Blood Velocity Profiles in the Origin of the Canine Renal Artery and Their Relevance in the Localization and Development of Atherosclerosis


Using a 20-MHz 80-channel pulsed Doppler velocimeter and 30-MHz high-resolution echo ultrasound, we investigated the in vivo hemodynamics at the origin of the renal artery by measuring the velocity profiles and bifurcation geometry of a surgically exposed left renal artery in 10 anesthetized dogs. The angle between the aorta and the renal artery ranged from 60° to 90° (mean, 84°) although the bifurcation did not lie in a single anterodorsal plane and the diameter of the renal artery ranged from 1.5 to 3.5 mm (mean, 2.4 mm). Despite different geometries, the velocity profiles in the different aortorenal bifurcations were similar. Although regions of reverse velocity were observed, the net flow in the renal artery was in the forward direction throughout the cardiac cycle. The peak Reynolds’ number was 486±63. The velocity profiles in the proximal renal artery in the plane parallel to the bifurcation showed velocity vectors directed toward the caudal wall throughout the cardiac cycle. Reverse flow, indicating flow separation, was observed near the cranial wall even during systole. When the probe was placed on the cranial wall perpendicular to the wall, a velocity component from the cranial side to the caudal side was observed. At a distance of four diameters from the renal ostia, velocity profiles were almost parabolic. These results indicate that the velocity pattern near the cranial wall at the renal ostia, at which atherosclerotic lesions are prone to develop, are characterized by 1) a low time-averaged shear rate, 2) separation of the flow, and 3) a time-varying oscillation of the flow. (Arteriosclerosis and Thrombosis 1992;12:626–632)

KEY WORDS • renal artery • blood velocity profile • atherosclerosis • pulsed Doppler velocimeter

Early atherosclerotic lesions tend to be found at bifurcations and the curved regions of arteries, and it has been suggested that hemodynamic factors play an important role in lesion development. This hypothesis is supported by the rapidly growing number of effects that blood flow appears to exert on the walls of blood vessels.1 Particular blood flow patterns may influence the shape and orientation of endothelial cells,2 the release of various autacoids,3,4 and the mass-transport properties of the vessel wall.5 Some of the effects appear to depend on not only the magnitude of the wall shear stress but also its time- and direction-dependent characteristics.6

The significance of these interactions between blood flow and the function of the vascular wall in the atherosclerotic process is unclear. In addition, the blood flow patterns in those arterial regions that are vulnerable to atherosclerosis have not been well defined because of the limitation of accessibility and the spatial and time resolution of blood flow measurements.

In the present study we measured in detail the blood velocity profiles at the origin of the renal artery in the dog. In humans, atherosclerotic lesions are prone to develop within the proximal 1–2 cm from the renal ostia7,8 on the cranial side.9 These lesions can be pathologically important because they may reduce renal blood flow and thereby cause renal hypertension by increasing the plasma level of renin. We investigated the velocity profiles in the renal branch of the aortorenal bifurcation using a high-frequency (20-MHz, 80-channel) ultrasound Doppler velocimeter. Because of the complexity of the local hemodynamics at the bifurcation, blood velocity profiles have been measured in different planes (horizontal and vertical) and at different beam angles to the long axis of the renal artery.

**Methods**

**Measuring Devices**

20-MHz, 80-channel pulsed Doppler velocimeter. Blood velocity profiles were measured with a high-frequency pulsed Doppler velocimeter developed at Kawasaki...
Medical School in collaboration with Fujitsu Lab Co., as previously described in detail.\textsuperscript{10,11} This device measures the Doppler shift from 80 channels in real time with sample volumes of about 0.8 mm\(^2\)×0.2 mm (radius, 0.5 mm; thickness, 0.2 mm). The high-pass filter had a cutoff frequency of 375 kHz. Specially designed plastic cuffs enabled the operator to place the transducer on the exposed renal artery at angles of either 60° or 90° to the long axis of the vessel (Figure 1). A layer of gel allowed normal wall motion. In the velocity profile displays the vertical axis indicates distance across the width of the vessel while the horizontal axis indicates both time and velocity (see Figures 4, 6, 7, and 8). Doppler ultrasound measures only the component of blood velocity in the direction of the beam. If blood flow is parallel to the wall of the artery, the absolute velocity can be calculated from the beam angle and the Doppler shift recorded by this device. However, when the direction of the velocity is not known, as is the case for blood flow in the region of a bifurcation, it is important to remember that the velocity profiles are really profiles of the component of velocity in the direction of the beam.

\textit{30-MHz high-resolution B-mode device.} Measurements of the geometry of the left renal artery and the abdominal aorta were made with a high-resolution B-mode echo device (prototype, Omron, Kyoto, Japan). The probe, which is 4 cm long, 3 cm wide, and 1 cm thick, had a linear array of 84 transducers for a total length of 15 mm. It was placed in contact with the exposed vessels and gave 30 images per second with a spatial resolution of 0.2 mm. Figure 2 shows typical examples of transverse and longitudinal images of the renal artery obtained with this device.

\textbf{Animal Preparation}

Measurements with both probes were made on the renal arteries of 10 mongrel dogs of either sex weighing between 13 and 19 kg. They were anesthetized with sodium pentobarbital (25 mg/kg i.v.) and after intubation were ventilated with room air with a Harvard respirator and supplemented with oxygen to maintain a normal arterial oxygen tension. After the chest was opened by a median sternotomy, an atrioventricular block was achieved by injecting 40% formalin into the atrioventricular node. The heart was then paced at 120 beats/min via wires sewn onto the right ventricle. The chest was closed, and the renal artery and the aortorenal bifurcation were exposed by bloodless tissue plane dissection through a median abdominal incision.

Blood pressure was measured with a catheter-tip pressure transducer (Millar microtipped catheter transducer, model SPC-784A, Millar Instruments, Houston, Tex.) that was advanced from the carotid artery to the suprarenal portion of the abdominal aorta. The mean blood pressure during the periods of experimental observation ranged between 84 and 105 mm Hg (91.5±5.5; mean±SD). The dogs were allowed to stabilize for
FIGURE 3. Sketch of the aortorenal bifurcation showing Doppler ultrasound measurement sites. The angle between the ultrasound beam and renal artery axis was 60° for sites 1, 2, 3, and 4V and 90° for site 1P. D, diameter of the proximal portion of the left renal artery.

approximately 1 hour after the operation. The experiments performed complied with and were approved by the Institutional Animal Care and Use Committee of Kawasaki Medical School.

Experimental Procedures

Bifurcation geometry. The branching angles between the aorta and the renal artery were measured from photographs taken during the operation. The diameters of the vessels as well as the configuration of the aortorenal bifurcations were observed with the B-mode ultrasound probe placed on the ventral surfaces of the aorta and the renal artery and on the cranial side of the renal artery (Figure 2). This device also confirmed that there was no stenosis due to atherosclerotic plaques in these vessels. After the experiments were completed and a radiograph was taken, a cast of the bifurcation was made by injection under hand pressure of methacrylate into the upstream vessel.

Velocity measurements. The exposed vessels were bathed in saline, and ultrasound gel was placed between the Doppler probe and the left renal artery. The position of the probe was adjusted until the velocity profiles were of maximum width, indicating that the ultrasound beam traversed the midline of the vessels.

Figure 3 is a three-dimensional illustration of the bifurcation indicating the measurement sites. The velocity profiles were measured both in the plane closest to that of the bifurcation (sites 1, 1P, 2, and 3) and perpendicular to that plane (site 4V). The probe holder enabled the beam angle to be positioned at 60° to the vessel axis at sites 1, 2, 3, and 4V and at 90° at site 1P, which was 0.5 diameter downstream from the ostium. In the horizontal plane site 1 was immediately distal to the renal ostia, and to confirm the velocity profiles at that site, we also measured the profiles in the contralateral direction (from the caudal side) in three of the dogs.

TABLE 1. Geometry and Hemodynamic Parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Range</th>
<th>Mean±SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Geometry</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Diameter (mm)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aorta</td>
<td>7.0-8.1</td>
<td>7.4±0.5</td>
</tr>
<tr>
<td>Renal artery</td>
<td>1.5-3.5</td>
<td>2.4±0.5</td>
</tr>
<tr>
<td>Angle (degrees)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aorta to renal artery</td>
<td>60-90</td>
<td>84.0±8.7</td>
</tr>
<tr>
<td>Hemodynamic parameters</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean blood pressure (mm Hg)</td>
<td>84-105</td>
<td>91.5±5.5</td>
</tr>
<tr>
<td>Pulse rate (pacing; beats/min)</td>
<td>120</td>
<td>120</td>
</tr>
<tr>
<td>Peak Reynolds' number</td>
<td>310-611</td>
<td>486±63</td>
</tr>
<tr>
<td>Unsteadiness parameter</td>
<td>2.1-4.6</td>
<td>3.1±0.6</td>
</tr>
</tbody>
</table>

n=10.

To evaluate the effect of the aortorenal angle on the velocity profile, an additional experiment was performed on three dogs (weight, 19-24 kg). In these dogs the kidney together with the main renal vessels was detached from the retroperitoneal tissues, and the aortorenal angle was changed to 90°, 60°, and 30° while the velocity profiles were measured at site 1. The change in diameter of the renal artery as observed by the 30-MHz echo was negligible following the changes in the angle.

A two-dimensional velocity pattern was calculated for the region just distal to the bifurcation by vector addition of the profiles measured at sites 1 and 1P. The flow in the renal artery was also characterized by calculating the peak Reynolds' number and a frequency parameter. The peak Reynolds' number is defined as

$$Re = \frac{Ud}{\nu}$$

and the frequency parameter (often called the Womersley number in the physiological literature) is defined as

$$\alpha = \frac{d}{2\sqrt{\frac{\omega}{\nu}}}$$

where $U$ is the maximum instantaneous velocity, $d$ is the vessel diameter, $\omega$ is the cardiac frequency, and $\nu$ is the kinematic viscosity of blood (assumed to be $4 \times 10^{-6}$ m$^2$/sec).

Results

Bifurcation Geometry

The aortorenal junction showed three-dimensional curvature, as illustrated in Figure 2. That is, the renal artery initially lies more ventrally than the abdominal aorta and then passes to the most dorsal surface of the abdominal cavity. The angles of the flow divider between the renal artery and abdominal aorta in the horizontal plane are given in Table 1, together with the vessel diameters and the calculated hemodynamic parameters. Turbulent flows are unlikely to occur at these Reynolds' numbers and frequency parameters.12

Blood Velocity Profiles

Sites 1, 1P, and 4V. Figure 4 shows a typical example of the velocity profiles observed at sites 1 and 1P during the first half of the cardiac cycle. Reverse velocities were
present at the cranial side of the vessel during the whole of systole in all dogs, with the greatest amplitude occurring close to peak flow and indicating maximum flow separation at this time. The velocity profiles during the later half of the cardiac cycle indicated low, axisymmetric, and nonreversing flow.

With the probe at site 1P (at 90° to the cranial side of the vessel), the velocity vector during systole was directed away from the cranial to the caudal side, indicating the presence of secondary flow in this direction. Vector addition of the velocity profiles from sites 1 and 1P gave rise to a two-dimensional vector map at the time of maximum flow (Figure 5). The velocity vectors of greatest magnitude were directed toward the caudal wall, and at the cranial side the vectors indicated a zone of flow separation. The velocity profiles measured at site 1V (Figure 6) were M shaped, with the velocity depressed in the center of the vessel. This M shape is also indicative of secondary flow.

Figure 7 shows the velocity profiles during one cardiac cycle at site 1 for three different aortorenal angles as this angle was altered in a single dog. Flow reversal at the cranial side wall during early systole was clearly seen for the 90° and 60° bifurcations. In the 30° bifurcation the reverse flow velocities were still visible but were very small.

Sites 2 and 3. Figure 8 shows the blood velocity profiles at sites 2 and 3 during the first half of the cardiac cycle. At site 2 the velocities still appeared to be greatest on the caudal side, but the flow reversals at the cranial wall seen at site 1 were either very small or not present. At site 3 the profiles were almost axisymmetric.

Discussion

This study has demonstrated the phasic characteristics of blood velocity at the bifurcation of the canine renal artery, as measured by a high-frequency pulsed Doppler velocimeter. Our conclusions and interpretations depend on three factors that include 1) a critique of the method used for measuring velocity profiles, 2) the relation between vessel geometry and the velocity profiles, and 3) the clinicopathological significance of the data.

The Doppler Ultrasound System

This or similar devices have been used to assess velocity profiles in a wide or a similar range of vessels. However, there are still several limitations to the measurement of velocity profiles. First, a measured velocity profile may deviate from the actual profile.

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**Figure 4.** Representative recordings of blood velocity profiles at sites 1 and 1P (see Figure 3). Vertical axis indicates distance across the width of the vessel, the horizontal axis indicates velocity, and the different curves represent velocity during the first half of the cardiac cycle. Note the difference in expression of blood velocities between site 1P and other sites. In the velocity profile at site 1P, the direction of blood flow is in the plane perpendicular to the renal artery. Backward flow velocity indicates flow from the cranial side to the caudal side wall. Dark shading indicates reverse flow velocities. D, diameter.

**Figure 5.** Diagram of two-dimensional velocity vectors calculated from two directional measurements made at sites 1 and 1P (see Figure 3) at the time corresponding to peak velocity. Lt., left.

**Figure 6.** Example of a typical blood velocity profile measured at site 1V (see Figure 3).
Because of the spatial distribution of acoustic power within the sample volume even though the sample volume of the system is small. According to the simulation study by Brown et al., the error in the measurement is small for a vessel with a diameter larger than about 2.0 mm. We confirmed the accuracy of our velocity profile measurements of blood flow in model tubes that were 2 and 3 mm in diameter. Second, broadening of the transducer's ultrasound beam and energy attenuation of the back-scattered Doppler signals from a distant sampling region may cause some errors in velocity profile measurements. Before in vivo measurements were made, the Fujitsu Lab Co. evaluated the acoustic pressure distribution experimentally. The beam width, defined as the distance between the points 12 dB below the center line intensity, was effectively constant for a distance of about 4 mm from the probe. Beyond 4 mm the "12-dB" lines diverged at an angle of 3°. The diameter of the canine renal artery was 2.4±0.5 mm. Thus, the underestimation due to the sample volume was small, and the beam path length was sufficiently short to avoid unacceptable diminution of the signal.

The algorithm used for calculating velocities from the Doppler shift frequency assumes that velocity vectors in the vessel are parallel to the vessel walls. In a region of complex flow, such as the neighborhood of a bifurcation, this will not be the case. For this reason we have used the technique of vector addition, assuming that all the Doppler shift measured with the probe at site 1P was...
associated with velocity vectors that were perpendicular to the long axis of the vessel.

**Bifurcation Geometry and Velocity Profiles**

Despite the variation in the geometry of the bifurcations, the velocity profiles measured in the renal arteries of the 10 dogs were not markedly different. All dogs showed flow reversal and apparent flow separation at the cranial side of the renal artery, even the dog with only a 60° angle between the renal artery and the aorta. This finding is contrary to previous observations, which suggested that branching angles may be a major factor in determining velocity profiles.17 Langille,18 however, also reported that local geometry may not be a predominant factor, as the alignment of endothelial cells at the ostia of the renal artery, the superior mesenteric artery, and the celiac trunk was similar from animal to animal despite considerable variations in branching angles from 60° to 90°. Our experiments in which the aortorenal angle was changed in the same dog also support this observation although for the smallest bifurcation angle (30°), very little flow separation was observed. By gradually moving the position of the transducer away from site 1, we were able to estimate the location of the reattachment point although the size of the transducer and the low-frequency cutoff made exact determination impossible. In each case, the length of the separation region was approximately one diameter from the origin of the vessel.

Because a high-pass filter (cutoff frequency, 375 kHz) is used to eliminate the large Doppler signal associated with the motion of the wall, there is uncertainty in the velocity profiles regarding the exact position of the wall, and therefore, it is not possible to determine the wall shear stresses exactly at any site. However, at site 3 the axisymmetry of the velocity profiles suggests that the wall shear stresses are likely to be similar at all points around the circumference of the vessel; at site 1 the shear stress at the caudal wall appears greater than that at the cranial wall. Perhaps more importantly, the shear stress appears to oscillate in direction at the cranial wall at site 1 whereas at all other positions in the vessel it appears to be unidirectional. Zarrir et al19 have suggested that shear stress oscillation may be a more important determinant of the location of atherosclerosis than the time-averaged shear stress.

Secondary flows occur in curved tubes or in branched tubes where there is a change in the primary direction of flow. There have, however, been few reports demonstrating its existence in arteries in vivo. Secondary flows were present in the upstream portion of the renal artery in this study, giving rise to the radial velocity component from the cranial to the caudal side at site 1P and the M-shaped profile at site 1V. Both secondary flows and a zone of separation might be expected at the origin of the renal artery on the basis of earlier mathematical and experimental model studies20,21 and the measured Reynolds' numbers and frequency parameters.

These measurements were made in canine renal arteries. The incidence of atherosclerosis in this animal is low, however, unless it is induced by hyperthyroidism or a rise in the level of dietary lipid induced by coconut oil feeding.22 In rabbits that were fed cholesterol, lesions were observed at the same site, and the distribution of lesions was altered by surgical modifications that would have altered the pattern of blood flow.23 In dogs there was a major difference in the velocity profiles measured at site 1 and those at site 3, which was four diameters downstream. In humans lesions occur within two diameters of the ostium,24 i.e., the region within which flow separation was observed in the dogs. Lesions have rarely been observed in the more distal parts of the artery; these regions were characterized by nonreversing flow in the dogs. Our preliminary measurements in the human aortorenal bifurcation, obtained during retroperitoneal surgery, suggest that the velocity patterns are similar to those observed in the dogs, which if confirmed, will support the validity of the present study for the prediction of hemodynamics in the human aortorenal bifurcation.

**Clinicopathological Interpretations**

The velocity pattern near the cranial wall of the renal artery just distal to the renal ostium, on which atherosclerosis is prone to develop, was characterized by a region of blood flow separation with relatively low but directionally oscillating shear stress. This environment could have many adverse effects on the adjacent vessel wall that could cause the initiation and development of atheromas. Such effects might include an increase in the residence time24 or the adhesion of circulating particles, such as monocytes, granulocytes, or platelets.25 Flow stagnation near the cranial wall might also delay the renewal of nutritive substances and the elimination of metabolites.26 There might also be direct detrimental effects of the oscillatory shearing stresses on the endothelium, thus causing changes in receptor or autacoid expression or blood tissue transport processes.27

**References**


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