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# **A Proposal for Redesigning Aortofemoral Prosthetic Y Graft for Treating Abdominal Aortic Aneurysms**

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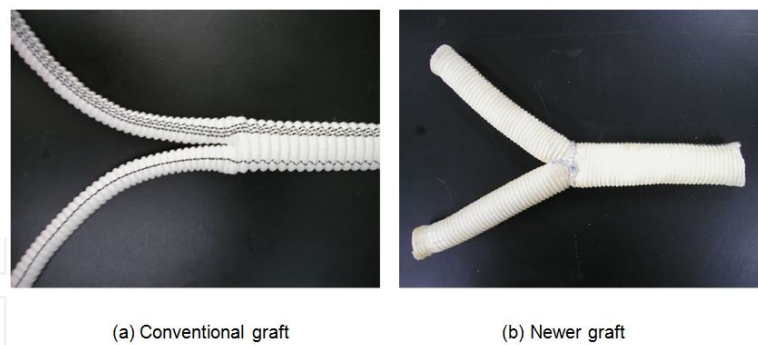
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<http://dx.doi.org/10.5772/52979>

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## **1. Introduction**

Aortofemoral Y grafts are applied commonly in the treatment on those abdominal aortic aneurysms which are at a high risk of rupture. Typically in these Y grafts, the diameter of the stem conduit is 16mm and that of the branch conduit is 8 mm. The ratio of the stem to the branch is different from that of the natural anatomical bifurcation: the branch diameter is small in comparison with that of the natural vessel. It is supposed that the branch diameter is too small to allow these graft implants to work effectively. Consequently, new grafts with a 9-mm branch diameter have been recently implemented in clinical applications as shown in Fig. 1 [1]. However, it has not been discussed fully from the viewpoint of hemodynamics whether this difference of scale at the bifurcation has an actual influence on the blood flow. Many prosthetic devices, including the Y grafts that are discussed here, are currently being used in clinical applications. Although some of these devices should be redesigned in order to improve their properties, methodologies for redesigning them have not been established. In this study, Y grafts have been selected as a subject of study; however, it is not that a completely newly-designed product is introduced, but that a new concept, by which a new product can be developed or redesigned, is presented. At first, the effect of bifurcation on pressure loss in the Y graft is explored by experiments conducted under conditions of steady flow. This is done in order to understand basically the characteristics of bifurcation flow, such as that which is found in a Y graft. Secondly, additional experiments were conducted in order to demonstrate the effects of an incremental increase in branch diameter in a newly designed aortofemoral Y graft under conditions of pulsatile flow.



**Figure 1.** Photographs of aortofemoral Y grafts

## 2. Effect of bifurcation on pressure loss in aortofemoral prosthetic Y grafts

### 2.1. Purpose of steady flow experiments

The flow of fluid such as blood in a circular tube with a constant inner diameter causes pressure to drop due to energy loss. If there is a bifurcation in the circular tube, generally change in the flow causes additional energy loss. It is important to understand how the bifurcation has an influence on the flow from the viewpoint of hydrodynamics in order to design Y grafts. Therefore, the experiments were conducted to attempt to indicate the effect of bifurcation on flow under a steady flow condition.

### 2.2. Method of steady flow experiments

#### 2.2.1. Bifurcation model

Three types of fluid models of rigid aortofemoral Y grafts were made as shown in Fig. 2 [1][2][3]. Epoxy was used as the material of the models that were produced by the lost wax method.

First of all, the mold, whose outer shape replicates the inner shape of the Y graft, is made from wax. The wax mold is immersed in melted liquid-state epoxy which solidifies in several hours. Secondly, after the material becomes a solid epoxy block, it is heated up and the wax in the block runs out. Finally, from the above process, the epoxy block, in which the flow channel that replicates the inner shape of the Y graft is formed, with a smooth inner surface, becomes the fluid model of the Y graft for experiments.

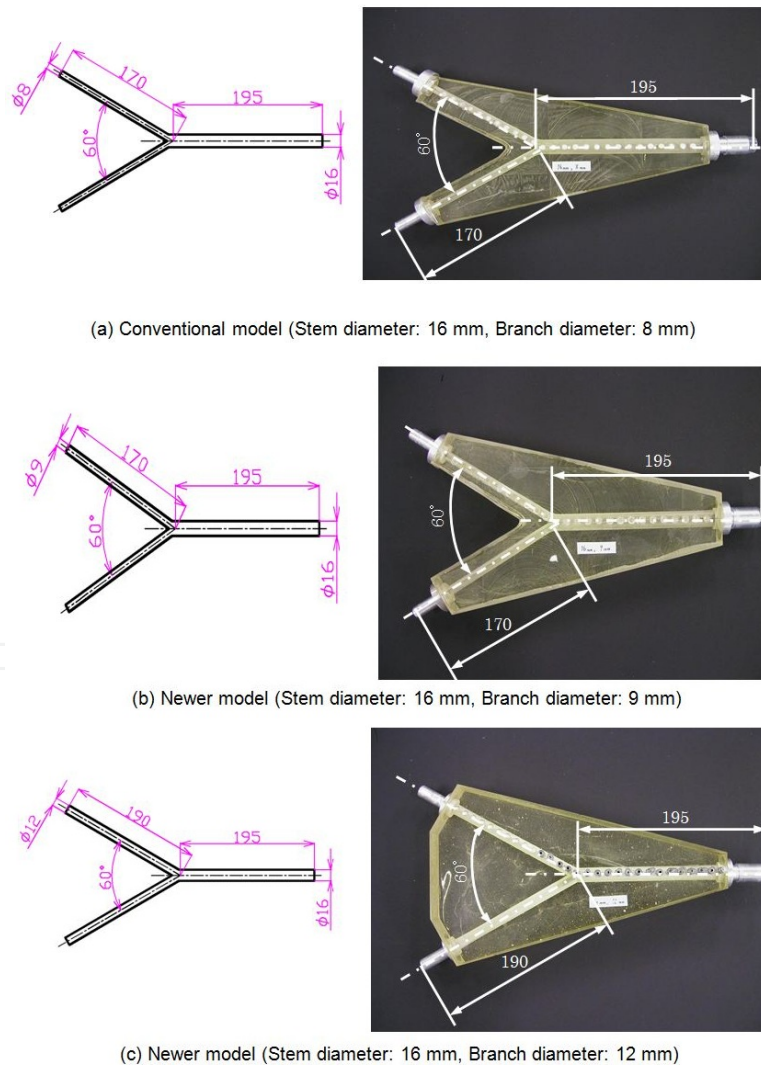
There are an inlet and two outlets in the bifurcation models. Correspondingly, the conduit of the inlet end is called the stem and the conduits of the outlet ends are called the branches. The first model, in which the diameter of the stem is 16 mm and that of the branches is 8 mm, replicates conventional grafts. In the second model, which replicates the newer graft, the branch diameter is 9 mm and the stem diameter is the same as that of the conventional

graft. In the third model, which replicates yet another style of newer graft, the branch diameter is 12 mm and the stem diameter is the same as those of the other two models.

In all models, the structure is symmetrical along the long axis of the stem conduit, the angle between the branch conduits is 60 degrees, and the lengths of the stem and branch conduits are 195 mm and 170 or 190 mm respectively. In order to measure the pressure in the flow, the ports are set in the models along the flow channels.

### 2.2.2. Test circuit for steady flow experiments

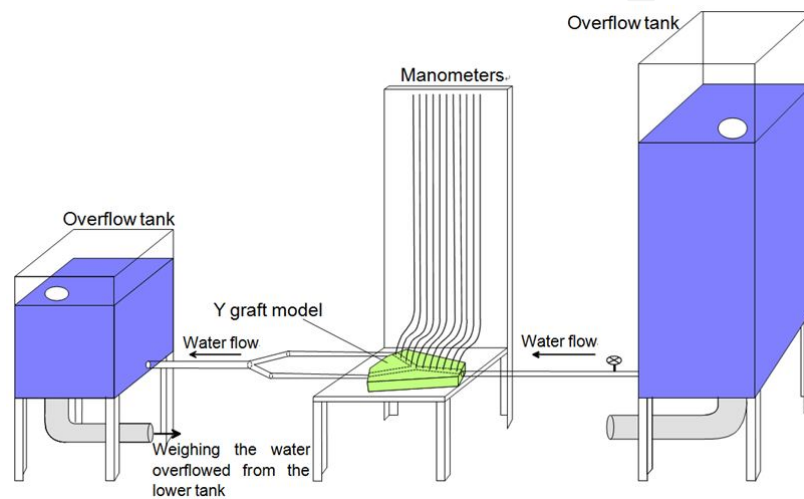
A fluid test circuit was set up in order to conduct experiments for these three bifurcation models as shown in Fig. 3 [1]. The circuit consists of a higher overflow tank, a lower overflow tank, a valve for regulating flow rate, and manometers for measuring pressure in these bifurcation models.



**Figure 2.** Three types of models of rigid aortofemoral Y grafts

Each model is installed between the higher and the lower overflow tank in the test circuit, in which tap water is used as the fluid. The height of the water in the manometer is called water head, which indicates static pressure in the flow channel. The higher tank loads the inlet of the bifurcation model with constant 1000 mm water head, or pressure, through the regulating valve and the lower tank loads the outlets of the model with constant 300 mm water head. The fluid in the circuit is driven by the constant pressure gradient between these two tanks and thus has a condition of steady flow.

The manometers are connected to the ports on the model to measure the pressures of the flow, and the flow rate is measured by weighing the water overflowed from the lower tank.



**Figure 3.** Test circuit used for steady flow study

### 2.3. Effect of bifurcation under condition of steady flow

The results from the conducted experiments were analyzed to explore the relationships between pressure and flow rate in the bifurcation model under the condition of steady flow.

Energy loss due to fluid flow through the bifurcation causes a reduction in the pressure of the flow (that is, the pressure gradient along the flow stream). At first, it is simply confirmed how the pressure gradient between the inlet and the outlet of the bifurcation model changes when the flow rate increases. The flow rate is shown here in the form of a Reynolds number as a dimensionless parameter to enable discussion of these flow properties generally.

In Fig. 4, it is shown that the pressure gradients change to Reynolds numbers that range from 2000 to 5000 [1][2][3]. These Reynolds numbers might be supposed to be large in comparison with those in the natural abdominal aorta, but these numbers were settled in order to figure out the characteristics of the bifurcation flow from the view point of hydrodynamics rather than the flow of the natural aorta.

Pressure gradients increase as Reynolds number increases in all bifurcation models; however, changes in these pressure gradients differ in degree among these models. An increase in

the branch diameter is accompanied by a decrease in the pressure gradient. In fact, the pressure gradient between the inlet and the outlets was only 1 mmHg in the case of a 12 mm branch, which was 4 mmHg less than that of the 8 mm branch at Reynolds number 5000. As mentioned above, the effect, that an increase in the branch diameter causes a decrease in the pressure gradient, grows in accord with increase in the Reynolds number.

The pressure gradient discussed above is caused by flow resistance in the stem conduit, the branch conduit, and the bifurcation itself. In other words, summation of the energy loss in the fluid flow at the stem conduit, the branch conduit, and the bifurcation is estimated as the pressure gradient. Next, it is important to discuss whether the difference in the structure of the bifurcation has an essential influence on the pressure gradient.

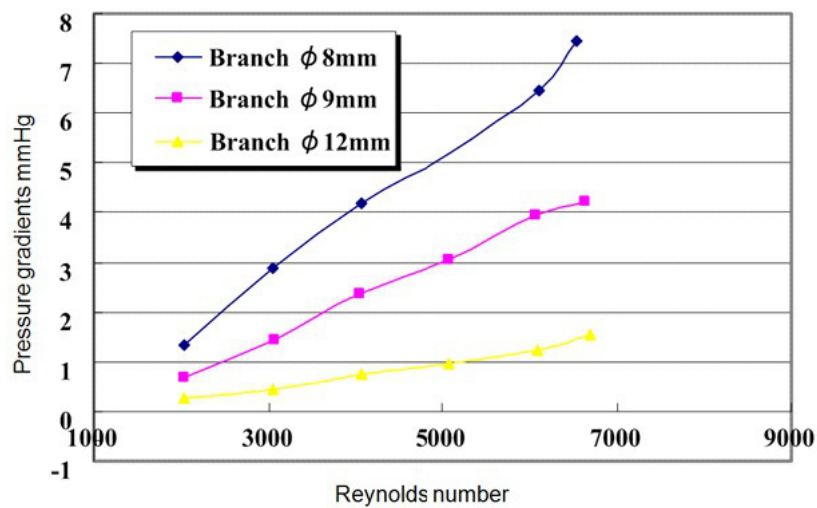


Figure 4. Pressure gradients at each Reynolds number

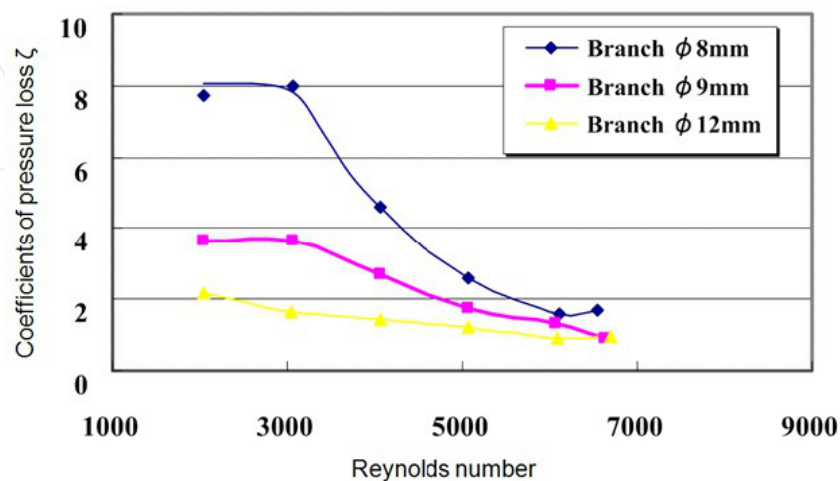


Figure 5. Coefficients of pressure loss at each Reynolds number

In order to evaluate the effect of the bifurcation not including the stem and the branch conduits, we focus on the flow at the bifurcation in itself.

When it is supposed that the pressure loss due to the bifurcation is proportional to the kinetic energy of the flow through it, the pressure loss  $\Delta P$  is presented as

$$\Delta P = \zeta \rho V^2 / 2 \quad (1)$$

where  $\zeta$  is the coefficient of pressure loss,  $\rho$  is density of the fluid, and  $V$  is the flow velocity.

From the measured pressures in the stem and the branch conduits, the pressure loss  $\Delta P$  due to the bifurcation in itself is calculated. Furthermore, the coefficient of pressure loss  $\zeta$ , by which the flow resistance can be compared among the different bifurcations, is calculated from the above formula based on the flow rate measurements.

In Fig. 5, the relations of the calculated coefficient of pressure loss  $\zeta$  to the Reynolds number are shown. In the 12 mm branch, the coefficient of pressure loss due to bifurcation is less than half of that in the 8 mm branch.

Summing up the results of these experiments, in which variation in the branch diameter in the bifurcation model was evaluated from the standpoint of hydrodynamics in epoxy-based models, it was revealed that in the bifurcations an incremental increase of only 4 mm in branch diameter affects hydrodynamic characteristics drastically under steady flow conditions.

### 3. Characteristics of a newly designed aortofemoral prosthetic Y graft under pulsatile flow conditions

#### 3.1. Purpose of pulsatile flow experiments

In the experiments on steady flow mentioned above, flow in bifurcations was studied. The results showed that an increase in the branch diameter of the bifurcation is accompanied by a decrease in the pressure gradient and that the size of the bifurcation has an essential influence on the flow through it. From these results, it might be suggested that a larger conduit branch for a Y graft would be more efficient, because there is a potential for an increase in downstream flow from implanted Y grafts. However, there are many differences between the experimental conditions in the test circuit and those in the natural aorta, of which the most notable is the condition of the flow, which was steady in the experiment but is pulsatile in the aorta. In pulsatile flow, other effects not confirmed in the steady flow experiments may occur. Thus, it is not clear whether the results obtained under the condition of steady flow can be applied directly to the natural aorta.

Additional study is necessary to evaluate the effects of the new graft model in the setting of the natural aorta, and there are several possible methods for this study. For instance, experiments either in an animal or in a mock circuit simulating the natural blood circulation are

employed typically. In the former, natural pulsatile flow conditions can be obtained, but it is difficult to measure stable hemodynamic conditions repeatedly. However, in the latter, it is not difficult to measure pressure and flow, but it is necessary to confirm whether it actually simulates the natural setting.

In this study, a mock circulatory system was set up to conduct experiments under pulsatile flow condition.

The mock circulatory system, in which bifurcation models replicating Y grafts are installed and fluid is fed through the grafts, simulates the left ventricle, the aorta, and the peripheral vessels. The structures of the bifurcation models and the mock system, as well as the reason that the mock system can be used to mimic hemodynamics in the aorta, are explained first. After that, the results obtained from the experiments are described.

### **3.2. Method of pulsatile flow experiments**

#### *3.2.1. Bifurcation models used for experiments*

In the pulsatile flow experiments, three models of rigid aortofemoral Y grafts, essentially the same as those used in the experiments of steady flow, were used as shown in Fig. 2. The type of epoxy, model material, and methods used in this experiment were the same as those in the steady flow experiments.

In all models, the diameter of the stem conduit is 16 mm. The diameters of the branch conduits in the first, the second and the third model are 8, 9, and 12 mm respectively; the first model corresponds to a conventional Y graft, and the second and the third models to newer grafts. The length of each conduit and the angle between the branch conduits are the same as those of the models used in the steady flow experiments.

#### *3.2.2. Mock circulatory system for pulsatile flow experiments*

A mock circulatory system was constructed as shown in Fig. 6 to conduct experiments under conditions similar to those in the natural blood circulatory system [1][4][5]. The mock system consisted of a pulsatile pump, valves for flow resistance, compliance units, and overflow tanks.

The pulsatile pump, which was driven by a pulse motor, had an inlet and an outlet port. The inlet port and the outlet port were connected to an overflow tank and a compliance unit respectively. The outlet port was also connected to another downstream-side overflow tank in order to regulate pressure and flow rate. The pump, which functionally simulated the left ventricle, could output as much fluid volume as was set preliminarily and could create given flow patterns.

Three valves were installed upstream from the overflow tanks to regulate flow resistance. When the resistance was increased by the valve, the flow rate into the overflow tank decreased. Thus, an increase in the resistance corresponded to constriction of vessels in the blood circulatory system. On the other hand, a decrease in the resistance corresponded to

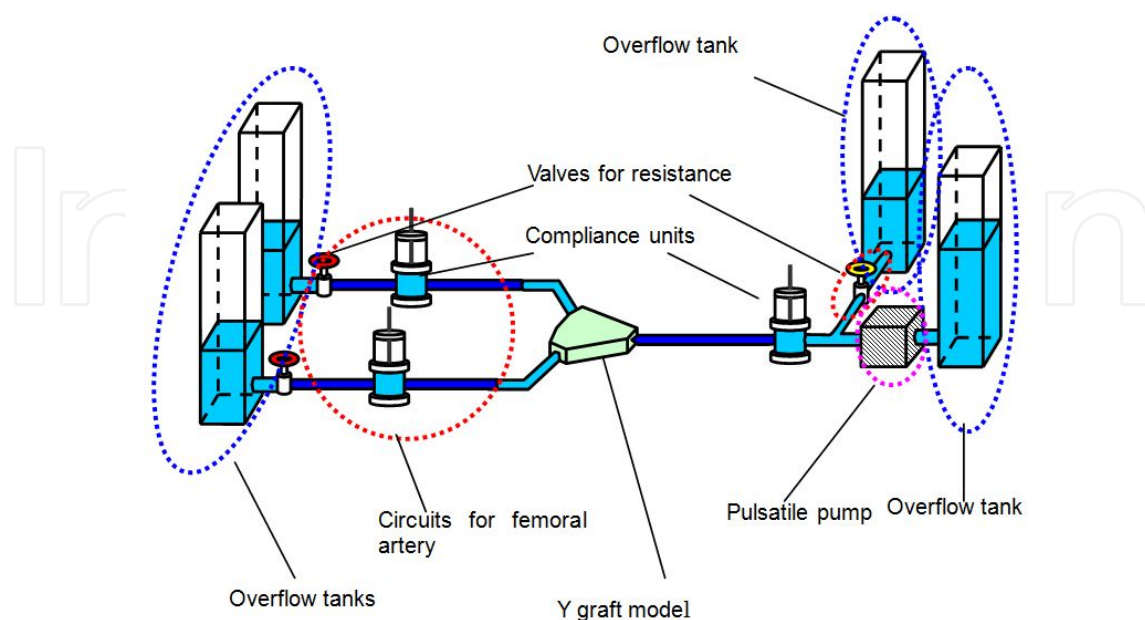


dilatation of blood vessels. By operating these valves, changes in flow resistance of the mock system represented changes in the diameters of actual blood vessels.

Three compliance units were installed in the mock system. Each compliance unit had a casing made from acrylic into which an elastic membrane and a spring were built. Pressure in the fluid was changed through contact with the membrane, which in turn was pressed by the spring. Owing to the restoring force of the spring built in the unit, the unit achieved the effect changing pressure in accordance with changes in the fluid volume. As fluid volume in the unit increased, so did the restoring force of the spring, and thus pressure intensified. In the contrasting situation, an increment in pressure attenuated in the case of a decrease in the restoring force of the spring. Blood vessels in the natural circulatory system are supposed to consist of elements of resistance and compliance from the viewpoint of dynamics. The element of compliance is caused mainly by the elasticity of the vessel wall. A stronger restoring force in the spring indicated higher elasticity, comparatively. On the other hand, a weaker force indicated that the elasticity was lower. Therefore, through inclusion of the compliance unit, the mock system could simulate the elasticity of the vessel wall.

By combination of the valve as a resistance element and the compliance unit as an elasticity element, input impedance of the vessel could be regulated, as will be discussed later.

Each bifurcation model was installed in the mock system and physiological saline was fed through the fluid circuit. Pressure and flow rate at the inlet and the outlet of each model were measured under conditions of pulsatile flow by pressure transducers and electromagnetic flow meters.



**Figure 6.** Mock circulatory system for pulsatile flow experiments

### 3.2.3. Experimental conditions

Experiments were conducted in the mock system described above to evaluate the newer Y graft under conditions of pulsatile flow. However, the question of whether or not the constructed mock system simulates effectively the natural blood circulatory system must be answered. How can the question be answered? In steady flow in a circular tube, pressure and flow rate are constant with time. Therefore, a property of flow is presented simply by a ratio of pressure divided by flow rate or velocity as a flow resistance.

In pulsatile flow such as blood flow in the artery, a property of flow is indicated generally as the input impedance instead of as the flow resistance as mentioned above [6][7]. It is difficult to evaluate the characteristics of pulsatile flow, in which pressure and flow rate change constantly with time; however, pressure and flow rate are periodic signals that are generally represented as an expanded series.

Next, a basic explanation about the input impedance is shown. Pressure  $P(t)$  and flow rate  $Q(t)$  are expanded in Fourier series as follows [5][8][9],

$$P(t) = P_0 + \sum_{n=1}^{\infty} P_n \cos(2\pi n f_1 t - \alpha_n) \quad (2)$$

$$Q(t) = Q_0 + \sum_{n=1}^{\infty} Q_n \cos(2\pi n f_1 t - \beta_n) \quad (3)$$

where  $P_0$  and  $Q_0$  are time averages of  $P(t)$  and  $Q(t)$ ,  $P_n$  and  $Q_n$  are amplitudes of n-order harmonic,  $f_1$  is fundamental frequency, and  $\alpha_n$  and  $\beta_n$  are phases of n-order harmonic.

Input impedance is represented as modulus  $Z_n$  and phase  $\Psi_n$  as follows,

$$Z_n = \frac{P_n}{Q_n} \quad (4)$$

$$\psi_n = \beta_n - \alpha_n \quad (5)$$

We conducted an animal experiment, in which a goat weighing 51 kg was used, in order to estimate the input impedance at the femoral artery [5]. The input impedance was calculated by the above process from the measured pressure and flow rate in the femoral artery of the animal. Some examples of the results are shown in Figs. 7 and 8. In Fig. 7 (a) and (b), the modulus and the phase are plotted corresponding to frequencies under a control condition. An increase in the frequency is accompanied by a decrease in the modulus and the phase changes from negative to positive with an increase in the frequency. They are typical changes in the input impedance of the artery. In Fig. 8, changes in the input impedance are

shown when a vasoconstrictor was injected into the experimental animal. The tendency is same in both cases; however, the modulus in the case of vasoconstriction is higher than that in the control condition. This indicates that change in the vessel can be represented quantitatively. The modulus at a frequency of 0 Hz indicates an average resistance of the artery system and is an especially important value. The value of the modulus increases from  $1.0 \times 10^4$  to  $3.4 \times 10^4$  dyne·sec/cm<sup>3</sup> when the vessel is constricted.

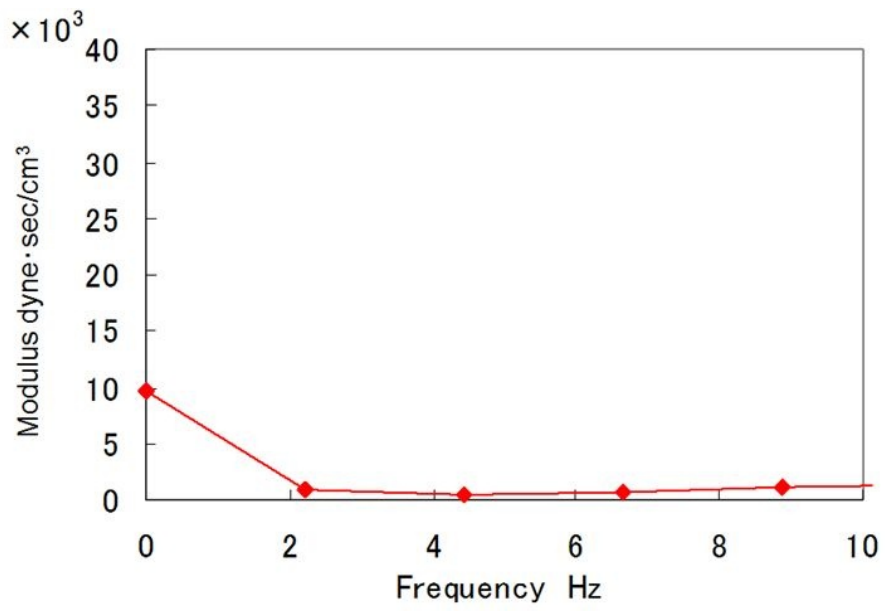
We assumed that a mock system can simulate the natural system in terms of evaluating the Y grafts from the view point of hydrodynamics, when the modulus of the mock system at the femoral artery approximates that of the natural system, because the branch conduits of a Y graft are connected to the femoral arteries.

In the mock system, by regulating the valves as resistance elements and the compliance units as elasticity elements, the mean flow rate at the inlet of the bifurcation model could be made to be 1.0 liter/min with the conditions of the input impedance in the femoral artery being  $1.0 \times 10^4$  and  $3.4 \times 10^4$  dyne·sec/cm<sup>3</sup> at 0 Hz as shown in Fig. 9. Each modulus obtained from the mock system was similar to those from the animal experiment.

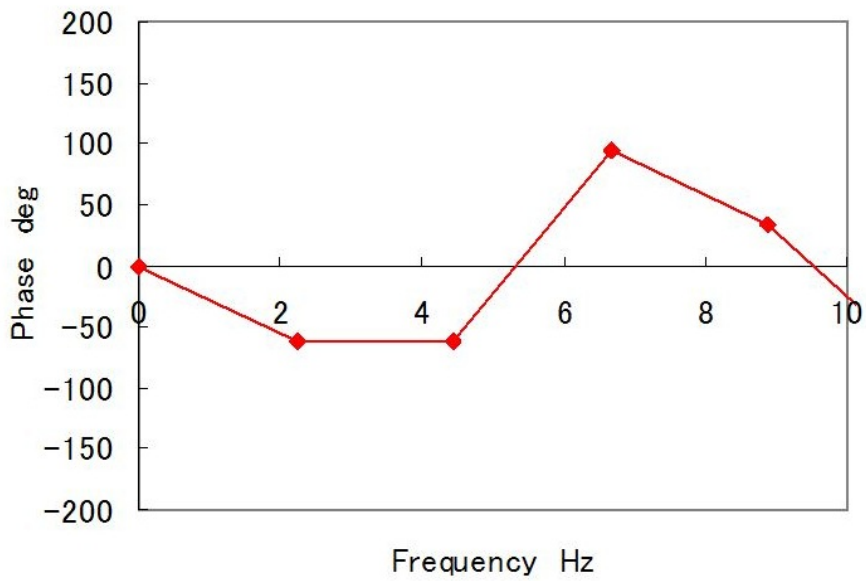
### 3.3. Characteristics of newly designed Y graft under condition of pulsatile flow

From the above discussion, the efficacy of the newly designed Y grafts, that is those with an increased diameter of the branch conduit, can be evaluated in the mock system, in which the input impedance in the femoral artery approximates that in the natural system. To do this, it was important to set suitable conditions by regulating the valves and the compliance units. In the experiments, suitable conditions were those in which the mean flow rate at the inlet of the bifurcation model was 1.0 liter/min and the input impedance of the femoral artery was  $1.0 \times 10^4$  dyne·sec/cm<sup>3</sup> at 0 Hz in the mock system as above mentioned.

The results of experiments obtained under these conditions are shown in Fig. 10 [1][5][10]. The pressure and the flow rate were measured at the inlet of the bifurcation model by a pressure transducer and an electromagnetic flow meter. The mean pressures were equivalent to 66 mmHg in all models, but the flow rates changed due to the diameter of the branch conduit, as shown in Fig. 10 (a) and (b). The mean flow rate was 1.0 l/min in both the 8 mm and the 9 mm branch. However, in the 12mm branch, the mean flow rate was increased by 0.2 liter/min under the same conditions. An increase in the branch diameter caused an increase in the flow through the branch even if the input impedance in the femoral artery was the same across different branch diameters.

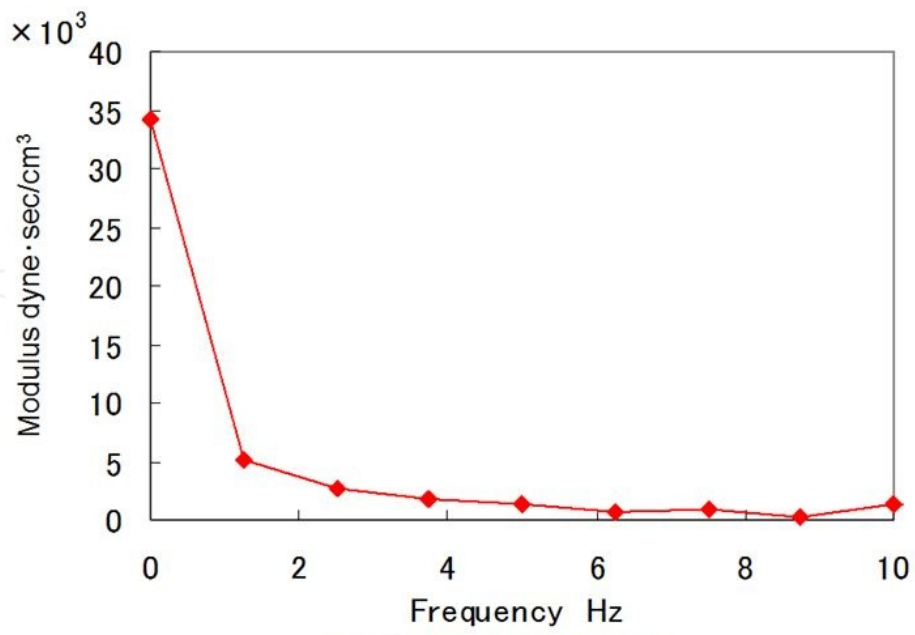


(a) Modulus of input impedance

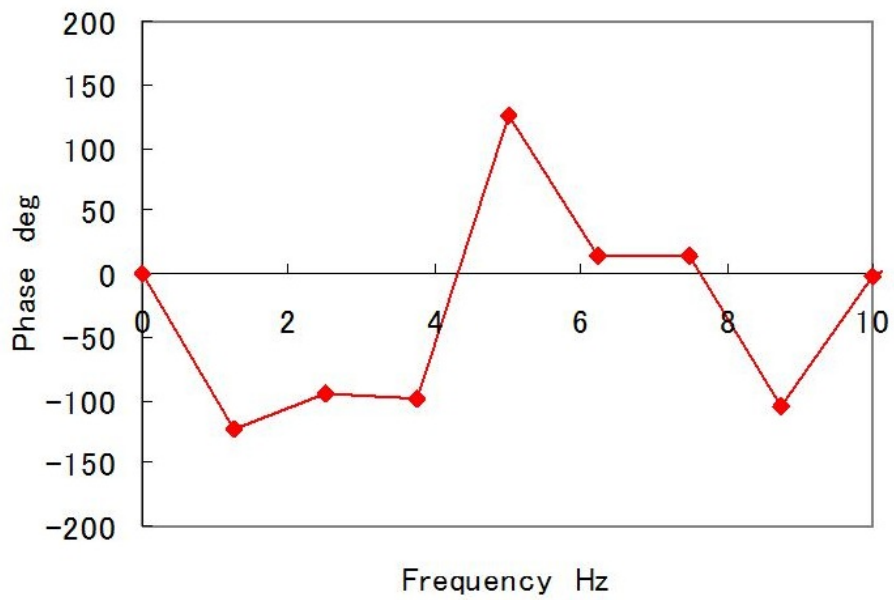


(b) Phase of input impedance

**Figure 7.** Input impedance in the femoral artery obtained in an animal experiment under control condition

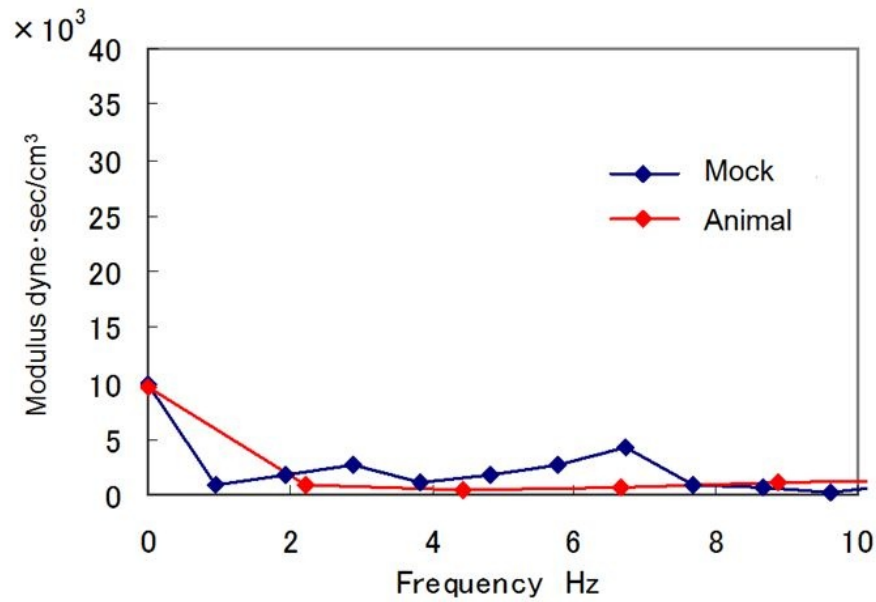


(a) Modulus of input impedance

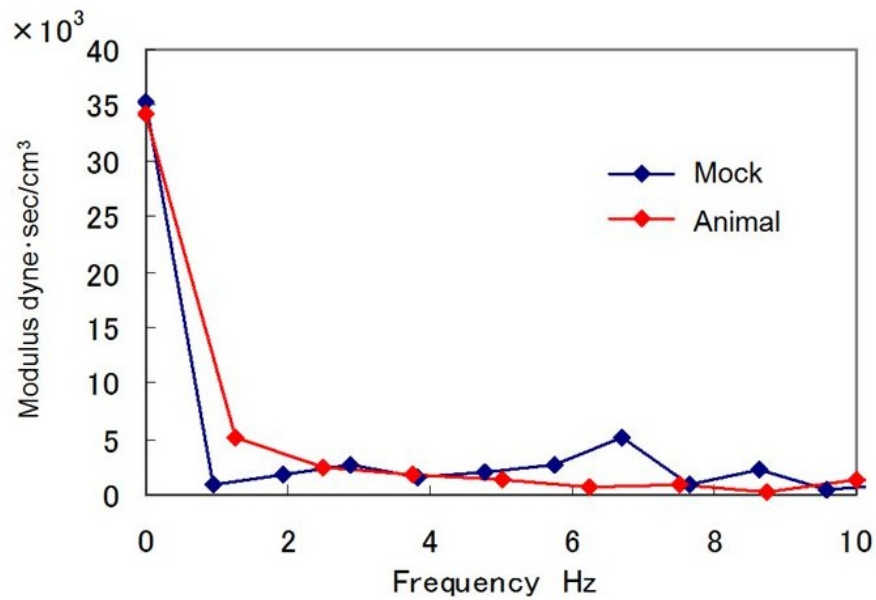


(b) Phase of input impedance

**Figure 8.** Input impedance in the femoral artery obtained in an animal experiment under vasoconstrictive condition



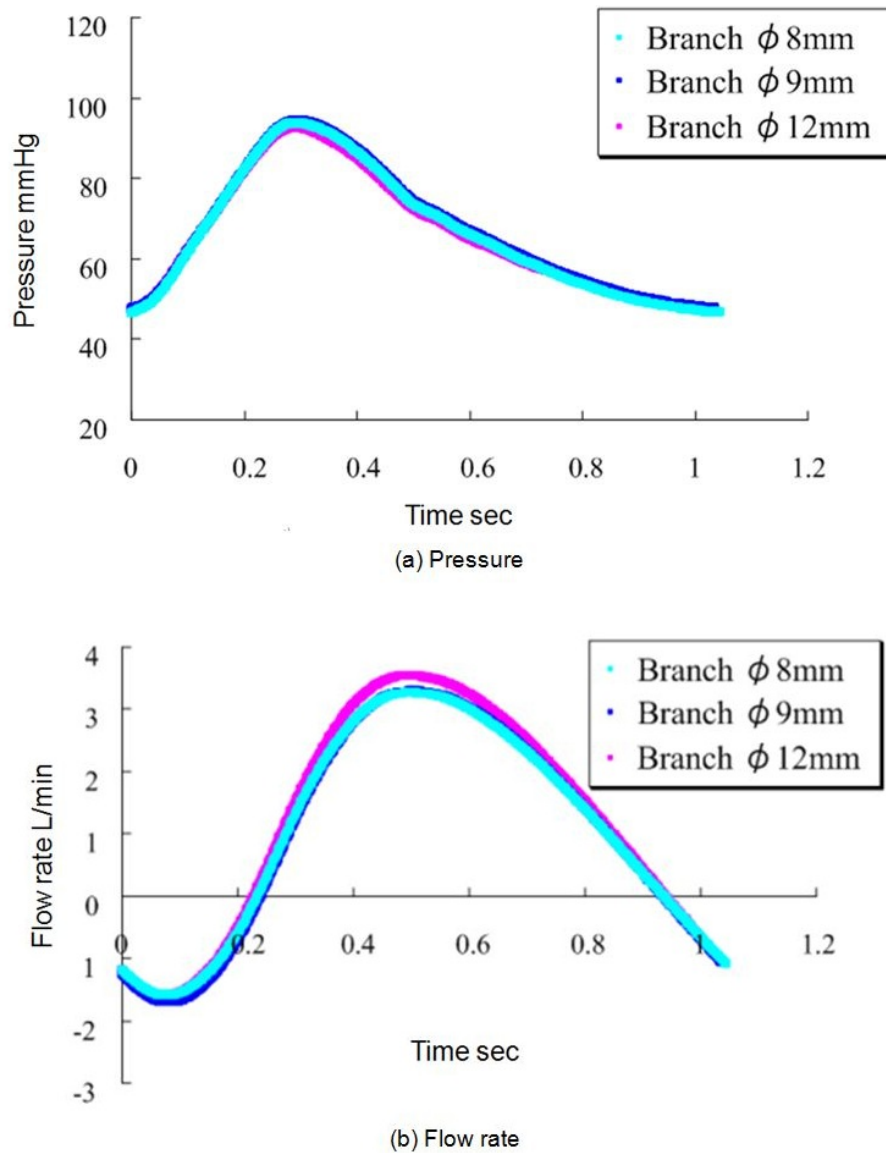
(a) Modulus obtained under control condition



(b) Modulus obtained under vasoconstrictive condition

**Figure 9.** Comparison of input impedance in the femoral artery obtained in the mock circulatory system with that in an animal experiment. Mock and Animal indicate experimental results in the mock circulatory system and the experimental animal respectively.

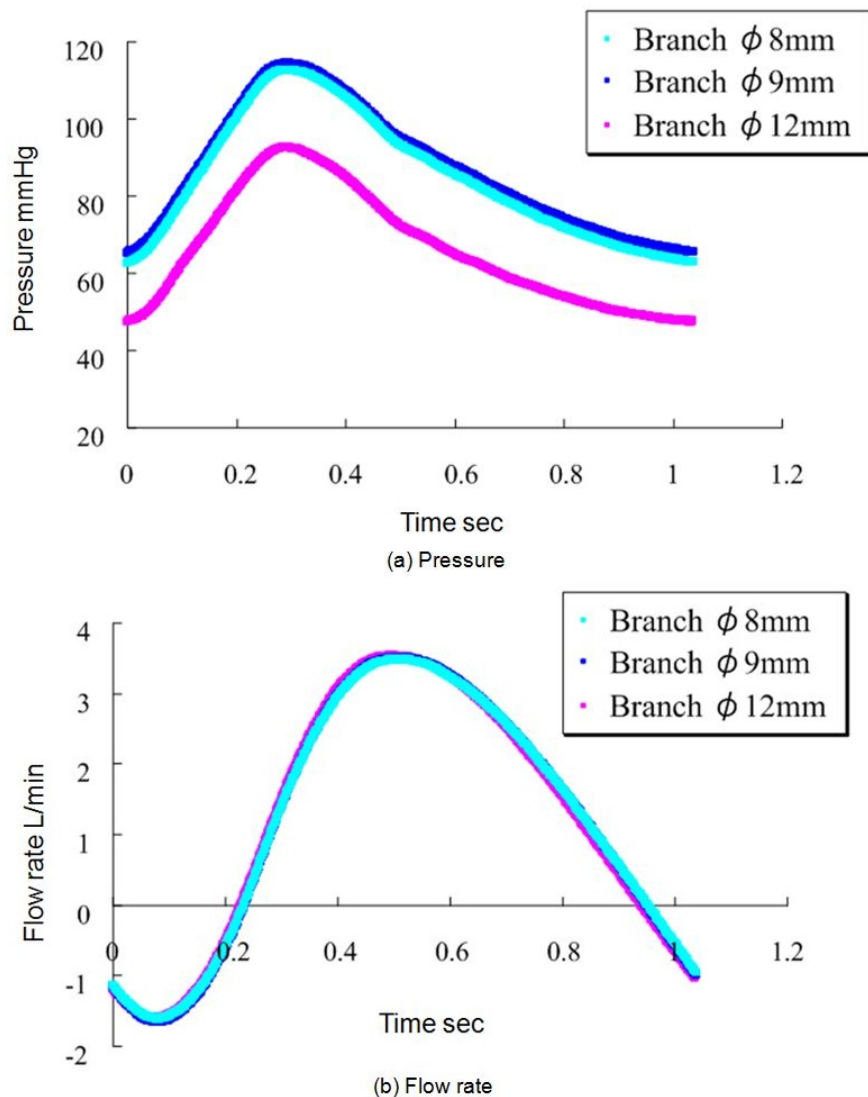
When the mean flow rates were equivalent to 1.2 l/min across all models, the pressures changed due to the diameter of the branch conduit as shown in Fig. 11 (a) and (b). The mean pressure was 85 mmHg in the 8 mm branch; however, the mean pressure decreased by 19 mmHg in the 12mm branch under the same conditions.



**Figure 10.** Results of pulsatile flow experiments under condition of constant mean pressure

In addition, when the input impedance was increased up to  $3.0 \times 10^4$  dyne·sec/cm<sup>3</sup> at 0 Hz by regulating the resistant unit in the mock circuit, the results were found to be the same as those in which the impedance was  $1.0 \times 10^4$  dyne·sec/cm<sup>3</sup> [5].

Summing up the results of these experiments, a newly designed aortofemoral Y graft was compared with a conventional graft from the standpoint of hydrodynamics using epoxy-based models under pulsatile flow conditions. As a result, it was revealed that an incremental increase in the branch diameter increases definitely the flow rate through the graft.



**Figure 11.** Results of pulsatile flow experiments under condition of constant mean flow rate

#### 4. Conclusion

In aortofemoral Y grafts for treating abdominal aortic aneurysms, the ratio of the stem conduit to the branch conduit is different from that in the native bifurcation anatomically. Because the branch diameter is small compared with that of the natural vessel, it is supposed that the difference of scale at the bifurcation in Y grafts has an influence on the blood flow. Thus, variation in the branch diameter in aortofemoral Y grafts was discussed from the standpoint of hydrodynamics.

First, a basic evaluation of the effect of variation in the branch diameter in the bifurcation was executed using epoxy-based models. As a result, it was revealed that in the bifurcations



an incremental increase of only 4 mm in branch diameter affects hydrodynamic characteristics drastically under steady flow conditions.

Secondly, using epoxy-based models, a newly designed Y graft with 12 mm branches was compared with a conventional graft with an 8 mm branch under pulsatile flow conditions in a mock circulatory system that simulated the natural system. The result revealed that an increase in the branch diameter undoubtedly increases the flow rate through the graft even under the same input impedance in the femoral artery.

These results are suggested to be useful for a proposal concerning the redesign of aortofemoral prosthetic Y graft for treating abdominal aortic aneurysms. In this study, Y grafts are selected as a subject of study. However, it is not that a completely newly-designed product is introduced so much as a new concept, by which a new product can be developed or redesigned, is presented.

## Author details

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