Computational Fluid Dynamics and Bifurcated Stent Graft Design

NOTES

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Early stent grafts were unable to withstand the effects of arterial pressure and flow. Long-term results were marred by migration, component separation, fatigue fracture, and graft erosion. Second- and third-generation devices have generally fared better, yet preclinical testing of new designs remains an imprecise art in the absence of information on the hemodynamic environment. Most previous studies of hemodynamic forces on stent grafts have assumed steady state conditions and a sac pressure of zero.

Methods

We used computational methods to map temporal and spatial variations in the flow, pressure, and shear and overall forces on real (computed tomography-derived) and hypothetical stent graft geometries. We based the pressure modeling of in situ bifurcated stent grafts on the sac pressures found in a variety of clinical circumstances relating to endoleak status. We calculated summated forces for the whole stent graft, the trunk alone, the bifurcation alone, and the legs. Hypothetical geometries allowed us to study the effects of individual parameters, such as tapering and angulation.

Results

All these studies showed that pressure-related forces were far larger than flow-related forces for all stent-graft geometries, real or hypothetical, whatever the sac pressure. For example, doubling the inlet blood pressure nearly doubled the total peak displacement force on the stent graft, whereas doubling the blood flow produced only a 6% increase on the total peak displacement force. These pressure-related forces depended on only two factors: the pressure gradient across the walls of the stent graft, and the relative areas and orientations of the inflow and outflow.

The area around the bifurcation was generally subject to the largest forces because it was the site of greatest change in total cross-sectional area. The axial component of the net force on the bifurcation was directed the caudally in all four cases. The forces on other parts of the stent graft were different in both direction and magnitude. Since the stent graft was wider inside the aneurysm than inside the neck, the axial component of the net force on the trunk was directed cranially in all four cases. The summated forces for the stent graft as a whole were generally larger in the transaxial direction (sagittal and coronal combined) than in the caudal-cranial direction, especially in cases with significant degrees of tortuosity.

None of the real geometries contained areas of tapering within the trunk, but this was a feature of some hypothetical geometries. Because the trunk generally has a large cross-sectional area, we found that a tapered trunk can make a significant contribution to the overall displacement force.

Sac pressure had a striking effect on net forces. For example, the axially directed systolic force at the bifurcation was 70% higher with the low sac pressures of a shrinking aneurysm than with high sac pressures of a dilating aneurysm. Although the absolute effect of sac pressure reduction on diastolic forces was the same, the proportionate was even greater (570% increase).

Implications for Stent Graft Design

It has been common practice to significantly oversize the main body of the stent graft to ensure apposition and attachment. Our CFD studies suggested that oversizing may ultimately affect displacement force and migration risk. Immediately following surgery, the proximal stent of an oversized stent graft is restricted by the diameter of the native abdominal aortic neck, but in time the neck dilates until it is limited by the diameter of the graft component of stent graft. Therefore, the degree of oversizing, and not the preoperative aortic neck diameter, determines the ultimate inlet diameter. Since inlet diameter is a principal determinant of displacement force and migration potential, the larger the stent graft, the higher the risk of migration. There is some support in the literature for the idea that greatly oversizing the stent graft may increase the risk of migration, although the true explanation for this clinical observation may lie elsewhere.

The relative lengths of the trunk and the limbs may influence force distribution and migration risk. Short stiff limbs are capable of transmitting force from the bifurcation of the stent graft, where most force is applied, to the bifurcation of the aorta. Long floppy limbs cannot do this, so all the displacement force on the bifurcation has to be borne by the proximal attachment between the trunk of the graft and the neck of the infrarenal aorta. Consequently, stent grafts with short stiff limbs are probably less prone to proximal stent migration than are stent grafts with long floppy limbs.

Stent graft-specific variations in the sac pressure may affect migration risk. High sac pressures lower the pressure gradient over the graft wall and reduce migration forces. Clinical studies have shown that some stent graft designs are associated with higher residual sac pressure, low rates of aneurysm shrinkage, and high rates of aneurysm dilatation, even in the absence of endoleak. This lack of change in the natural history of the aneurysm has prompted manufacturers to reduce the porosity of the offending stent grafts. However, it is probable that, if effective, these changes will have the paradoxical effect of increasing migration rates because lower sac pressure increases both the displacement force and the migration rate.

Conclusion

Bifurcated stent grafts with short, wide, stiff limbs are probably more stable than those with long, narrow, flexible limbs. Stent grafts that are effective in reducing sac pressure must withstand more force than those that fail to do so. Changes in stent graft design, such as reductions in porosity, that are intended to reduce sac pressure will likely increase migration rates unless accompanied by improved fixation.