Experimental Evaluation of Streamline Patterns and Separated Flows in a Series of Branching Vessels with Implications for Atherosclerosis and Thrombosis

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SUMMARY  Flow conditions in four models representing the aortic bifurcation, iliac bifurcation, and a renal artery branch were investigated at volumetric flow rates corresponding to Reynolds numbers from 1000 to 4000 over the complete range of flow division between daughter vessels. Qualitative flow streamline patterns and quantitative definition of those flow conditions leading to disturbed flow (flow separation) were determined primarily at steady flow with a limited set of pulsatile experiments. Under conditions of no flow separation, common characteristic streamline patterns not parallel to the center lines of parent or daughter tubes were found for all models. These effects were accentuated with increasing Reynolds number. Flow separation was inducible through alteration of flow division between daughter vessels or by an increase in flow rate. Each of the four models had distinct combinations of flow division ratio and flow rate which gave: (1) no flow separation, (2) flow separation at the outside of the right daughter tube, and (3) flow separation at the outside of the left daughter tube. Models representing the renal artery also had regions of simultaneous left- and right-hand separation on the outside of their daughter tubes. The separated flows observed here displayed streamlines forming an open vortex with flows entering and leaving. These regions, which occur only at distinct combinations of flow rate and flow division, may be key centers where platelet aggregates may form, release constituents, and cause vessel injury.

THE QUALITY of flow in the vascular system has been related to the sites of predilection for both atherosclerosis and thrombosis. Conditions of high and low shear rate, as well as turbulence and disturbed flow, have been implicated. Flow separation, which is characterized by reverse fluid movement adjacent to the vessel's surface and a recirculating vortex region, can occur as a result of changes in cross-sectional area along a conduit. A transition to fluid motions within the recirculating region (vortex) which are more random, displaying fluctuating bursts, also is possible at higher flow rates. These flow conditions are referred to as disturbed flows. Sites of branching display such cross-sectional area variation where, during pulsatile flow both in vitro and in vivo, vortex motion and random fluctuation in local velocity have been reported. Recirculating flows of this kind have been further demonstrated while also indicating the importance of the relative flow division at branches in the generation of such conditions.
The cause of vessel injury has become a key question in a "response to injury hypothesis" for atherogenesis. Fluid mechanical factors as well as biochemical ones have been implicated. Positions adjacent to disturbed flow regions may be doubled ill-fated in terms of potential endothelial damage: first, due to high shear stress from disturbed flows and, second, due to biochemical damage, as a result of substances released by platelets, since in vitro evidence for the formation of platelet aggregates within vortices now exists.

Of the shapes prone to flow separation, the most extensively investigated has been the symmetrical branch; studies have been made on models using computer techniques and in vitro flow simulation methods and in vitro flow simulation methods. These investigations have dealt with definition of velocity fields, wall shear rates, and flow separation over a wide flow rate range and with some geometrical variance. Conclusions with respect to disturbed flow (flow separation) display some disagreement; the studies utilizing two-dimensional models report flow separation at considerably lower flow rates than those reporting upon three-dimensional tubular models. The flow division ratio between daughter tubes was not treated as an experimental variable in these studies.

This investigation of flow in vitro presents qualitative flow streamline patterns and quantitative definition of those flow conditions leading to disturbed flow. The study was done primarily under steady flow conditions with a limited set of pulsatile experiments. Four three-dimensional models were selected as representative of geometries important in arterial disease. Volumetric flow rate (presented as Reynolds number—a dimensionless quantity proportional to flow rate) and flow division ratio between daughter vessels were selected as the key experimental variables, since substantial variation in these is possible under normal physiological conditions.

Methods

Experimental Design

Techniques for defining fluid motions are numerous. They may be used for both quantitative evaluation of velocities and qualitative determination of streamlines. In the latter category, dye injection, smoke tracers, hydrogen bubbles, and polymer tracer particles have proven useful for a variety of applications. In the present study, a neutrally buoyant tracer-particle system similar to others formulated by our group was used with ciné photography for streamline observations. Rigid transparent models were constructed for this study, thus omitting the effects of vessel elasticity. The implications of this have been examined in detail elsewhere, with the conclusion that the class of fluid mechanical phenomena examined here would not be appreciably altered by the degree of elasticity expected.

As a first step in model preparation, angiograms of the abdominal aorta and its major branches (celiac, renal, and superior mesenteric arteries), the aortic bifurcation and the iliac bifurcation were studied. Measurements of vessel diameter, take-off angle, and radius of curvature of the transitions between parent and daughter tubes were made and comparisons done with data from the literature. Although a certain degree of variability was observed, common geometrical characteristics were found, and these were incorporated into the four models described in more detail below. The rigid nature of the models did not allow for changing either the parent vessel curvature or the curvature within the transition from parent to daughter vessels as experimental variables. Figure 1 shows drawings for the models with dimensions given in Table 1. The entrance tube for all the models was for convenience made 2.2 cm in diameter, thus requiring scaling from anatomical information for fabrication, since not all parent vessels were 2.2 cm in diameter and some vessel size information was available only as diameter ratios. Since we report the Reynolds number for our observations, direct comparison is possible with other data at equal Reynolds numbers due to geometrical and dynamical similarity.

Model 1

This is a model of a symmetrical branch typified by the aortic bifurcation. Anatomically, this bifurcation has an angle between its two branches in the range of 60°-90°. It has been reported that the descending aorta undergoes a 40% reduction in cross-sectional area over its length. Experimental measurements have led to an exponential dependence for this area as a function of distance measured from the proximal site. The equation describing this is:

$$A_m = A_0 e^{-\beta x/R_0}$$

where $x$ is the distance coordinate, $A_0$ and $R_0$ are the mean area and radius at the proximal site, and $\beta$ is the taper factor. Such a taper was incorporated into this model. The parent and daughter tube diameters were chosen to provide equal mean velocities in parent and daughter tubes with conditions of equal flow division. An area ratio of 1.0 resulted (total outflow to total inflow area) and is in the physiological range.

Model 2

This model represents the asymmetrical iliac bifurcation. In some cases, the external iliac artery is a straight continuation of the common iliac. The take-off angles selected were 20° and 45°. The daughter tube diameter ratio then was selected with the aid of a mathematical relationship stating that the product of daughter tube flow area and sines of the take-off angle is equivalent for both branches. This equation has been shown to be applicable in a number of physiological cases. An inflow-to-total
The outflow area ratio of 1.0 was then selected to give nonaccelerating flow when the division of flow between daughter tubes is proportional to their cross-sectional areas, completing the geometrical definition of this model. This area ratio is in the physiological range.

Models 3 and 4

The renal arteries form two main branches emanating from nearby positions along the abdominal aorta. The take-off angle varies between 45° and 90° for these vessels. Two models each having a single branch at the limits of this span were selected. Thus, these models are simplified versions constructed principally to study take-off angle. The tube diameters for these models were chosen to comply with anatomical measurements and our own observations. These models also represent other branches of the aorta, e.g., celiac and superior mesenteric arteries.

The models were machined from blocks of Perplex. Powered drilling operations were used, as well as manual forming, to shape the transition sections. Clear transparent specimens resulted after polishing. The accuracy of the machining operations was checked by comparing projected cross-sections of the models from cine films with the design drawings. Deviations of less than 0.05 cm were found. Rigid plastic tubes with lengths equal to 15-20 diameters and having diameters equal to those of the models were attached.

Experiments were designed to examine the quality of steady flow over a wide range of parent tube Reynolds number \( DVp/\mu \), where \( D \) is tube diameter, \( V \) mean velocity, \( \rho \) fluid density, and \( \mu \) fluid viscosity, which is proportional to flow rate. A range of 1000–4000 was selected to cover a variety of physiological flows in both laminar and turbulent regimes. Although a value of 4000 may seem high when compared to resting values in the thoracic and abdominal aortic regions of 1500 to 2500, upon moderate exercise, flow may be elevated to a Reynolds number of 4000. The other key experimental variable was flow division between daughter tubes which was made both measurable and controllable through a system described later. The entire range, from total to zero flow in each daughter tube, was examined, since this was feasible.

A limited investigation aimed at defining differences between pulsatile and steady flows also was undertaken. Pulsatile flow in the aorta can be approximated by a Fourier series. The first harmonic, a sine wave having the frequency of the heart beat, gives the largest contribution. Thus, in this study, physiological pulsatile flow was approximated by adding the flows from a steady flow pump and an oscillating piston which yielded a volumetric flow having the form of a displaced sine wave.

The three parameters necessary to characterize single-harmonic pulsatile flow are: the mean flow

<table>
<thead>
<tr>
<th>Model</th>
<th>Parent tube D</th>
<th>Left branch</th>
<th>Right branch</th>
<th>Area ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(cm)</td>
<td>dL (cm)</td>
<td>aL (0)</td>
<td>AL (cm)</td>
</tr>
<tr>
<td>1</td>
<td>2.2*</td>
<td>1.27</td>
<td>37.5</td>
<td>1.8</td>
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<td>2</td>
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<td>1.27</td>
<td>45</td>
<td>2.2</td>
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<tr>
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<td>45</td>
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<tr>
<td>4</td>
<td>2.2</td>
<td>0.90</td>
<td>90</td>
<td>1.1</td>
</tr>
</tbody>
</table>

Definitions for \( D, dL, \) and \( dR \) may be found in Figure 1; \( aL \) and \( aR \) are left and right take-off angles measured from the parent tube center line to the daughter tube center line; \( \lambda_L \) and \( \lambda_R \) are left and right radii of curvature of the transitions between parent and daughter tubes.

* This model’s parent tube is tapered; the tapered length, \( l \), is 6.6 cm, the upstream diameter is \( D \), and the diameter at the junction, \( D_0 \), is 1.8 cm (see Fig. 1).
† The area ratio for this model was computed from \((dL^2 + dR^2)/D_0^2\).
Reynolds number, unsteadiness parameter ($\alpha$), and the flow fluctuation ratio. A range of 1200–1900 was chosen for the Reynolds number. The unsteadiness parameter is defined as $\alpha = R(2\pi f p/\mu)^{1/2}$, where $R$ is the tube radius, $f$ the pulse frequency, $p$ the fluid density and $\mu$ the fluid viscosity. For the human aorta and large arteries, the range of $\alpha$ is 6–12 at rest with 18 possible during moderate exercise. Frequencies from 0.2 to 0.6 sec$^{-1}$ were chosen corresponding to $\alpha$ values from 10 to 18. The flow fluctuation ratio is defined as the quotient of the difference between maximum and minimum flow rate values with the mean. To create a base condition (in the parent vessel) in which no reverse flow was present, an additional constraint upon the flow fluctuation ratio was necessary. This aided the experimental recognition of separated flow and resulted in a flow fluctuation ratio range of 0.8–1.2.

Experimental Equipment and Procedures

The flow circuit used in this study has been modified from a system used for previous work (Fig. 2). Flow rates were monitored with two rotameters calibrated to have an accuracy of ±1% of maximum reading. They were placed downstream from the pump and downstream from one of the model’s daughter tubes (the larger daughter tube, usually). Flow resistance in each daughter tube was controlled by screw-clamps on sections of plastic tubing, providing for smooth continuous control of flow rate. In the pulsatile case, valve settings were made for the steady flow conditions having the same total mean flow and flow division conditions required.

Pulsatile flow was obtained by means of a piston acting on the fluid within a mixing chamber upstream of the test section. The piston was powered by a variable speed DC motor through a scotch-yoke mechanism (Fig. 2). By varying the main pump total flow rate and frequency and amplitude of the piston, a wide range of mean flow rates, frequencies, and amplitude ratios could be obtained.

The circulating fluid was a mixture of water and glycerol (14.2% glycerol by volume) which had a density ($\rho$) of 1.40 g/ml, a viscosity ($\mu$) of 1.53 cp, and a refractive index of 1.36 at 20°C. Polystyrene particles of size range 100–300 μm were present at a concentration of 10 particles/ml and were neutrally buoyant in the suspending fluid. There was a difference in the refractive indices between the perspex of the models (1.48) and the suspending fluid (1.36). This would result in some optical distortion which was not considered important for the qualitative observations made here. Checks were made frequently to see if particle paths were dependent upon particle size. The absence of such a dependence was taken to indicate that particle paths and fluid paths were equivalent. Although it is improper to speak of a “streamline” in connection with turbulent flows, the particle motions observed here did not display perturbations perpendicular to the particle’s main path, which would be characteristic of turbulence. This may be due either to the absence of local turbulence or to a scale of turbulence that is small with respect to the particle’s size, 100–300 μm. Thus, within the Results section to follow, particle paths will be referred to as streamlines throughout.

Photographic information was obtained with a ciné camera (Lo Cam, Red Lake Laboratories) having a frame rate range of 16–500/sec. A 55-mm, f-2 Asahi Pentax lens with extension tubes was used to provide an image for continuous recording of particle movement within the models. A film projection system previously described was used. Particles were followed from upstream positions within the parent tube where Poiseuille flow (parabolic velocity profile) existed to downstream positions 2 to 3 diameters into the daughter tubes. Since the fluid entering the daughter tubes divides, and since secondary flows (flows not parallel to the center line of the conduit) exist, it was of interest to select streamlines emanating from a number of places within the parent tube cross-section and to follow these into the daughter tubes. Within the parent tube, particles near the median plane may be located by focusing on this plane, using a lens configuration of small depth of field. Particles from other parent tube positions may be located, after filming with a large depth of field, through their linear velocity, since they are in a symmetrical parabolic velocity profile having a defined relationship between radial position and local velocity.

Results

The steady-flow results of this study will be presented first in the form of drawings of streamlines to indicate the relative importance of flow rate (Reynolds number), vessel shape, and flow division ratio upon flow paths and formation of separated flows. Second, a figure will be presented for each
model on coordinates of Reynolds number and flow division ratio with areas delineated for flow separation (disturbed flow) at the locations at which it occurs within each model. Individual vessel flows will be designated as Q_L, Q_R, and Q_t, with L and R referring to the left and right branches (from the viewpoint of the observer) and t being total flow in the parent tube. A qualitative picture of pulsatile flows then will be given.

Steady Flows

The streamline patterns for the four models are shown at a common parent tube Reynolds number of 1800 (Fig. 3). Flow division in each case was made proportional to the respective flow areas of the daughter tubes. No flow separation was observed at these conditions, although some reverse flow was found for model 3.

There exists a number of similarities in the flow patterns for the four models. A group of streamlines emanating from the median plane (plane of symmetry of model) of the parent tube (labeled 1, 2, 3, 4, and 5 for models 1 and 2, and 1 and 2 for models 3 and 4 of Figure 3) moved radially from the outside wall toward the flow divider while the distance between them was reduced upon entering their respective daughter tubes. This resulted in a flow of fluid from daughter tube outside wall to inside wall within the median plane and is in part responsible for the paired counter-rotating vortices observed by us and others in the daughter tubes of symmetrical bifurcations. A complementary group of streamlines (labeled 6, 7, and 8 for all models) moved from their respective positions on either side of the median plane within their parent tubes to the outside of the daughter vessels. Model 3 displayed an additional characteristic in which streamlines 7 and 8 first enter the right branch (continuation of the aorta) before reversing direction and entering the daughter tube.

There was a change in the streamline patterns with increase in flow rate (Reynolds number). This is shown in Figure 4 with model 1, as an example. At the higher flow rate (Reynolds number equal to 3600), streamlines 1, 2, 3, 4, and 5 were more distinctly displaced toward the flow divider than at the lower flow rate (Reynolds number equal to 900). Likewise, streamlines 6, 7, and 8, which crossed from their starting positions in the parent tube to the outside of the daughter tubes, did so further upstream, i.e., closer to the daughter tube entrance, and with an apparent increase in radial velocity component. Flow separation was not observed at either flow rate.

A major shift to separated flow was observed for models 1 and 2 when the flow division ratio was altered from those specified in Figure 3, without changing the parent tube flow rate (Reynolds number). The left branch flow rates were decreased to levels of 30% (Q_L/Q_t = 0.30) and 15% (Q_L/Q_t = 0.15) of the parent tube flow rates for models 1 and 2, respectively, and flow separation (disturbed flow) resulted. Separated flow of this type could be similarly induced in model 4 by increasing the parent tube Reynolds number while maintaining the flow division ratio of the base case (Fig. 3). These flow conditions are depicted in Figure 5.

Unlike simpler two-dimensional flows which have a single separation point with a downstream set of closed streamlines forming a recirculating vortex, three-dimensional flows, such as those described above, display a locus of separation points creating a more complex pattern resulting in streamlines which do not form a closed vortex. Here flows enter and leave the vortex region, having a residence time dependent on flow rate and vessel shape. A detailed study of these variables was not made.

An additional type of flow disturbance was found to exist near the carina (flow divider) within model 3. A number of vascular flow studies have characterized this region as a high shear rate area in which locally separated flows have not been previously reported. In this case, an open vortex flow disturbance was observed for a limited range
of flow division ratio and a Reynolds number of at least 3000 and is depicted in Figure 6 by an arrow marked S.

**Quantitative Designation of Flow Disturbances**

The phenomena described in the previous section are present only for certain defined combinations of Reynolds number and flow division ratio. They may occur singly or in combination. This information is presented on a separate diagram for each model studied and is given by Figures 7-10. The lines on each figure separate areas in which different flow conditions exist. Zone A of Figure 7, for example, is a region in which flow separation occurs on the outside of the left branch. This means that all of the combinations of Reynolds number and flow division ratio to the left of the lines through the filled circles (●) give separated flow.

All of the models displayed regions where no flow disturbance occurred. Flow disturbance was absent in models 1 (Fig. 7) and 2 (Fig. 8) for a range of flow division ratio up to a Reynolds number of 4000. A change in the slope of the demarcation lines for model 1 at a Reynolds number of 2000 (Fig. 7) could be associated with transition from laminar to tur-
butions on the separation of flow and the formation of vortices were not observed for models 2, 3 and 4. Studies of model 3 (Fig. 9) and model 4 (Fig. 10) indicated regions in which simultaneous flow separation was observed at the outside of both daughter tubes. Flow disturbance near the carina was observed at elevated Reynolds numbers only for model 3.

The size of the separation bubbles, but not the points of first separation, also were found to be dependent upon Reynolds number and flow division ratio. Figures 11 and 12 are plots of projected separation bubble length measured at the intersection of the median plane with the daughter tubes of models 1 and 2. The sizes shown in Figure 11 correspond to zones A and C of Figure 7, whereas, for Figure 12, the lefthand portion corresponds to zone C of Figure 8, and the right portion to zone A of that figure. Those conditions of Reynolds number and flow division ratio farthest from the zone of no separation, B, gave the largest vortex sizes.

Pulsatile Flows

The first set of observations was made at flow division ratios that would not have led to flow separation for steady flow within the Reynolds number range chosen for pulsatile investigation (1200-1900). These operating points come from zone B for models 1, 2, 3, and 4 (Figs. 7, 8, 9, and 10, respectively). At frequencies less than 0.3 sec\(^{-1}\) (less than 12), the instantaneous flow patterns were similar to those observed in steady flow (see Fig. 3) and included a shift in pattern from the maximum to minimum of the flow pulse akin to that observed in steady flow between Reynolds numbers of 900 and 3600 (see Fig. 4). The instantaneous patterns closest in form to the comparable steady-flow pat-
FIGURE 9 Flow conditions for model 3 showing zones of separation and no separation as functions of Reynolds number and flow division ratio.

Since reverse flow may be an integral part of pulsatile flow, for some fraction of the pulse cycle, depending on the value of $\alpha$ and the flow fluctuation ratio, it is necessary to distinguish this from flow separation, which also displays reverse flow. Despard and Miller have, based on experimental evidence, established a criterion for flow separation during pulsatile flow which requires reverse flow

FIGURE 10 Flow conditions for model 4 showing zones of separation and no separation as functions of Reynolds number and flow division ratio.

FIGURE 11 Definition of separation and reattachment positions for model 1 as a function of Reynolds number and flow division ratio. This figure corresponds to conditions of zones A and C of Figure 7. Subscript $i$ refers to either left or right of this symmetrical branch.
during the entire pulse cycle. According to this criterion, flow separation was not present during the experiments described above.

A second set of observations was made at flow division ratios which would have led to flow separation for steady flow within the Reynolds number range chosen for pulsatile investigation (1200–1900). These experiments dealt only with models 1 and 2 with operating points taken from zones A and C of Figures 7 and 8. Using the criterion of Despard and Miller, we observed flow separation under conditions approximately equal to those for flow separation in steady flow. The location of the separation points for these experiments was indistinguishable from those of the comparable steady flow separation points (same mean flow Reynolds number and flow division ratio). Both flow conditions for separation and separation point were found to be independent of pulse frequency in the range of this work (0.2–0.6 sec⁻¹).

**Discussion**

The complex nature of three-dimensional flows within models of arterial branches has been clearly demonstrated here and in other studies. In part, this arises from the geometrical shapes of the transition sections at the junctions of branching tubes. These result in a potential extensive expansion and contraction of effective tube cross-sectional area for flow. Stehbens has made measurements of such geometrical changes in a detailed study on the rabbit aortic bifurcation. As has been demonstrated here, asymmetry is a second factor leading to complex flows in general and nonrecirculating vortices in particular. It has been shown previously that vortex motion is developed downstream of an asymmetrically placed obstruction in a circular tube and that the recirculating streamlines in such a flow complete a finite number of cycles before leaving the recirculation region. This is in contrast to flows downstream of a symmetrical expansion which display continuously circulating vortex streamlines. The separated flows found in the current study are analogous to those of the asymmetrical case. Although two-dimensional channel flows (in contrast to three-dimensional flows within branching circular tubes) may demonstrate separation and disturbed flows, the occurrence of these would necessarily be at flow conditions (Reynolds number and flow division ratio) quite different from those for the comparable three-dimensional cases. The lack of any disturbed flow for model 1 (aortic bifurcation) and model 2 (iliac bifurcation) for a wide range of flow division ratio and up to a Reynolds number of 4000 is an example. This has been explained previously in

**Figure 12** Definition of separation and reattachment positions for model 2 as a function of Reynolds number and flow division ratio. The lefthand portion of this figure corresponds to conditions of zone C of Figure 8 and the righthand portion to zone A of Figure 8.
terms of a flow cross-over phenomenon wherein high velocity fluid from the central core region of the parent tube crosses over to the outside of the daughter tube, thus preventing boundary layer separation, and is consistent with the streamline patterns demonstrated here. The forces necessary to develop such flows are not possible in a two-dimensional channel configuration.

This work suggests that there is a wide range of physiological conditions under which disturbed flow is not apt to occur (zone B of Figures 7-10). However, exercise is a condition in which cardiac output is increased and flow redistribution occurs, thus, potentially creating conditions suitable for disturbed flow. During moderate exercise, cardiac output may increase 300% (6 liters/min to 18 liters/min) while renal flow decreases by 50-80% (1.2 liters/min normally) and flow to the mesentery decreases substantially. Digestion is another condition resulting in redistribution of blood flow; here, flow to the superior mesenteric artery may increase by as much as 132%. At rest, the mean flow to a single renal artery is at least 15% of flow in the abdominal aorta, corresponding to conditions of no flow separation for most of the cardiac cycle, i.e., zone B of Figures 9 and 10 for models 3 and 4. With exercise, Reynolds number increases and flow division ratio \( Q_L/Q_I \) decreases, thus creating conditions more suitable for flow separation, i.e., zones A or AC of Figures 9 and 10 for models 3 and 4. It is unlikely, from our data, that digestion would create flow separation but, more likely, a shift from the left to right portion of zone B of Figures 9 and 10.

The separated flows observed here all had open recirculating regions. This means that fresh upstream fluid entered and left the separation bubble simultaneously. Platelet aggregates have been visually observed to form within the downstream vortex of a symmetrical-tubular expansion exhibiting closed circulating streamlines. In ex vivo stagnation flow, platelet aggregates likewise have been generated both for vortices with closed (symmetrical) and open (asymmetrical) streamlines. Under conditions of higher degrees of asymmetry and greater flow, aggregates were not formed; this was explained in terms of decreased platelet collision frequency and residence time within the vortex. The convective removal of aggregation stimuli also may have been a factor. It is possible then that the separation regions demonstrated here could be areas where platelet aggregates form, possibly dependent on the degree of vortex asymmetry. It is also known that platelet aggregate formation occurs concurrently with the platelet release reaction which liberates such substances as adenine nucleotides (AMP, ADP, ATP), serotonin, and a factor necessary for smooth muscle cell proliferation in culture. One may then speculate that over long periods of time, months to years, vessel walls adjacent to continuously or discontinuously disturbed flow could be injured biochemically by such substances.

In vitro pulsatile flow studies with whole blood have shown that boundary layer separation occurs at a threshold flow rate (Reynolds number) and that localized velocity perturbations within the separation zone are present. The model used by these workers is similar to model 3 of this study. Turbulence and localized velocity fluctuations akin to those found in vitro occur in vivo at the junction of the abdominal aorta and iliac artery in dogs. Similar flow disturbances were found in vivo at the ilioaortic junction in pigs; these correlated with ultrastructural degenerative changes of endothelium and loss of endothelial cells. Although flow separation is implied in these in vivo studies, they do not contradict our findings, that no flow separation was observed for a wide range of flow division ratio in a model of an abdominal aortic bifurcation. This may be explained by the fact that the dog and pig have aortic trifurcations in which the area ratio changes within the junction may be sufficiently elevated to induce flow separation not found in a bifurcation.

The present study supports the work of others and further indicates that arterial shapes are prone to generate flow disturbances at certain focal locations. This study additionally points out that the selection of a vascular location alone is insufficient to unequivocally state that flow disturbance must occur at that point. To do this, the flow rate and flow division conditions also must be known. In vivo these variables may be affected by a host of stimuli which include detailed vessel shape, exercise, and the resistance states of downstream muscle beds. It thus becomes particularly important that in vivo studies which purport to link pathological conditions such as atherosclerosis and thrombosis with hemodynamic phenomena carefully define flow variables.

References

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