醫工學會誌：第 13 卷, 第 2 號, 1992
J. of KOSOMBE : Vol. 13, No. 2

研 究 論 文
92-13-2-10

Flow Dynamics Near End-To-End Anastomoses
—Part Ⅰ. In Vitro Compliance Measurement—

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=Abstract=

Compliance mismatch across an end-to-end anastomosis was measured in the in vitro experimental setup. A 35mm camera was used and image process was done in Gould/DeAnza image processor. The results showed that compliances of Penrose tubing and synthetic PTFE grafts were in good agreement with the previously reported in vivo data. PTFE grafts exhibited a nonlinear behavior with compliance decreasing with increasing transmural pressure, whereas the compliance of the Penrose tubing remained relatively constant within the range of the pressures in which data were obtained. The lumen cross-sections at the anastomosis were affected by the suture and the mismatch in compliance between the Penrose tubing and vascular grafts. The variations in the lumen diameter at the anastomosis was more pronounced with increasing transmural pressures. From the present study, it was clearly demonstrated that the compliance of prosthetic grafts is much lower than that of the arteries. In addition to the hemodynamic consequences, compliance mismatch across the anastomosis has been known to lead to increased anastomotic and suture stresses with resultant suture line dehiscence and false aneurysm formation. Thus, there are good hemodynamic reasons to suppose that introduction of a less compliant arterial graft into the arterial circulation will be damaging and that grafts should be made to match the elastic behavior of their host arteries as closely possible.

INTRODUCTION

Changes in compliance have been emphasized in cases of atherosclerosis, aneurysms, and subintimal hyperplasia. Compliance mismatch between the host artery and the vascular graft has been believed to be the most probable cause of graft failure [1-4]. It is possible that those areas of compliance mismatch may have abnormal flow dynamics, particularly wall shear stress, at the anastomotic site, inducing thrombus deposition. There are several synthetic grafts available for graft implantation. Dacron and polytetrafluoroethylene (PTFE) are most commercially available synthetic graft material. The clinical introduction of expanded polytetrafluoroethylene (e-PTFE) prostheses has so improved the short-term patency of synthetic small-artery grafts that subintimal hyperplasia, especially at the distal anastomosis, has been noted more frequently. However, Walden et al. [5] analyzed long-term patency rates of grafts, based on the life table method and correlated

<접수 : 1992년 7월 14일>
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them with graft compliance. The authors demonstrated that more compliant grafts such as saphenous vein graft or glutaraldehyde-treated umbilical vein graft performed much better than stiffer synthetic grafts.

Nicolaides[6] reported that the clinical performance of the graft depends on the mechanical properties of the graft and the response of the body to the prosthesis in terms of fibrous tissue reaction and thrombus deposition. Cengiz et al. [7] performed in vitro and in vivo studies on the effects of compliance changes on healing of a porous Dacron prosthesis in the thoracic aorta of the dog. Abbott et al. [2] examined the effect of lower compliance upon patency in autogeneous vein grafts. Even though the showed that the compliance was decreased in both two weeks and three months. Again, it appeared in their studies that the compliance of a graft should be closely matched with that of the host artery. Hasson and Abbott[8,9] performed compliance studies both in vivo and in vitro. Their studies revealed the presence of varying compliance along the conduit and across the anastomosis. Additionally, these authors described the presence of a consistent zone of hypercompliance at certain points in proximity to the anastomotic suture line both proximally and distally. The para-anastomotic hypercompliant zone (PHZ) represents a transmitted effect of the suture line to adjacent areas of the graft and the artery. Hasson et al.[10] reported later that a suture line created additional local compliance mismatches near the anastomosis of artery and graft and suggested that it might affect the transmission of the stress in the anastomotic region, regardless of anastomotic geometry. The major purpose of the present in vitro compliance study is to determine the lumen cross-sections and compliance in the vicinity of the anastomosis for various intraluminal pressures.

METHOD

1. Specimens

Real canine arteries have too many variables in the in vitro experiments such as tethering, taper, branches and their different sizes. In order to simulate the compliance of the human femoral artery, the Penrose surgical drainage tubing (6.35mm diameter) was used for control studies, since it has approximately the same compliance as femoral arteries. 6mm-diameter Gore expanded PTFE vascular grafts (W. L. Gore & Associates, Inc., Elkton, M.D.) of two

Table 1. Number of specimens in the compliance measurement study.

Categories		Number of specimens
case 1	Penrose tubing segments	3
case 2	Standard wall PTFE graft segments	3
case 3	Thin wall PTFE graft segments	3
case 4	Penrose tubing segments with silicone gel	3
case 5	Standard wall PTFE graft segment with silicone gel	3
case 6	Thin wall PTFE graft segments with silicone gel	3
case 7	Penrose tubing-Penrose tubing anastomosis	3
case 8	Penrose tubing-standard wall PTFE graft anastomosis	3
case 9	Penrose tubing-thin wall PTFE graft anastomosis	3

different thicknesses were used as graft specimens. 7-0 Prolene (Ethicon, Inc., Somerville, N.J.) suture was used with continuous suturing technique for the anastomosis by the same vascular surgeon. Thin layer of silicone gel was applied on the place under the magnification lenses where leak occurred across the anastomotic region.

Table 1 shows the numbers of the specimens used for this experimental study. At least three specimens were performed into the experiments in each category in order to establish the statistical significance of the results. Figure 1 shows various anastomosed specimens used in the compliance measurement study.

Fig. 1 Specimens used in the compliance measurement studies.

Before synthetic grafts were sutured onto the arterial specimen using Prolene 7.0, they were immersed into fresh non-heparinized canine blood in order to block pores present in the graft walls, and hence, prevent leakage of the fluid during the experimental study. The effect of silicone gel was observed by the compliance measurements in the Penrose surgical drainage tubing and synthetic graft segments.

2. Experimental Method

The experimental set-up, shown in Figure 2, is made up of an air pressure regulator unit, a pressure chamber (chamber Ⅰ), a flow chamber (chamber Ⅱ) and a constant-temperature bath. One end of the chamber Ⅰ is clamped, and the other end is connected to the chamber Ⅱ. To control the temperature during the experiment, a constant-temperature bath is connected to chamber Ⅱ. The regulated air pressurizes the tap water inside the pressure chamber. Kodak Ectakrom ASA 400 black-and-white films were used. Each experiment was performed in the following procedures.

1) Mount the specimen to the chamber Ⅱ, with providing in situ tension.
2) Run the constant-temperature bath to maintain a temperature of 37℃ in the chamber Ⅱ.
3) Initially, take a picture on the specimen without providing any applied pressure.
4) Pressurize the chamber Ⅰ from 50 to 250mmHg in increments of 25mmHg.
5) Take pictures on the specimen at each increment of pressure from the same camera position of step 3).
6) Process the film to digitize the images.

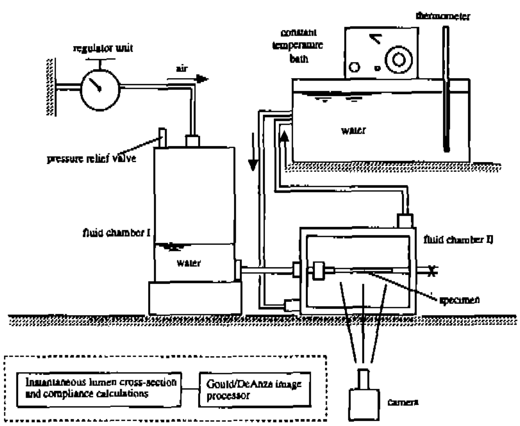


Fig. 2 Experimental set-up of the compliance measurement study.

The images in each film have informations on the instantaneous diameters along the specimen. With a Fortran program in Gould/DeAnza image processor of 512×512 pixels, the digitization were performed with the following steps.

1) Mark four reference points with know length.
2) Mark a line connecting two end points along the center line of the specimen.
3) Put a mark at a point along the suture line or center point.
4) Carefully digitize both wall boundaries of the specimen on the image. Each image was digitized three times in order to reduce digitization errors.
5) Repeat 1) through 4) for images at various transmural pressures.

Instantaneous diameters as a function of transmural pressure and distance from the anastomotic site were computed from the images. Since the coordinate of the image processor given by the pixcels is the integer, the calculation of the instantaneous diameter was required to find the approximate real values in the coordinates. The pixel had an unavoidable error of about 0.03mm in the radial direction for the present study. The data of initial diameter at p=0mmHg of all measurements were aborted in order to prevent the initial effects of the specimen with kinking and others. Instantaneous diameters which were calculated by the conversion of the data from three digitizations for one picture at a given pressure were averaged, then diameter changes of each specimen were calculated at each pressure levels. Instantaneous diameters at the region of silicone gel were recalculated with those at 50mmHg. With these data of all the pressure levels, compliance distribution of the specimen was determined at each pressure level for each experiment, as defined by the following equation.

$$Co=\frac{\Delta d/d}{\Delta p} \tag{1}$$

Where d is the instantaneous diameter at that position for the last pressure step and Δp is the intraluminal pressure. The analysis of vari-ance (ANOVA) was performed to evaluate the statistical difference between the results obtained in the different categories.

RESULTS AND DISCUSSION

Figure 3 shows mean instantaneous diameters of three different categories of specimens. It is noted that Penrose tubing behaved like a linear-elastic material, but PTFE vascular grafts exhibited non-linear characteristics. Figure 4 shows compliance of the Penrose tubing segment and synthetic grafts as a function of transmural pressures. In vascular grafts, the compliance decreased with increasing transmural pressure, while the compliance of Penrose tubing was approximately constant in the physiological range of transmural pressures. Compliance of the Penrose tubing was determined to be of $57.0\pm3.7\times10^{-2}$%/KPa ($7.6\pm0.5\times10^{-2}$%/mmHg) at transmural pressure of

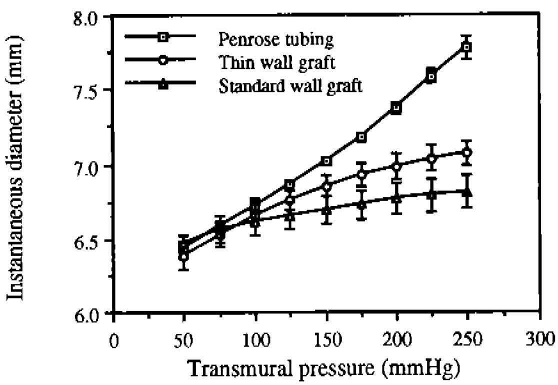


Fig. 3 Mean instantaneous diameters of the Penrose tubing and PTFE vascular graft segments with respect to transmural pressures.

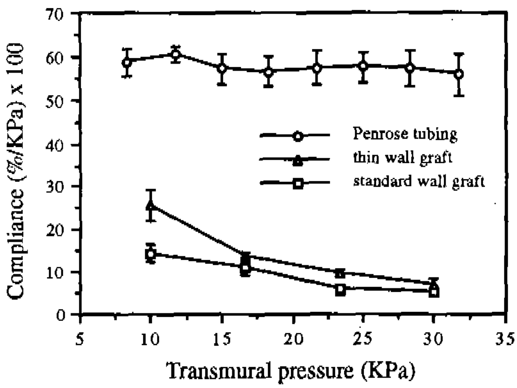


Fig. 4 Compliances as a function of transmural pressure of the Penrose tubing and PTFE vascular grafts.

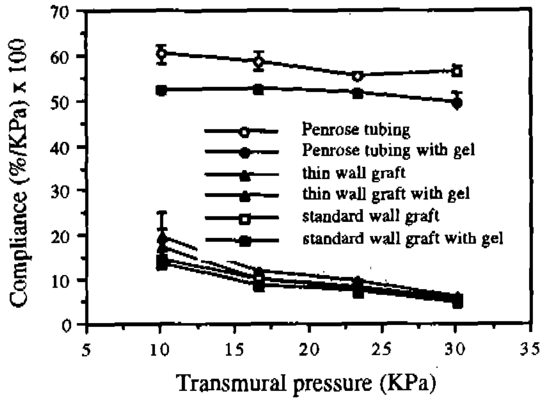


Fig. 5 The effect of silicone gel on the compliance distribution.

15.0KPa (112.5mmHg). Compliances of the standard wall graft and the thin wall graft were about $11.2 \pm 2.2 \times 10^{-2}$%/KPa ($1.5 \pm 0.3 \times 10^{-2}$ %/mmHg) and $13.5 \pm 1.5 \times 10^{-2}$%/KPa ($1.8 \pm 0.2 \times 10^{-2}$%/mmHg) respecitively at 16.7KPa (125mmHg). Previous in vivo studies, as shown in Table 2, revealed that the compliance of the human femoral artery and synthetic grafts are about 67.5×10^{-2}%/KPa (9.0×10^{-2}%/mmHg) and $10.5-14.3 \times 10^{-2}$%/KPa ($1.4-1.9 \times 10^{-2}$% /mmHg) respectively[5,11,12], which showed good agreements with those in the present study. Figure 5 shows the effect of silicone gel on the compliance. It was found that the silicone gel application to prevent leak did not affect the compliance significantly, when compared to the difference in compliance magnitudes between the Penrose tubing and the graft specimen.

Figure 6 (a) shows the compliance distriution across the Penrose tubing-Penrose tubing anastomosis, simulating an artery-artery junction. The compliance of Penrose tubing was fairly uniform about 3mm distal to the anas-

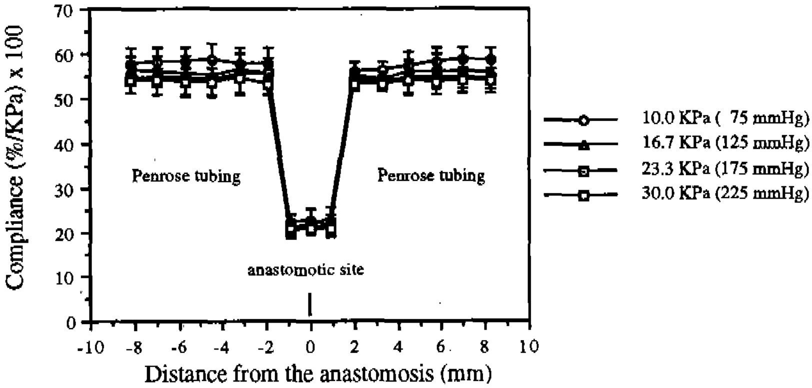
Table 2. Comparison of measured compliance in vitro in the present study with reported in vivo measurements.

Specimens	Compliance		Reference
	(%/KPa)×100	(%/mmHg)×100	
Human femoral artery	44.36±3.76	5.9±0.5	Walden et al.[5][a]
Canine femoral artery	63.16±15.79	8.4±2.1	White et al.[12]
Canine femoral artery	73.68±24.06	9.8±3.2	Stewart and Lyman[11][b]
Penrose tubing	57.0±3.7	7.6±0.5	The present study[c]
Dacron	14.29±2.26	1.9±0.3	Walden et al.[5][a]
PTFE	12.03±1.50	1.6±0.2	Walden et al.[5][a]
PTFE	10.53±4.51	1.4±0.6	White et al.[12]
Standard wall PTFE	11.2±2.2	1.5±0.3	The present study[b]
Thin wall PTFE	13.5±1.5	1.8±0.2	The present study[b]

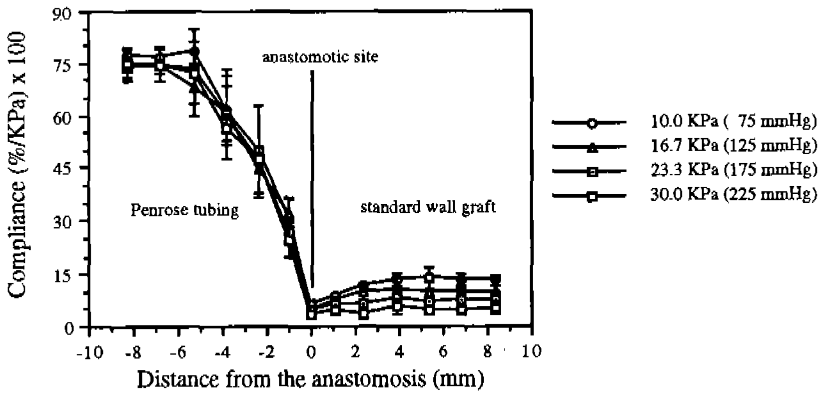
Note : Compliance were measured in the following range of transmural pressures :

a : 6.7−22.7 KPa (50−170mmHg). b : 13.3KPa (100mmHg).

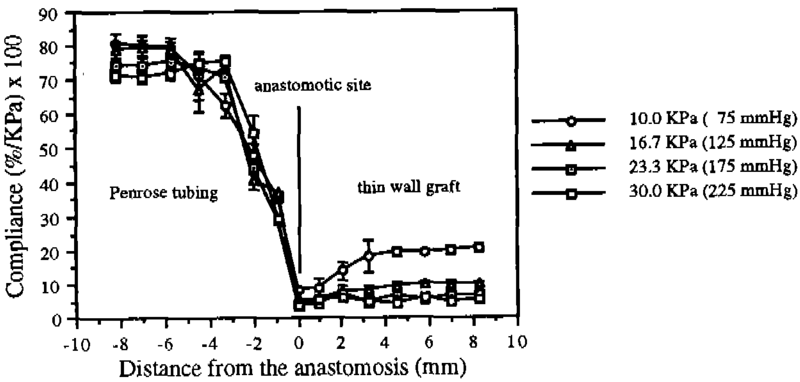
c : 15.0KPa (112.5mmHg). d : 16.7KPa (125mmHg).



(a)



(b)



(c)

Fig. 6 Compliance distribution as a function of distance from the anastomosis.

(a) Penrose tubing-Penrose tubing anastomosis.

(b) Penrose tubing-standard wall PTFE graft anastomosis.

(c) Penrose tubing-thin wall PTFE graft anastomosis.

tomotic site. Figure 6 (b) and (c) show compliance distributions across the Penrose tubing-standard wall PTFE graft anastomosis and the Penrose tubing-thin wall PTFE graft anastomosis respectively. Remarkable compliance mismatches were observed across the anastomotic site. The variations in the lumen cross-sections at the anastomotic site resulted from the effect of compliance mismatch and the suture, which is evident from these figures. With the increase of the intraluminal pressure, the compliance mismatch across the anastomosis seems to be much more serious. When people have arterial disease with endothelial wall thickening, the abnormal increase in the intraluminal pressure occure. Then this compliance mismatch will be much more severe than that in the normal pressure ranges. It can be observed that the cross-sections varied significantly due to increase in transmural pressure and the flow dynamics will be affected by these variations.

The analysis of variance (ANOVA) method was used to distinguish the significant difference in the compliance of each group. This statistics led the result that the compliance of Penrose tubing, thin graft and thick graft are significantly different ($p<0.001$). In other words, it can be said that the compliance of the Penrose tubing, thin graft and thick graft are all different among one another, with which Penrose tubing has the largest compliance, and the thick graft is the smallest one.

Away from the anastomotic site, changes in diameter and the compliance were about the same as the case in the segment studies. However, the results suggested that the suture line created an additional stiffness to the specimen. Even without the suture line, the compliance in the graft side is smaller than that in the Penrose tubing side, thus led to a compliance mismatch at the anastomosis region.

Hyperplasia is consistently greater at the downstream anastomosis as compared with the upstream anastomosis[13,14]. The significant compliance mismatch produces a converging conduit at the proximal end of a graft and a diverging conduit at the distal end with the assumptions of the equal radii at the diastolic pressure. However, at systolic pressure the compliant host artery distends, whereas the less compliant graft relatively remains at the diastolic diameter. Therefore, at the distal anastomosis, there may be a decrease in both flow velocity and volume, leading to local stagnation and increased potential for local thrombosis.

In the present compliance measurement study, there was no flow through the specimens, and static pressures were applied as transmural pressures. In vivo study[9] also suggested that postsurgical stiffening of arteries resulted in reduced compliance mismatch across the anastomosis. Loss of arterial compliance after surgery may be due to periarterial scar formation and changes in the elastic properties of the wall, resulting from possible ischemic injury following dissection, or to a combination of these factors.

The cross-sectional variations at the anastomosis of Penrose tubing (simulating an artery) and PTFE graft specimens at transmural pressures within the physiological range were experimentally determined. From the present study, the following conclusions can be drawn :

1. The PTFE graft exhibited a nonlinear behavior with compliance decreasing with increasing transmural pressure, whereas the compliance of the Penrose tubing remained relatively constant within the range of the pressures in which data were obtained.
2. The lumen cross-sections at the anastomosis were affected by the suture and the difference in compliance between the Penrose

tubing and vascular grafts. The suture line itself presents a regin of compliance mismatch. The variations in the lumen diameter at the anastomoses was more pronounced with increasing transmural pressures. In patients with hypertension, this effect might result in severe abnormal flow dynamics.

From these compliance measurement studies, it was clearly demonstrated that the compliance of the prosthetic grafts is much smaller than that of the host arteries. In addition to the hemodynamic consequences, compliance mismatch across anastomoses has been known to lead to increased anastomotic and suture stresses with resultant suture line dehiscence and false aneurysm formation. The elastical properties of the arterial system are essential to their functions, which include the transmission of pulsatile energy to the periphery, the smoothing of the pulse wave, and the matching of the low impedance output of the heart to the high input impedance of the peripheral circulation[16-19]. Thus, there are good hemodynamic reasons to suppose that introduction of a less compliant arterial graft into the arterial circulation will be damaging and that grafts should be made to match the elastic behavior of their host arteries as closely as possible.

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