Conditioned Viscoelastic-Chracteristics of Human Aorta

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Abstract

Human aorta has viscoelastic behavior. The test of tissues such as aorta, skin, muscle, and etc. is required to consider visco effect on deformation behavior. Creep and slow recovery are main aspects of viscoelasticity of tissue engineering. Volumatric strain plays a important role in determine slow recovery of human arota. This study is to suggest the method avoiding viscous effect in tissue experiment. The results shows the time scale when the specimen can be fully recovered from slow deformation. Also, this study observes the qualitative creep-effect on elastic strain in 1 minute at the same loading.

1. Introduction

Human aorta is chatracterized to elastic, nonlinear behavior. When it is pressurized , it deform elastically. Also, its behavior with loading can not explain linear manner (Hookean material). It shows large deformation at low loading and small deformation at high loading. Specially, threshold between low and high loading is 80mmHg in human aorta. However, its behavior is more comlicate with visco effect (time effect). It is deformed by loading and recovered instantaneously by unloading. The recovery of human aorta is not completed as much as the one of pure elastic material. It takes time to recover the original configuration such as unloaded state. Another visco effect in human aorta is hysteresis. The recovery path at unloading is different from the deformation path at loading. When a aorta is tested, the deformation-behavior with same loading condition depends on experimental trials. Precondition is repeat loading condition to give a specimen. This conditioning can prevent hysteresis. The gap between loading and unloading behavior of deformation is narrowed down to be considered as a one line of deformation-loading relation. However, precondition is only for hysteresis prevention but creep and slow recovery which are also visco effect. They ramains in question for testing human aorta such as soft tissue. Therefore, this study is to observe creep in limited time interval and slow recovery in terms of volumatric strain.

2. Method

Measurement system in this study was the CCD camera system by computer-controlled motor maneuvering (Whang, et al., 1995). Cut ring of human aorta was used for characterizing viscous effect in tissue mechanics. The mesurement system setup in this study was aimed on the cross-section of human aorta (Whang, et al, 1994). The specimen was initially preconditioned so that it was repeatedly loaded by a mount of physiological condition (80-120 mmHg). The silicon carbite particles (50μm) were scattered on the cross-sectional area of the specimen. Then, the circumferential-deformation image was captured at 360 mmHg of loading. The measurement images included initial deformation (instantaneous elastic deformation) and the further deformation with time at the same loading. The respective measurements started at 0, 1, and 11 minutes. For volumetric strain, the measurement system is viewed at side of the ring specimen. The captured image of side view was rectangular. The separating measurement was done by dividing the cylindrical specimen circumferentially into 8 sectors each subtending angle of $\pi/4$ radian. The measurements were a total of 8 longitudinal images at constant loading (120 mmHg). The time scale was three hour and the measuring points were 2, 20, 60, 120, and 180minute.

3. Analysis

For creep determination, the selected particles were tracked at loading and unloading by using particle images which were grabbed during the experiment. The partilce coordinations are determined from the measured images The coordintations determined the circumferential variations (Whang, et al., 1995). The radius changes were observed at unloading and at loading for 11 minute so that instantaneous deformations and deformation progress with time were determined.

The volumatric strains were determined by longitudinal rectangular images (wall thickness x length) captured using a side view camera. Figure 1 is the schematic view of how to determine sector volume. The longitudinal area of each sector is determined by using the x z coordinates of 7 randomly selected boundary points with 4 corners in the longitudinal rectangular image. Radii in the top and the bottom are measured for each inner and outer longitudinal length for each sector. Then they are averaged to determine the centroidal radius of the rectangular area. For each sector, the volume (v) is determined by multiplying the centroidal radius (R) by the longitudinal area (A) and sector angle (Beyer, 1984). Therefore, sector volume is

$$v=\int r\,d\theta drdz$$

where (r, θ) is cylindrical coordinate.
Since

$$\int rdrdz=\int rdA=RA$$

Therefore

$$v=\int RAd\theta$$

However, the centroidal radius R and area A are not constant with circumferential angle . These are assumed to be linearly related to θ within each sector. The volume at each sector is determined by integrating a linear function of area $A_i(\theta)$ times a linear function of centroidal radius $R_i(\theta)$ over the sector angle($\pi/4$) where i is the number of sector (i=1...8). Therefore, total volume(V) is,

$$\begin{aligned} V &= \int_0^{2\pi} A(\theta)R(\theta)d\theta \\ &= \int_0^{\pi/4} A_1(\theta)R_1(\theta)d\theta + \int_{\pi/4}^{\pi/2} A_2(\theta)R_2(\theta)d\theta \\ &\quad \ldots + \int_{7\pi/4}^{2\pi} A_8(\theta)R_8(\theta)d\theta \end{aligned}$$

4. Results

It is well known that the effect of hysteresis is remarkably reduced by preconditioning. The preconditioning process loads and unloads the specimen and tube repeatedly in a given range. By repeated cycling of loading and unloading, eventually a steady state is reached at which further changes are negligible in the stress-strain relation for loading and unloading (Fung, 1981). In this study, the tube is preconditioned in the pressure range of 0-360 mmHg before pressurizing the specimen. Preconditioning of the specimen is then performed 5-10 times in the pressure range of 0-120 mmHg lumen pressure.

Creep for 1 minute at static loading is measured. A particle is selected near the inner layer of the specimen cross-section where maximum radial deformation occurs. The radial movement of the particle is measured at lumen pressure of 120 mmHg for 1 minute after initial rapid movement. This additional radial movement is 0.04 mm out of a total deformation of 5.04 mm. The resulting radial creep deformation is less than 1%, and specimen creep for 1 minute is negligible.

Slow relaxation of radial deformation upon removal of the load can result in a different reference configuration depending on the loading magnitude and time. Such slow relaxation was observed in this study. Pressure was rapidly increased from 0-120 mmHg, held for 1 minute and then rapidly decreased to 0 mmHg. The overall particle movements on loading were 4-6 mm. However, the difference between the radial position of the particles just before loading and after unloading was 0.01-0.02 mm. The resulting radial movement is less than 1% and is negligible.

Volume change of the specimen may be caused by tissue compressibility, fluid transfer, and/or viscoelasticity. If these effects are significant, specimen volume change depends on loading and its time, fluid transfer and its rate, and deformation and its rate. An experiment was performed to measure the volume change of a specimen with time and pressure.

Figure 2 shows results of volume change with loading time (thoracic aorta, 53yr, female). Volume is rapidly reduced by about 12% when the pressure is increased to 120 mmHg for 2 minutes and held static for 4 minutes for volume measurement. Then, pressure is continuously held at 120 mmHg, and volume shows about 10% decrease over the next 1 hour. Pressure is then released to 0 mmHg, and about 12% volume is rapidly recovered. Then the volume appears to be slowly and asymptotically recovered to within 5% of the original volume over the next 2 hours.

5. Conclusion and Discussion

The initial volume reduction at the point of increased loading may be due to compressibility and/or rapid fluid loss (Nerem and Cornhill, 1980). To address the causes of this volume change, an experiment was performed. If loading time is too short to transfer fluid, say 2 seconds, then volume change in such a short time scale is considered to be due to compressibility. Volume is measured before and after loading to 120mmHg (lumen pressure) for 2 seconds. The result shows that the difference between initial volume and volume after quick loading and unloading is 4 %. Furthermore, as shown in Figure 2, the initial 12% volume change with quick loading is the same as the elastic volume recovery observed when the pressure is released. Therefore, most volume change with quick pressurizing is due to compressibility (Chuong and Fung, 1984). Continuous volume decrease for the next 1 hour at constant loading may be due to creep and continuous fluid loss as shown in Figure 2. Volume recovery may be from fluid gain, viscoelastic deformation recovery and elastic deformation recovery. The causes of volume deformation and recovery with time and loading are not clear. However, Figure 2 does, at least, show that there is a volume change with time at constant loading and volume recovery with time upon unloading. The time for volume recovery appears to depend on static loading time. Therefore, these observations could indicate that the time scales for loading and unloading should be the same to have consistent volumetric references and configurations. The time scale for loading and unloading could be different when sequential measurements at several pressure steps. In order to reduce the volume change at static loading, the loading time should be reduced. Also, in order to obtain a consistent volumetric reference state, the unloading time should be considered based on the loading time. If deformation measurement for one pressure step takes 1 minute in the experiment, the volume recovery against volume change for 1 minute should be predicted. Since both viscoelasticity and fluid diffusivity could be explained as exponential behavior (Harrison, R.G., and Massaro, T.A., 1976), an exponential model is proposed for predicting volume change in any time scale;

$$V(t)=V_o(1-ae^{-\frac{t}{\tau}})$$

where Vo is the asymptotic volume and t, τ and α are time, time constant, and constant respectively. Figure 3 shows the results of the exponential curves which have been fitted to experimental data as shown Figure 2 by varying τ and α. Based on these exponential models, volume change in 1 minute is predicted for both static loading and unloading. Volume change in 1 minute is calculated at static loading and the recovery time of this volume change is calculated. Volume changes 1% in 1 minute at static loading and this 1% volume change is recovered in 1.8 minutes. Therefore, deformation measurement in this protocol is within 1% volume change. If constant loading for 1 minute is followed by unloading for 1.8 minute at each measurement, the specimen has consistent volume reference since volume is mostly recovered in this time scale. In light of this, the experimental protocol requires that each measurement at each loading is followed by unloading for intervals which are twice the static loading time. In this manner, consistent volumetric reference states are achievable. Although the time scale is one minute loading as a example in this study, this result may be applicable to more time scale and loading.

6. Reference

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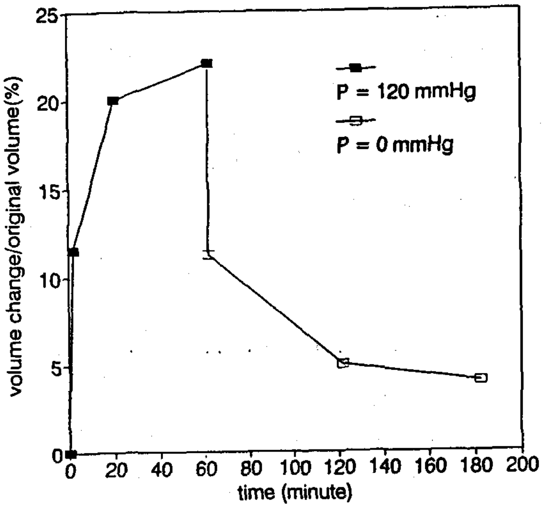


Figure 2. Volume change with time at increasing lumen pressure (0-120mmHg) and unloading to 0mmHg lumen pressure (thoracic aorta, 53yr, female). The data plotted assuem that the volume of the specimen is constant over the measurement time.

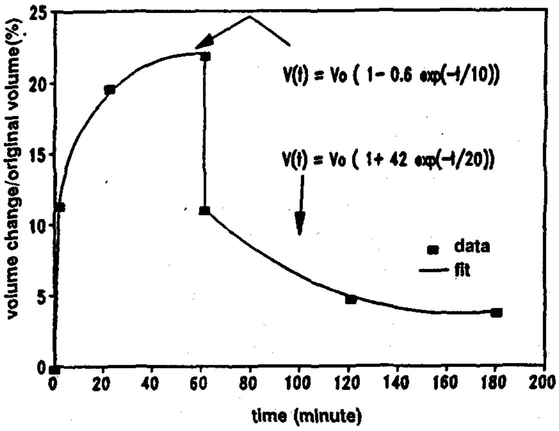
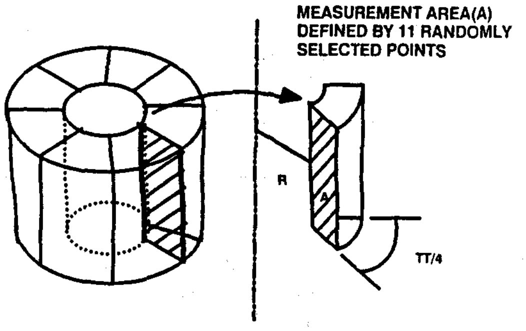


Figure 3. The resultss of experimental data are approximated by two exponential models: one for volumen decrease at static lumen pressure of 120mmHg for 1 hour and one for volume recovery at unloading



$$V_{total} = \sum_{n=0}^{8} \int_{(n-1)\pi/4}^{n\pi/4} ARd\theta \qquad v = \int_{(n-1)\pi/4}^{n\pi/4} ARd\theta$$

Figure 1. Schematic view of determination of sector volume.