 11. This equation is given along with the definitions of the convective time derivative and Laplacian operator [9]. The relevant variables are fluid velocity (denoted by \( \mathbf{u} \)), time (denoted by \( t \)), density (denoted by \( \rho \)), pressure (denoted by \( p \)), the gravitational constant (denoted by \( g \)), and kinematic viscosity (denoted by \( \nu \)). The subscripts \( z \), \( r \), and \( \Theta \) denote the longitudinal, radial, and angular coordinates, respectively.
Conservation of z momentum: \( \frac{\partial v_z}{\partial t} + (V \cdot \nabla) v_z = -\frac{1}{\rho} \frac{dp}{dz} + g_z + \nu \nabla^2 v_z \)

Convective time derivative: \( V \cdot \nabla = v_r \frac{\partial}{\partial r} + \frac{1}{r} v_\theta \frac{\partial}{\partial \theta} + v_z \frac{\partial}{\partial z} \)

Laplacian operator: \( \nabla^2 = \frac{1}{r} \frac{\partial}{\partial r} \left( r \frac{\partial}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2}{\partial \theta^2} + \frac{\partial^2}{\partial z^2} \)

In order to remove unnecessary terms assumptions must be applied to the governing equation. In uniform cross section duct flow it can be assumed that it is axisymmetric (labeled as assumption 1). That is to say that the fluid is moving only along that length of the duct (in this case a latex tube). Since the test section was mounted horizontally gravity may be disregarded (labeled as assumption 2). Also, it will be assumed that the fluid flow is fully developed (labeled as assumption 3). This means that the velocity profile and boundary layer thickness do not change with longitudinal distance. These assumptions are shown below:

1) \( u_r = 0, v_\theta = 0, \frac{\partial}{\partial \theta} = 0; \) 2) \( g_z = 0; \) 3) \( \frac{\partial u}{\partial z} = 0 \)

The assumptions do a great deal to simplify the conservation of momentum equation. They completely eliminates the convective derivatives term and simplify the Laplacian operator to radial terms only. This removal of extraneous variables is displayed below:

\[
V \cdot \nabla = v_r \frac{\partial}{\partial r} + \frac{1}{r} v_\theta \frac{\partial}{\partial \theta} + v_z \frac{\partial}{\partial z} = 0 \quad \text{&} \quad \nabla^2 = \frac{1}{r} \frac{\partial}{\partial r} \left( r \frac{\partial}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2}{\partial \theta^2} + \frac{\partial^2}{\partial z^2} = \frac{1}{r} \frac{\partial}{\partial r} \left( r \frac{\partial}{\partial r} \right)
\]

\[
\frac{\partial v_z}{\partial t} + (V \cdot \nabla) v_z = -\frac{1}{\rho} \frac{dp}{dz} + g_z + \nu \nabla^2 v_z \quad \rightarrow \quad \rho \frac{\partial v_z}{\partial t} = -\frac{dp}{dz} + \frac{1}{r} \frac{\partial}{\partial r} \left( r \frac{\partial v_z}{\partial r} \right)
\]

Since viscosity (\( \mu \)) is a fluid property and constant along the axial (and any, for that matter) direction this coefficient may be moved inside of the parentheses to the right. By doing this the derivative function inside of the parentheses becomes the definition of shear stress (\( \tau \)) and
can be replaced. At this point the variable \( v_z \) will be redefined as \( u \), due to the fact that there are no longer any other velocity variables and the fact that this is more fitting with common syntax. This process is illustrated in the following equations:

\[
\rho \frac{\partial u}{\partial t} = - \frac{dp}{dz} + \frac{1}{r} \frac{\partial}{\partial r} \left( r \mu \frac{du}{dr} \right) \quad \text{&} \quad \tau = \mu \frac{du}{dr}
\]

\[
\rho \frac{\partial u}{\partial t} = - \frac{\partial p}{\partial z} + \frac{1}{r} \frac{\partial}{\partial r} (r \tau)
\]

However, one final step is necessary to get the above equation to contain the required variables. This goal can be achieved by integrating the governing equation in the radial direction from the tube centerline to the inner wall. By evaluating this integral the shear stress and fluid velocity (both of which are functions of radius) change from being radially dependent to radially independent. The velocity \( u \) becomes \( \bar{u} \), the mean velocity of the flow, and the shear stress \( \tau \) becomes \( \tau_\omega \), the shear stress at the vessel wall. This transformation is depicted in the following lines.

\[
\int_0^{D/2} \rho \frac{\partial u}{\partial t} \, dr = \int_0^{D/2} - \frac{\partial p}{\partial z} + \frac{1}{r} \frac{\partial}{\partial r} (r \tau) \, dr \quad \rightarrow \quad \int_0^{D/2} r \rho \frac{\partial u}{\partial t} \, dr = \int_0^{D/2} - r \frac{\partial p}{\partial z} + \frac{\partial}{\partial r} (r \tau) \, dr
\]

\[
\frac{r^2}{2} \rho \frac{\partial \bar{u}}{\partial t} \bigg|_{0}^{D/2} = - \frac{r^2}{2} \frac{\partial p}{\partial z} + \tau_\omega r \bigg|_{0}^{D/2} \quad \rightarrow \quad \frac{D^2}{8} \rho \frac{\partial \bar{u}}{\partial t} = - \frac{D^2}{8} \frac{\partial p}{\partial z} + \tau_\omega \frac{D}{2}
\]

As can be clearly seen the remaining variables are diameter, density, time derivative of average velocity, and pressure drop. These are all parameters which can be measured with the given test apparatus (except for density, which will be discussed in the subsequent section). Of course, the end of the above iteration must be refined and make wall shear stress (\( \tau_\omega \)) an explicit function. This final form is displayed below:

\[
\rho \frac{\partial \bar{u}}{\partial t} = - \frac{dp}{dz} + \frac{4 \tau_\omega}{D} \quad \rightarrow \quad \tau_\omega = \frac{D}{4} \left( \rho \frac{\partial \bar{u}}{\partial t} + \frac{dp}{dz} \right)
\]
This derivation was confirmed using a control volume approach instead of integrating the derivative form. The three main forces to be considered in duct flow are inertia, drag, and pressure drop. The inertial force is defined by Newton’s 1st Law while the other two are very basic fluid mechanical concepts. The definitions of these total forces on the control volume are shown below in Figure 12. The variable SA denotes the wetted surface area while V represents control volume.

\[ F = ma = \rho (\pi/4)D^2 z \left( \frac{d\bar{u}}{dt} \right) \]

\[ F_r = \tau_w (SA) = \tau_w \pi D z \]

\[ F_{\rho} = \frac{dp}{dz} V = (\frac{dp}{dz})(\pi/4)D^2 z \]

**Figure 12: Control Volume Approach**

By creating a simple force balance a basis for the governing equation can be established. It is subsequently apparent that there are several extraneous geometric terms contained within the governing equation. These were systematically eliminated and the equation was put into an explicit form for wall shear stress. This is shown in the proceeding lines.

\[
\rho \frac{\pi}{4} D^2 z \frac{d\bar{u}}{dt} = \pi D z \tau_w + \rho \frac{\pi}{4} D^2 z \frac{dp}{dz} \\
\rho \frac{\pi}{4} D^2 z \frac{d\bar{u}}{dt} = \pi D z \tau_w + \frac{\pi}{4} D^2 z \frac{dp}{dz} \rightarrow \rho \frac{D \frac{d\bar{u}}{dt}}{4} = \tau_w + \frac{D \frac{dp}{dz}}{4} \\
\tau_w = \frac{D}{4} \left( \rho \frac{d\bar{u}}{dt} + \frac{dp}{dz} \right) \]
5.2 Density of Fluid Mixtures

Of course, the density variable isn’t quite as simple as just looking it up in a table. With the presence of gas emboli in an aqueous glycerin solution there are multiple fluids to be considered. Fortunately, for non-polymer low molecular weight fluids it is rather simple to calculate density of mixtures. A table of the densities of each the fluids involved in this work at standard temperature and pressure (1 atm and 20˚C) is displayed in Table 6 [9].

<table>
<thead>
<tr>
<th>Fluid</th>
<th>Glycerin</th>
<th>Water</th>
<th>Air</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density</td>
<td>1260 kg/m³</td>
<td>998 kg/m³</td>
<td>1.2 kg/m³</td>
</tr>
</tbody>
</table>

The density of each constituent of the mixture is weighted by the volumetric percentage, a value known as the void fraction and denoted by \( \alpha_k \) [24]. This value is used as a coefficient for the density in the summation to calculate apparent density (the density expressed by the mixture). It should be noted that this approach to calculating density assumes that the mixture is reasonably well mixed and does not have any significant discontinuities. This point will be subsequently demonstrated in the Continuum Body/Flow Regimes section. This relationship is shown in the equations form in the subsequent lines [24].

\[
\alpha_k = \frac{V_k}{V} \quad \text{and} \quad \rho = \sum_{k=1}^{N} \alpha_k \rho_k
\]

5.3 Assumptions

In a mathematical approach, such as the one used in this thesis, it is important to not only make assumptions in order to make equations solvable but also to further explain and validate those assumptions. The suppositions of axisymmetric flow and neglected gravity were both very valid for this setup. Even if the test tube were not level the influence of gravity would be
completely overwhelmed by the pressure and momentum of the flow stream [1] [7]. Also, even with disparate turbulence the flow of the fluid will remain overwhelmingly longitudinal. However, some other inherent assumptions require more in-depth analysis.

5.3.1 Continuum Body/Flow Regimes

It was taken for granted that this body was a well mixed continuum by not taking interfaces or discontinuities into consideration. This supposition was based on statements within the relevant literature. The flow created by the circulatory simulator remained well within the bubbly regime throughout testing. Flow regimes are important for characterizing the properties of a given two phase stream. A graphic displaying prominent examples of this concept is shown in Figure 13 [63]. Although some publications introduce transitional regimes as well the ones illustrated below (liquid, bubbly, slug, annular, droplet, and gas) are almost universally considered the major breakdowns of multiphase flow.

![Flow regime examples](image)
Bubbly flow is typified by bubbles dispersed in the liquid. When shear forces are dominant the bubbles tend to disperse uniformly throughout the lumen [63]. This best describes the properties of asymptomatic gas emboli [64]. In fact, the transition between flow regimes is largely governed by the void fraction (defined earlier as $\alpha_k$). With a maximum void fraction of 11.5% (15 mL/min maximum gas infusion and 130 mL/min minimum blood simulant flow) the presence of the fluid is not nearly enough to create significant coalescence or occlusion of the lumen [64] [65]. In the actual experimental testing this bubbly flow/continuum body was visually confirmed. The bubbles produced by the test apparatus were small and evenly distributed longitudinally with a very slight bias toward the top of the tube. Also, significant jumps were not seen in the wall shear stress data.

5.3.2 Fully Developed Flow

One of the other key assumptions was that the flow was fully developed. In order to numerically confirm this assumption is will be necessary to complete a multistep process. The length of the entrance region is closely tied with the Reynolds number of the flow. However, it is first necessary to calculate the viscosity and density of the blood simulant approaching the test section. Although there are several methods for calculating multiphase mixture viscosity the equation shown below was a consistent presence in the relevant literature [66] [67] [68].

$$\ln \mu = \sum_{k=1}^{N} \alpha_k \ln \mu_k$$

Additionally, a table displaying the viscosities of the test fluids at STP conditions is given in Table 7 [9].

<table>
<thead>
<tr>
<th>Fluid</th>
<th>Glycerin</th>
<th>Water</th>
<th>Air</th>
</tr>
</thead>
<tbody>
<tr>
<td>Viscosity</td>
<td>1.2 Pa-S</td>
<td>8.94 * 10^{-4} Pa-S</td>
<td>18.12 * 10^{-6} Pa-S</td>
</tr>
</tbody>
</table>
The material properties found in Table 6 and Table 7 were used to calculate the density and viscosity for each of the flow rate pairings (blood simulant and embolism) according to the equations outlined above. The results, produced in Microsoft Excel, are displayed in Table 8 and Table 9.

**Table 8: Fluid density for various flow pairings**

<table>
<thead>
<tr>
<th>Gas Infusion Rate (mL/min)</th>
<th>0</th>
<th>5</th>
<th>10</th>
<th>15</th>
<th>Density (kg/m³)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time Average Blood</td>
<td>130</td>
<td>1102.8</td>
<td>1062</td>
<td>1024.114</td>
<td>988.8414</td>
</tr>
<tr>
<td>Simulant Flow (mL/min)</td>
<td>280</td>
<td>1083.474</td>
<td>1064.814</td>
<td>1046.786</td>
<td></td>
</tr>
</tbody>
</table>

**Table 9: Fluid viscosity for various flow pairings**

<table>
<thead>
<tr>
<th>Gas Infusion Rate (mL/min)</th>
<th>0</th>
<th>5</th>
<th>10</th>
<th>15</th>
<th>Viscosity (Pa-s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time Average Blood</td>
<td>130</td>
<td>0.01594</td>
<td>0.0124</td>
<td>0.009821</td>
<td>0.007905</td>
</tr>
<tr>
<td>Simulant Flow (mL/min)</td>
<td>280</td>
<td>0.01594</td>
<td>0.014152</td>
<td>0.012617</td>
<td>0.011292</td>
</tr>
</tbody>
</table>

These data were then used to calculate the Reynolds number of the time averaged flow leading up to the test section. The equation defining the unitless Reynolds number (Re) is shown below and the spreadsheet results can be seen in Table 10 [9].

\[
Re = \frac{\rho UD}{\mu}
\]

**Table 10: Reynolds number for various flow pairings**

<table>
<thead>
<tr>
<th>Gas Infusion Rate (mL/min)</th>
<th>0</th>
<th>5</th>
<th>10</th>
<th>15</th>
<th>Re</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time Average Blood</td>
<td>130</td>
<td>30.05732</td>
<td>38.638227</td>
<td>48.78597</td>
<td>60.61628</td>
</tr>
<tr>
<td>Simulant Flow (mL/min)</td>
<td>280</td>
<td>64.73884</td>
<td>72.916874</td>
<td>81.79184</td>
<td>91.39054</td>
</tr>
</tbody>
</table>

The Reynolds numbers presented were much lower than any threshold for onset of turbulent flow. Thus, the conclusion was made to use the laminar form of the pipe entrance
length equation illustrated below, where $L_e$ is entrance region length [69]. The results may subsequently be found in Table 11.

$$Turbulent: L_e = D * 4.4 * Re^{\frac{1}{6}} \quad Laminar: L_e = 0.06 * D * Re$$

Table 11: Entrance length for various flow pairings

<table>
<thead>
<tr>
<th>Gas Infusion Rate (mL/min)</th>
<th>0</th>
<th>5</th>
<th>10</th>
<th>15</th>
<th>Entrance Length (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time Average Blood</td>
<td>130</td>
<td>0.011452</td>
<td>0.01472116</td>
<td>0.018587</td>
<td>0.023095</td>
</tr>
<tr>
<td>Simulant Flow (mL/min)</td>
<td>280</td>
<td>0.024665</td>
<td>0.02778133</td>
<td>0.031163</td>
<td>0.03482</td>
</tr>
</tbody>
</table>

From Table 11 it can be concluded that for the test setup an approximate maximum length of the entrance region would be 3.5 cm (1.378”). The length of uniform tube leading up to the test section was measured to be 6” in length. Thus, it is safe to assume that the pulsatile flow in the test section is fully developed. Keep in mind, however, that this is an approximation on time averaged flow rates and transient spikes might increase it somewhat. However, it is doubtful that a transient wave would increase Reynolds number that dramatically.

5.4 Statistical Considerations

Some statistical considerations were also included in the analysis. As previously discussed the relevant data was to be averaged from the duplicate data runs in order to gain more reliable results. Additionally, it was desired that the standard error be calculated for each grouping of data. Standard error is very useful for this particular because it takes sample size into consideration. The square root of the number of samples is inversely proportional to the standard error. This would provide a visual cue on charts as to the spread experienced within each test setting. It should be duly noted that standard error does not necessarily reflect an error in calculation or measurement but is closely related to the standard deviation of the data.
\[ \bar{x} = \frac{1}{N} \sum_{k=1}^{N} x_k \quad \rightarrow \quad s = \sqrt{\frac{\sum_{k=1}^{N} (x_i - \bar{x})^2}{N - 1}} \quad \rightarrow \quad SE = \frac{s}{\sqrt{N}} \]

5.5 MATLAB Program

A MATLAB program was written to process the large amounts of data generated by the testing. A copy of the program code is provided in Appendix A for reference. The first step taken by the program was to define the preset constants relating to the data acquisition process. These values were consistent for every test and thus did not need to be altered by the user.

Subsequently, the code utilized a pair of inputs (embolism and blood analog volumetric flow) to calculate the average density of the bubbly test mixture. In order to go further with the data analysis it was then necessary to import the data from one of the ASCII files generated by the SignalExpress software. In order to more easily process the raw experimental data it was split from its matrix form into a group of separate vectors (volumetric flow, pressure drop, and time).

The volumetric flow was subsequently converted into the far more useful mean flow velocity variable. Unfortunately, early efforts with the program produced unreliable results due to the noise present in this velocity signal. While this same noise was present in the other data it was magnified when the time derivative of the velocity was calculated. In order to alleviate this instability a MATLAB digital filter algorithm (a Savitzky-Golay filter, to be exact) was utilized. This type of filter utilizes a local polynomial regression to smooth the data curve. This method is advantageous because it tends to retain wave features like local maxima and minima. The cutoff frequency was set to 3 Hz, which didn’t result in any aliasing or unwanted effects.
Figure 14: Sample wall shear stress waveform

Subsequent to this filtering the approximate derivative of the velocity data was taken and fed into a for-loop. In the for-loop wall shear stress was calculated using the data vectors and user inputs on the test setup. This string of time dependent wall shear stress values was then averaged to obtain the mean wall shear stress for that experimental setting. The wall shear stress was then exported as a text file of the same name as the raw data. Finally, graphs of the wall shear stress, mean flow velocity, and pressure drop were created to verify correct operation. A sample printout is shown in Figure 14. Though this program took a bit of time to develop and troubleshoot it was of great assistance in reliably analyzing the test data.
Chapter 6. Results

After the various mean wall shear stresses were calculated by the aforementioned MATLAB program they were placed in a spreadsheet and graphed. The relationships between mean wall shear stress and embolism presence as well as complications with heart rate, blood flow, etc. were analyzed. The results and their discussion are provided below.

6.1 Static Flow

The average mean wall shear stress and standard error were calculated for the 4 samples taken at each test setting. The end results of the static testing are collected below in Table 12 for visual reference. It can be almost automatically seen that embolism rate has an inverse relationship with mean wall shear stress. This trend seems to be true in the hydrostatic experiments no matter what the peripheral setting. It can also be noted that increased fluid flow rate significantly increased wall shear stress, which is not particularly surprising (that drag force would increase with flow). It appears at first glance that standard error increases with increased embolism presence. This value climbs steadily with embolism rate for coronary and carotid flow levels but starts at .184 dynes/cm^2 and .192 dynes/cm^2, respectively. Indeed, the level of precision demonstrated in Table 12 was quite surprising to the author.

Table 12: Synopsis of static mean wall shear stress results

<table>
<thead>
<tr>
<th>Flow Rate</th>
<th>Embolism Rate</th>
<th>Sample Recording</th>
<th>Mean Wall Shear Stress (dynes/cm^2)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coronary Artery (130 mL/min)</td>
<td>0 mL</td>
<td>1  2  3  4  AVERAGE  SE</td>
<td></td>
</tr>
<tr>
<td></td>
<td>5 mL</td>
<td>5.22 5.58 5.78 6.10 5.67 0.1843</td>
<td></td>
</tr>
<tr>
<td></td>
<td>10 mL</td>
<td>3.61 4.16 3.55 4.52 3.96 0.2317</td>
<td></td>
</tr>
<tr>
<td></td>
<td>15 mL</td>
<td>3.02 3.79 2.89 3.54 3.31 0.2129</td>
<td></td>
</tr>
<tr>
<td>Carotid Artery (280 mL/min)</td>
<td>0 mL</td>
<td>8.13 8.95 8.78 8.18 8.51 0.2081</td>
<td></td>
</tr>
<tr>
<td></td>
<td>5 mL</td>
<td>8.23 7.64 8.18 7.27 7.83 0.2295</td>
<td></td>
</tr>
<tr>
<td></td>
<td>10 mL</td>
<td>6.84 7.34 7.44 6.46 7.02 0.2282</td>
<td></td>
</tr>
<tr>
<td></td>
<td>15 mL</td>
<td>5.99 6.75 6.57 5.69 6.25 0.2473</td>
<td></td>
</tr>
</tbody>
</table>
However, a table alone does not properly reveal the lessons that a body of data has to tell. This information was also graphed to further understand the underlying currents. As seen in Figure 15 this same downward trend in MWSS with increasing embolism rate is apparent. However, not simply the downward slope but also the details of this descent should be noted. Apart from beginning at different y-intercepts the two data sets have very similar slopes (-.1518 and -.1584). The $R^2$ values are particularly interesting as well. A rating of 1 in this category would mean that a numerical set is a perfect line. The carotid and coronary flows demonstrated $R^2$ values of .9988 and .9958, respectively. This evidence would seem to support that the relationship between mean wall shear stress and embolism rate is linear for hydrostatic flows.

![Figure 15: Static mean wall shear stress vs. embolism rate](image)

Although the static coronary flow data is completely beneath the atherosclerotic onset limit this insight should be taken with a grain of salt. Static flow does not occur physiologically...
in the body. Therefore this data would not be completely relevant when considering atherosclerotic risk factors. It was included, however, to provide a valuable baseline dataset to see the effects of heart rate on the severity of gas emboli.

Standard error was also an important parameter in testing. A relationship between standard error (a function of standard deviation) and other parameters would indicate a contribution towards instability by that characteristic. Thus, standard error for the static test was graphed versus embolism rate in Figure 16. A slight upturn in MWSS standard error can be seen, but fortunately nothing particularly definitive can be gleaned from the data.

![MWSS Standard Error vs. Embolism Rate](image)

**Figure 16: Static mean wall shear stress standard error vs. embolism rate**

6.2 Pulsatile Flow

While it can be reasonably assumed that the pressure drop term was dominant for static tests it was also the key driving factor behind the wall shear stress for pulsatile experiments. Typically, this term had approximately twice the influence of the acceleration portion. The same spreadsheet analysis process conducted for the static tests was subsequently repeated for the pulsatile data. The average and standard error mean wall shear stress for each test setting was calculated and tabulated. Of course, with the inclusion of the resting and elevated heart rates
there were more data to consider with the pulsatile setting. Some of the previously discussed
trends were recurrent in this set of results while several different phenomena were also observed.
The results of the pulsatile analysis are shown in Table 13.

**Table 13: Pulsatile mean wall shear stress results**

<table>
<thead>
<tr>
<th>Flow Rate</th>
<th>Pulse Rate</th>
<th>Embolism Rate</th>
<th>Sample Recording</th>
<th>Mean Wall Shear Stress (dynes/cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coronary Artery</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(130 mL/min)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Resting (70 bpm)</td>
<td>0 mL</td>
<td>7.99</td>
<td>8.21</td>
<td>7.16</td>
</tr>
<tr>
<td></td>
<td>5 mL</td>
<td>6.17</td>
<td>4.69</td>
<td>5.63</td>
</tr>
<tr>
<td></td>
<td>10 mL</td>
<td>4.16</td>
<td>5.58</td>
<td>4.36</td>
</tr>
<tr>
<td></td>
<td>15 mL</td>
<td>4.22</td>
<td>4.45</td>
<td>3.11</td>
</tr>
<tr>
<td>Elevated (135 bpm)</td>
<td>0 mL</td>
<td>11.50</td>
<td>10.60</td>
<td>11.42</td>
</tr>
<tr>
<td></td>
<td>5 mL</td>
<td>10.88</td>
<td>9.18</td>
<td>10.04</td>
</tr>
<tr>
<td></td>
<td>10 mL</td>
<td>9.86</td>
<td>8.29</td>
<td>9.67</td>
</tr>
<tr>
<td></td>
<td>15 mL</td>
<td>8.62</td>
<td>7.73</td>
<td>9.51</td>
</tr>
<tr>
<td>Carotid Artery</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(280 mL/min)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Resting (70 bpm)</td>
<td>0 mL</td>
<td>13.98</td>
<td>13.08</td>
<td>12.31</td>
</tr>
<tr>
<td></td>
<td>5 mL</td>
<td>12.66</td>
<td>11.19</td>
<td>12.77</td>
</tr>
<tr>
<td></td>
<td>10 mL</td>
<td>10.13</td>
<td>10.26</td>
<td>11.63</td>
</tr>
<tr>
<td></td>
<td>15 mL</td>
<td>10.45</td>
<td>9.98</td>
<td>9.04</td>
</tr>
<tr>
<td>Elevated (135 bpm)</td>
<td>0 mL</td>
<td>15.16</td>
<td>16.41</td>
<td>16.44</td>
</tr>
<tr>
<td></td>
<td>5 mL</td>
<td>14.23</td>
<td>14.46</td>
<td>15.79</td>
</tr>
<tr>
<td></td>
<td>10 mL</td>
<td>14.02</td>
<td>15.08</td>
<td>14.23</td>
</tr>
<tr>
<td></td>
<td>15 mL</td>
<td>13.86</td>
<td>12.67</td>
<td>14.15</td>
</tr>
</tbody>
</table>

For the pulsatile results this persistent decrease in mean wall shear stress with embolism rate is once again present. The standard error of mean wall shear stress, on the other hand, increased consistently with embolism rate. Also, both MWSS stress and standard error increased as a function of increased fluid flow rate. Higher heart rates were seen to generally increase both mean wall shear stress and standard error as well. It should be noted that the standard error levels were significantly higher for pulsatile results than for static, regardless of peripheral test settings.
In order to provide a more visual representation of the test data a graph was created of mean wall shear stress vs. embolism rate for the multiple experimental parameters. It can be seen in Figure 17 that although the MWSS does decrease for all of the configurations with embolism rate they do not all do so linearly. The descent seems to level off in the 10-15 mL/min region. This, of course, differs from the static dataset.

The order of the flow categories from largest to lowest mean wall shear stress is elevated heart rate and carotid flow, resting heart rate and carotid flow, elevated heart rate and coronary flow, and finally resting heart rate and coronary flow. It would appear that the coronary cardiovascular flow setting consistently produces lower mean wall shear stresses. The same could, of course, be said for the resting heart rate. The above observation may also indicate that fluid flow rate has a greater effect on mean wall shear stress than does heart rate.

![MWSS vs. Embolism Rate (Pulsatile Flow)](image-url)

Figure 17: Pulsatile mean wall shear stress vs. embolism rate
It should also be noted that the resting heart rate with coronary flow was the only series to dip below 6 dynes/cm². As previously stated, this shear stress regime is closely associated with the onset of atherosclerosis. Such a result, of course, is cause to take note.

In addition to the average shear stress the standard error of the shear stress was also analyzed. A plot of the standard error versus embolus rate is shown in Figure 18. Much like Figure 16 this graph displays a slight increase with air infusion rate, but nothing consistent or significant. However, what is notable is the change in magnitude between standard error in the pulsatile tests and the static tests. The maximum standard error observed for the hydrostatic tests was .2473 dynes/cm² while it was .3885 dynes/cm² for pulsatile experiments.

**Figure 18: Pulsatile mean wall shear stress standard error vs. embolism rate**

In order to gain a slightly different perspective on the data they were organized along lines of heart rate as opposed to embolism injection rate. Often a single group of results will
yield far more insight if only examined from a different angle. For these purposes static tests were simply considered to be 0 bpm tests. The fully tabulated data are displayed in Table 14.

<table>
<thead>
<tr>
<th>Flow Rate (mL/min)</th>
<th>Coronary</th>
<th>Carotid</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>5.7</td>
<td>70</td>
</tr>
<tr>
<td>10</td>
<td>4.0</td>
<td>5.6</td>
</tr>
<tr>
<td>15</td>
<td>3.3</td>
<td>4.7</td>
</tr>
<tr>
<td></td>
<td>8.5</td>
<td>3.7</td>
</tr>
<tr>
<td></td>
<td>7.8</td>
<td>13.2</td>
</tr>
<tr>
<td></td>
<td>7.0</td>
<td>12.1</td>
</tr>
<tr>
<td></td>
<td>6.3</td>
<td>10.8</td>
</tr>
<tr>
<td></td>
<td></td>
<td>9.6</td>
</tr>
</tbody>
</table>

Table 14: Experimental results rearranged according to heart rate

What is particularly striking about this table of results is how the individual values fit so well together within this new organization. All of the mean wall shear stresses increase in a consistent manner with heart rate. This definitely lends support to the idea that heart rate may have an adverse effect on the severity of gas emboli.

A graph was constructed to illustrate this concept and is depicted in Figure 19. It should also be noted that the four largest magnitude data series are all carotid flow regimes. It would appear that the coronary and carotid artery data differs even more significantly when viewed in this light.
Figure 19: Mean Wall Shear Stress vs. Heart Rate
Chapter 7. Conclusions

This study was undertaken to investigate the possible effects of chronic asymptomatic gas emboli on the formation of atherosclerosis by measuring the effect of gas emboli on mean wall shear stress under a variety of cardiovascular circumstances, namely varying fluid flow rates and pulse frequencies. This subsequently uncovered some very helpful and interesting conclusions.

First, several general correlations were established between relevant fluid mechanical factors and the level of mean wall shear stress. It was established that mean wall shear stress had an inversely proportional relationship with gas bubble infusion rate and a directly proportional relationship with circulatory volume and heart rate. These points are also illustrated below in Table 15.

<table>
<thead>
<tr>
<th>Mean wall shear stress decreases as…</th>
<th>Embolism rate increases</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Volumetric blood flow decreases</td>
</tr>
<tr>
<td></td>
<td>Heart rate decreases</td>
</tr>
</tbody>
</table>

But what do these observations say about cardiovascular health? This indicates that increased volume of asymptomatic gas emboli in the bloodstream would most likely decrease local arterial mean wall shear stress, particularly in regions with low volumetric flow or diminished pulse. There is theoretical work to support the idea that reduced wall shear stress is a significant contributor to the formation of atherosclerosis.

However, the author was unable to find any significant evidence for or against a definitive causal link. Under in-vitro testing one situational instance was found (in the coronary artery at a resting heart rate with gas embolus rate in excess of 5 mL/min) of a mean wall shear stress low enough to induce atherosclerosis. But this evidence is certainly not sufficient to
demonstrate a real-world medical issue with any certainty. This also does not, by any means, represent the refutation of a causal relationship either.

Rather, this study laid the ground work for future investigations into the effects of gas emboli on endothelial function. The research was able to clearly establish a positive link between bubbles in the blood stream and decreased mean wall shear stress. It was also able to identify peripheral circumstances that would exacerbate this fluid mechanical response.

7.1 Future Work

The first step of expanding upon this thesis would be to improve the flow simulator for greater accuracy. Utilizing more advanced techniques such as laser velocimetry or video microscopy in addition to the current sensors would allow for increased accuracy and a degree of redundancy. Also, using actual blood would help to better approximate the hemodynamic environment.

However, in order to really move significantly the current status of this research it would be necessary to perform similar tests on actual biological specimens. By exposing a culture of endothelial cells to conditions similar to those used in this research a more direct link between gas emboli and atherosclerotic onset could be established, or at least more so than an extensive literature review.

Further into the future, though, it would most likely become necessary to perform a survey of clinical cases or closely monitored patient database. As close as one can come in a laboratory to replicating the outside world it certainly can’t beat the real thing. Fortunately, organizations like NASA and the Diver’s Alert Network are several decades ahead in this regard. Although they tend to focus their efforts on more catastrophic events there is no reason they couldn’t monitor atherosclerosis rates as well.
References


Appendix A: Wall Shear Stress MATLAB Code
clear all

%glossary of variables
%dt: sampling period (s)
%dp: pressure drop (Pa)
%dz: tube length (m)
%D: tube diameter (m)
%f: sampling frequency (hz)
%Q: volumetric flow rate
%Qb: time average volumetric blood analog flow (mL/min)
%Qg: time average volumetric gas flow (mL/min)
%rho: density of test fluid (kg/m^3)
%rhob: density of blood analog (kg/m^3)
%rhog: density of air (kg/m^3)
%t: time (s)
%tauw: wall shear stress (dynes/cm^2)
%tauwm: mean wall shear stress (dynes/cm^2)
%u: mean flow velocity (m/s)
%uf: filtered mean flow velocity (m/s)

%defined preset constants DO NOT TOUCH
f=30;
dt=1/f;
dz=.1524;
%6" = 152.4 mm = .1524 m
D=.00635;
%1/4" = 6.35 mm = .00635 m
rhob=1102.8;
rhog=1.2;

% !!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!
% Test dependent variables MUST FILL OUT
Qb=130;
Qg=0;
% !!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!

%solving for test fluid density based on experimental setting (user input)
rho=Qg/(Qg+Qb)*rhog+Qb/(Qg+Qb)*rhob;

%import experimental data
%naming syntax: Flow Type/Flow Volume/Pulse Rate/Embolus Volume/Run #.asc
data = importdata('S/Co/N/0/1.asc');

%define length of data to set for-loop limit
loop=length(data);
%open blank arrays
%note: matrix component syntax: variable(row,column)
t=zeros(row,1);
dp=zeros(row,1);
Q=zeros(row,1);
tauw=zeros(row,1);

%fill blank arrays with data
%data columns:
%1 - time
%2 - pressure drop
%3 - mean fluid velocity
for i=1:1:loop
    t(i)=data(i,1);
    dp(i)=data(i,2);
    Q(i)=data(i,3);
end

%ascertain mean flow velocity from volumetric flow rate
u=Q/(pi/4*D^2);

%smooth velocity data to reduce issues with differential operation
uf=sgolayfilt(u,3,41);
%calculate finite difference/approximate derivative of mean velocity
ud=diff(uf);

%calculate instantaneous wall shear stress in dynes/cm^2
%Tw=D/4*(rho*du/dt+dp/dz)
for i=1:1:loop-1
    tauw(i)=D/4*(rho*ud(i)/dt+dp(i)/dz)*10;
end

%calculate mean wall shear stress
tauwm=mean(tauw);

%export wall shear stress data to text file named after source data ASCII
%file
export(tauw,'file',S/Co/N/0/1)
%!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!
%CHANGE NAME TO MATCH TEST SETUP
%!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!
figure (1)
plot(t,dp), grid on
title('Pressure Drop vs. Time')
xlabel('Time (s)')
ylabel(’Pressure Drop (Pa)’)

figure (2)
plot(t,u), grid on
title(’Mean Flow Velocity vs. Time’)
xlabel(’Time (s)’)
ylabel(’Mean Flow Velocity (m/s)’)

figure (3)
plot(t(1:loop-1),tauw), grid on
title(’Wall Shear Stress vs. Time’)
xlabel(’Time (s)’)
ylabel(’Wall Shear Stress (dynes/cm^2)’)