Abnormal Wall Strain at Distal End-to-Side Anastomoses

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Cyclic stretch has been demonstrated to induce proliferative and secretory activities by cultured arterial endothelial and smooth muscle cells, cellular processes that contribute to the development of intimal hyperplasia. A model of an end-to-side anastomosis was developed to examine the hypothesis that regions of the artery at such anastomoses are subjected to focally increased cyclic stretch, which may stimulate the development of intimal hyperplasia. Polytetrafluoroethylene grafts were anastomosed end to side to latex rubber tubes that have elastic properties similar to those of the human femoral artery. Pulse waves with physiologic pressure, rate, and contour were applied, and systolic and diastolic diameters were measured in two planes at longitudinal intervals. Circumferential strain imposed on the latex "artery" was calculated at each interval. Strain imposed perpendicular to the suture line was also measured. Circumferential strain was consistently maximal at a distinct region of the "artery" along the proximal third of the anastomosis (6.0 ± 1.1% vs. 3.3 ± 0.5% at other regions of the "artery"). The maximal strain across the suture line was found at precisely the same region (3.9 ± 0.3% vs. 2.0 ± 0.4%). The anastomotic region of the recipient artery in a distal end-to-side anastomosis is subjected to cyclic circumferential strains two times greater than those experienced by the remainder of the artery. This corresponds to a common location of intimal hyperplasia. Such strains may be a stimulus for intimal hyperplasia. (Ann Vasc Surg 1993;7:14-20.)

The end-to-side anastomosis is the most common configuration used at the distal end of arterial bypass grafts. In contrast to the end-to-end anastomosis, the end-to-side anastomosis preserves antegrade and retrograde flow in the outflow artery, maintains flow through collateral channels, and is easier to construct with small recipient vessels. The flow characteristics through the end-to-side anastomotic configuration have been extensively studied. Such studies have examined the effect of graft angle, graft diameter, and proximal and distal patency of the outflow vessel on flow turbulence, energy loss of the flowing fluid, and boundary layer separation. These studies were undertaken in an attempt to define the physiology of the end-to-side anastomosis in order to determine the ideal configuration to maximize volume flow and to understand the pathophysiology of graft occlusion. However, another major aspect of the physiology of the end-to-side anastomosis, the mechanical strains acting on the vessel wall, has not been studied directly. At present, the effect of these strains is unknown, as these parameters have not been described. The radial stretching of an arterial wall can be described and quantitated using the concept of circumferential strain (change in circumference/baseline circumference). Experimental evidence suggests that cyclic arterial stretching may be a stimulus for the development of intimal hyperplasia.

Thus a model was developed to evaluate the focal circumferential strain imposed on the recipi-
ent artery by a distal end-to-side anastomosis. The configuration of a stiff “graft” anastomosed to a compliant “artery” was used, as this simple mechanical configuration most closely mimics the clinical conditions of a prosthetic graft (which is virtually noncompliant) anastomosed to an artery. A profile was constructed of the focal strains applied to the wall of the “artery” with each pulse cycle. As cyclic stretch has been shown to induce endothelial cell mitogenesis and in turn secretion of extracellular matrix proteins by smooth muscle cells, it was proposed that focal areas of increased circumferential strain would correlate with those regions in the artery known to be common foci of intimal hyperplasia.

MATERIAL AND METHODS

The model consisted of an expanded polytetrafluoroethylene (PTFE) tube (the “graft”) anastomosed end to side to an isodiametric latex rubber tube (the “artery”).

Three individual models were constructed and tested (Fig. 1). A longitudinal incision 1.4 to 1.6 cm long was made in the wall of a ¼-inch diameter Penrose latex rubber tube (Argyle, St. Louis, Mo.) at least 12.5 cm in length. An appropriate bevel, at about 60 degrees, was cut in the end of a 6 mm internal diameter thin-wall PTFE tube (Impra Inc., Tempe, Ariz.) of 7 cm length. The end of the PTFE tube was then anastomosed to the side of the latex rubber tube with a continuous 6-0 polypropylene suture (Surgilene, Davis and Beck, Pearl River, N.Y.). To seal the suture holes and prevent leakage, liquid latex rubber molding compound (ETI, Fields Landing, Calif.) was diluted with water, poured into the lumen of the model, drained immediately, and allowed to dry overnight. This resulted in a very thin, highly compliant, virtually watertight lining adherent to the lumen of the model.

Both ends of the latex rubber tube and the free end of the PTFE tube were cannulated with steel nozzles, and the model was mounted on an adjustable rig (Figs. 2 and 3). The “proximal” end of the latex rubber tube was occluded, and the “distal” end was connected to a membrane pressure transducer (P23ID, Gould Inc., Oxnard, Calif.). The PTFE tube was connected to a side arm of an hydraulic pulse-generating circuit utilizing a bellows pump (SP80-30, Waltham Chemical Pump Corp., Waltham, Mass.). This system provides a physiologic pulsatile pressure wave at a stable pressure and rate (130/80 mm Hg at 120 “beats”/min).

The motion of the walls of the model was recorded with a video-based motion analysis system. A modified (noninterlaced) video camera (TI-23A CCD camera, NEC Corp., Japan) coupled to a video cassette recorder (Panasonic AG-6300, Matsushita Electric Ind. Co., Ltd., Secaucus, N.J.) was used to record an image of the model while the pulse wave was applied. The video images were then analyzed using a processor (VP 110, Motion Analysis Corp., Santa Rosa, Calif.) that follows the motion of the high-contrast edges delineating the border between the model and background. Motion is quantitated as image pixels per unit time sampled at a rate of 60 Hz. Applying an appropriate scale factor, the true change in diameter with time was determined. The data were processed and stored in a 386-PC computer (Amdek Corp., Elk Grove, Ill.).

To quantitate motion other than that of the vessel edges, an array of ink dots was applied to the surface of the PTFE and latex rubber tubes in the region of the anastomosis (Fig. 1). The relative motion of these dots was also recorded with the video system and used to calculate the strain across the suture line.

Images of the pulsating “vessel” were sequentially obtained in the lateral and posteroanterior planes. This allowed determination of the lateral and posteroanterior diameters at diastole and systole. These diameters were measured at longitudinal intervals of 1.0 mm. These measurements permitted calculation of the diastolic and systolic circumference of the model at 1 mm intervals, assuming that the luminal cross section outside of the anastomotic region was a circle and that between the heel and the toe of the anastomosis it approximated a regular ellipse:

![Fig. 1. A 6 mm PTFE “graft” anastomosed to a ¼-inch latex rubber “artery.”](image-url)