BIOMECHANICS OF ABDOMINAL AORTIC ANEURYSMS: BIAXIAL EXPERIMENTS AND RELATED MODELING, TISSUE DISSECTIONS AND MICROSTRUCTURAL CHARACTERIZATION

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Summary An investigation of the biomechanical properties of abdominal aortic aneurysms (AAA) provides a basis for a refined understanding of growth and remodeling. We performed biaxial experiments and peeling tests on 61 human AAA samples to investigate the biaxial responses and the dissection properties of the tissue. Microstructural characterization was used to determine the relative thrombus age. The results suggest that thrombus aging due to microstructural changes leads to wall weakening in the lesion.

INTRODUCTION

Abdominal aortic aneurysms are localized, balloon-like dilatations of abdominal aortas that exceed the normal diameter by more than 50%. An untreated AAA may be at high risk of rupture, which has an overall mortality rate between 75 and 90% [1]. The decision to electively repair AAA is frequently determined by the likelihood of rupture, which is mainly based on the unreliable 'maximum diameter criterion', i.e. 5-5.5 cm. To date no reliable criterion can yet predict the risk of AAA rupture and help in the final clinical decision. From a biomechanical point of view, the lesion rupture is a mechanical failure of the aneurysm wall, which occurs when the peak wall stress (PWS) exceeds the local strength of the aortic tissue. Therefore, the balance between PWS and wall strength is a critical point. An intraluminal thrombus (ILT), present in the majority of AAAs, is a three-dimensional fibrin structure incorporated with blood proteins, blood cells, platelets and cellular debris [2]. Serving as a mechanical cushion, the presence of the ILT alters the wall stress distribution and reduces the PWS in AAAs [3,4]. Current work encompasses four major goals, which are (i) systematically exploring biaxial mechanical properties of the ILT and the AAA wall underneath the thrombus, and developing more appropriate 3D material models to characterize the biaxial mechanical behaviors of both tissue types, (ii) quantitatively investigating dissection properties of AAA, (iii) histologically analyzing the mircrostructure of ILT for thrombus age determination, and (iv) correlating ILT age with changes in the biomechanical properties of AAA. For a more detailed discussion on open problems of the biomechanics, mechanobiology, and the modeling of AAAs see [5]. The findings may provide us with a more comprehensive understanding of the mechanical role of the ILT in the time course of lesion growth and remodeling.

METHODS AND MATERIALS

Biaxial Extension Tests and Material Modeling

Square specimens, approximately 2×2 cm, were cut from the thickest part of ILT samples and the corresponding AAA wall segments. Four nylon sutures were fixed to each side of the square specimens with fish hooks, and four markers were placed in a distance of approximately 6 mm from each other at the center of the specimen. Square specimens were mounted in a biaxial testing device by hooked nylon sutures via four connecting carriages, see Fig. 1(a), and were then submerged into a bath filled with DMEM and maintained at 37.0 ± 1.0°C by a heater-circulation unit. A stress-driven protocol was used during the testing procedure, where the first Piola-Kirchhoff stresses served as a measure, i.e.

$$P_{\theta\theta} = \frac{f_\theta}{TX_L}, \qquad P_{LL} = \frac{f_L}{TX_\theta}, \tag{1}$$

where f_θ and f_L denote the measured forces in each direction, T is the mean thickness in the unloaded configuration, and X_θ and X_L are the initial dimensions of the square specimen, i.e. 2 cm. Each specimen was tested using the following protocols: $P_{\theta\theta} : P_{LL} = 1:1$, $0.75:1$, $1:0.75$, $0.5:1$, $1:0.5$, $0.25:1$, $1:0.25$ and $1:1$. We started with 15 kPa to do the biaxial protocol for the individual ILT layers and with 150 kPa for the degenerated AAA wall. Throughout the test 2 kPa/sec and 20 kPa/sec were maintained during the loading cycles for the ILT and the AAA wall, respectively.

To model the mechanical responses of both ILT and thrombus-covered wall, we used a strain-energy function Ψ which is based on a model developed for arterial walls [6], i.e.

$$\Psi = \mu(I_1 - 3) + \frac{k_1}{k_2}(\exp\{k_2[(1-\rho)(I_1-3)^2 + \rho(I_4-1)^2]\} - 1), \tag{2}$$

where $\mu > 0$ and $k_1 > 0$ are stress-like parameters with dimension kPa, and $\rho \in [0,1]$ and $k_2 > 0$ are dimensionless. The data from the five biaxial protocols ($P_{\theta\theta} : P_{LL} = 1:1$, $0.75:1$, $1:0.75$, $0.5:1$, $1:0.5$) for each specimen

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(a) (b)

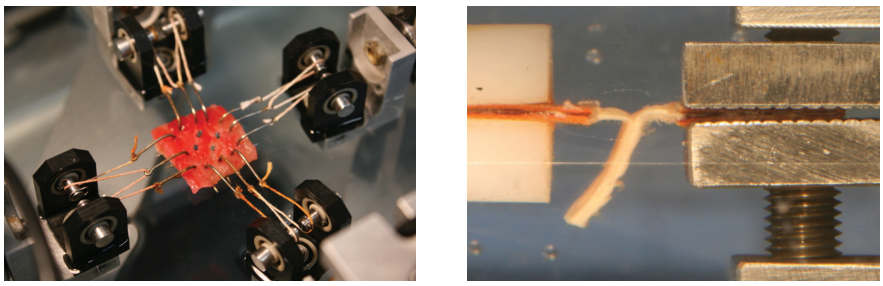

Figure 1. Schematic sketch of a square specimen fixed by four-sided nylon sutures associated with four markers (a); representative image of a specimen mounted in the testing machine before peeling (b).

associated with the circumferential and longitudinal directions were simultaneously fitted to the material model (2) using the optimization toolbox 'lsqnonlin' in Matlab 7.0, and the related parameters were obtained.

Peeling Tests

Peeling tests [7] were performed to dissect within the individual ILT layers, through the ILT thickness and within the aneurysm wall using two extension rates, i.e. 1 mm/min and 1 mm/sec. The dimension of rectangular strip used for the peeling tests was 18.0×6.0 mm (length $\times$ width). In order to better control the initiation of tissue failure, each strip was given an initial cut (incision of about 2.0-3.0 mm in length) using a surgical scalpel in the specimen preparation. Two 'tongues' were then obtained for mounting them into the testing machine, see Fig. 1(b). The dissection energy per reference area, say W^{dissect}, during the (circumferential c and longitudinal l) peeling (p) was quantified by subtracting the elastic energy W^{elastic} from the external work W^{ext}, i.e.

$$W^{\mathrm{dissect}}_{\mathrm{pc}} = (W^{\mathrm{ext}}_{\mathrm{c}} - W^{\mathrm{elastic}}_{\mathrm{pc}})/L_{\mathrm{pc}}, \qquad W^{\mathrm{dissect}}_{\mathrm{pl}} = (W^{\mathrm{ext}}_{\mathrm{l}} - W^{\mathrm{elastic}}_{\mathrm{pl}})/L_{\mathrm{pl}}, \tag{3}$$

where L_{pc} and L_{pl} denote the reference lengths of the strips in the circumferential and longitudinal directions, respectively.

Microstructural Characterization

In general, the thrombi within the AAAs consist mainly of four types of components including (i) erythrocytes, (ii) loose fibrin networks with thin bundles, (iii) fibrin networks with thick bundles and (iv) condensed small proteins [8]. We propose a four-phase thrombus evolution within the AAA. In phase I (very fresh), the thrombus initially formed with almost only the erythrocytes from the blood stream. It might contain some other components such as loose fibrin network, leucocytes, or thrombocytes, but they were all in low percentages. In phase II (young), fibrin network started to grow in a very loose manner and caught erythrocytes in between. During this phase, the thrombi had a large percentage of loose fibrin network (thin bundles) combined with relatively lower percentages of both erythrocytes and fibrin networks with thick bundles. In phase III (intermediate), the erythrocytes disrupted and proteins were washed out of the fibrin network, in which fibrin network remained and became more condensed with thick bundles. In phase IV (old), fibrin networks disrupted and residue small proteins were more condensed as compared to the previous phase.

RESULTS AND CONCLUSION

If correlating the above-mentioned relative thrombus age with both biaxial and dissection properties of aneurysmal tissue, we find that a higher ILT age leads to (i) an increase in the anisotropy and a decrease in the longitudinal strength of the aneurysmal wall, and (ii) a higher risk of dissection initiation within the ILT and the intima-media composite of the aneurysmal wall. This finding suggests that ILT aging relates to wall weakening in the lesion.

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