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An Improved Panoramic Digital Image Correlation Method for Vascular Strain Analysis and Material Characterization

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Abstract

The full potential of computational models of arterial wall mechanics has yet to be realized primarily because of a lack of data sufficient to quantify regional mechanical properties, especially in genetic, pharmacological, and surgical mouse models that can provide significant new information on the time course of adaptive or maladaptive changes as well as disease progression. The goal of this work is twofold: first, to present modifications to a recently developed panoramic – digital image correlation (p-DIC) system that significantly increase the rate of data acquisition, overall accuracy in specimen reconstruction, and thus full-field strain analysis, and the axial measurement domain for in vitro mechanical tests on excised mouse arteries and, second, to present a new method of data analysis that similarly increases the accuracy in image reconstruction while reducing the associated computational time. The utility of these advances is illustrated by presenting the first full-field strain measurements at multiple distending pressures and axial elongations for a suprarenal mouse aorta before and after exposure to elastase. Such data promise to enable improved inverse characterization of regional material properties using established computational methods.

Keywords

panoramic DIC; mouse artery; biomechanics; inverse methods

1. Introduction

Advances in computational mechanics and computer technology have enabled sophisticated finite element analyses in vascular biomechanics, including solutions of initial and boundary value problems having complex, subject-specific geometries. Nevertheless, a common limitation to most such analyses is the continued prescription of uniform material properties (cf. Humphrey and Holzapfel, 2012). This situation results not from computational or theoretical limitations, but rather from the lack of experimental quantification of actual regional variations in material properties. The existence and potential importance of non-uniform properties is supported by advances in vascular mechanobiology (cf. Humphrey,

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2008), which imply that cells should be expected to build in regional variations in material properties to offset complexities in geometry or applied loads that would otherwise result in non-uniform stress fields that are sub-optimal for biological function. Computational results based on both traditional finite element analyses (e.g., Ryan and Humphrey, 1999) and newer growth and remodeling simulations (e.g., Wilson et al., 2012) further support this expectation.

To begin to address the need for new methods of quantification of regional material properties in blood vessels having complex geometry, we introduced a panoramic-Digital Image Correlation (p-DIC) method (Genovese et al., 2011a, b). Although the basic method can be implemented to study various soft tissues having different overall sizes and shapes, we built and tested a system having a spatial resolution targeted specifically for studying the mechanics of mouse arteries, which typically are 500 to 1000 microns in diameter. This system and general approach have proven very useful in this regard (e.g., Genovese et al., 2012), but an inherent limitation restricted the rate at which data could be collected. The purpose of this paper, therefore, is to present both a novel modification to our p-DIC system, which increased significantly the rate of data collection while increasing the axial domain of measurement and overall reconstruction accuracy, and a new method of data analysis, which similarly improved the reconstruction accuracy while decreasing the computational time. To illustrate the utility of these two advances, we present new data on short-term, elastase-induced changes in regional material behaviors in the mouse abdominal aorta during in vitro experiments. Such data can contribute to the interpretation of this well used method to study the pathogenesis of abdominal aortic aneurysms.

2. Material and Methods

2.1 Improved p-DIC Approach

Briefly, our original p-DIC concept and experimental system allows geometric reconstruction and tracking of surface displacements, and thus calculation of strains, along much of the length and around the entire circumference of an excised mouse artery. To accomplish this full-field measurement, the specimen is placed co-axially within a 45° concave conical mirror. Because the resulting peculiar optical scheme of image formation requires a small stereo angle (~1°), our original p-DIC system uses a single camera rather than the multiple cameras that are used in standard stereo-DIC (cf. Sutton et al., 2009) or related methods of soft tissue strain analysis (Hsu et al., 1994; Everett et al., 2005). A “pseudo” stereo-capability is thus achieved by generating four polar-symmetric views of the specimen (namely Right and Left, Up and Down) by sequentially reflecting the image of the sample to the single fixed camera. This sequence of views is achieved by manually tilting a flat mirror four times about two perpendicular axes through a gimbal mount. Each set of four tiltings of the mirror requires about 3 minutes, however, and consequently the data cannot be collected continuously and the deformed state must remain unaltered during this measurement period. In addition to this primary limitation, minor drawbacks to our original system include a reduced sample gauge length (about 4 mm) due to the presence of a fixed calibration pattern on the uppermost inner surface of the conical mirror and a circumferentially uneven illumination of the sample achieved via four gooseneck-pipe fiber optic illuminators. See the original papers for a schema of the overall experimental set-up and the method of stereo-image generation.

Notwithstanding these practical limitations, our original p-DIC method offers many new advantages. Full-field surface deformations are inferred from 3-D information on surface geometry from successive configurations measured via the two pairs of stereo views. Each image of the specimen thus appears as a ring-like reflection on the conical mirror, which has two main advantages: (i) the full circumference of the artery is reconstructed without

interruption, that is, without border effects, and (ii) the magnification due to reflection from the conical mirror allows measurements of small samples at high spatial resolution with standard off-the-shelf lenses. The modifications proposed herein retain these advantages but overcome the primary stated limitations as well. Specifically, we replaced the gimbal-based tilting flat mirror system with a goniometer-based flat mirror that is mounted on a fixed rotational stage. Starting from the 45° position (i.e., from the optical axis of the camera being coaxial with the conical mirror), the goniometer is first set to the desired stereo angle θ_s and a full rotation of the mirror provides a “continuous” series of stereo views that effectively encircle the sample in time (Figure 1). This simple change allows one to track nearly continuously the surface deformations in successive configurations and to collect data from many more (i.e., well more than two) stereo pairs, which increases the accuracy of reconstruction as well. We also modified the conical mirror to have a “double-opposing cone” design whereby the fixed calibration pattern was applied on an outward sloping 30° surface rather than on the uppermost inner sloping 45° surface of the conical mirror (Figure 2). This simple change left the inner reflecting surface entirely free for imaging, which increased the sample gauge length from 4 to 10 mm (albeit effectively 8 mm to avoid difficulties that occur in image analysis near the upper and lower edges of the conical mirror), and allowed the calibration pattern to cover a larger depth of the measuring volume, which increased the accuracy of the Direct Linear Transformation (DLT) algorithm used for calibration (cf. Abdel-Aziz and Karara, 1971; Genovese et al., 2011a). Finally, the previously used four gooseneck-pipe fiber optic illuminators were replaced by a single fluorescent ring lamp, which rendered the illumination much more uniform circumferentially and thereby enhanced the accuracy of image correlation in data analysis.

By using a rotating flat mirror, rather than the prior tilting flat mirror, one can perform time-resolved panoramic stereo measurements with a significantly reduced lag time between two successive configurations. This lag time is limited primarily by the camera acquisition frame rate (FR) and the speed of the rotation of the mirror. A complete rotation around the axis r of the rotational stage (Figure 1) allows one to collect a set of $N = FR \cdot \text{rotation period}$ virtual stereo views, with N measured in frames per second (fps) and rotation period measured in seconds (s). Using a standard camera (e.g., having a FR of 30 fps) and a high speed motorized rotational stage (e.g., with a rotation speed of $720^\circ/\text{s}$), the time resolution of the measurement could be reduced from the original ~ 3 minutes to ~ 0.5 seconds. In the present study, however, we used a FR of 25 fps and we “spun” the mirror manually, which yielded a data acquisition period of ~ 3 seconds, which is ~ 2 percent of the original period. As noted above, increasing the number of stereo views beyond the original two pairs (i.e., Right-Left and Up-Down) also improves the accuracy of the 3D reconstruction process. To appreciate this improvement, consider the simple example of $N = 2, 4$ (i.e., equal to the original number), or 8 (used in subsequent calculations herein) polar-symmetric stereo views of a 1 mm diameter cylindrical tube that served as a phantom (Figure 3a). Figure 3b shows that a single image pair (e.g., stereo views 1-5 in Figure 3a) yields a poor overall reconstruction because of the small stereo-angle and the peculiar image formation (cf. Genovese et al., 2011a). That is, reflections from the curved surface of the conical mirror virtually increase the stereo-angle for points circumferentially distant from the plane containing the axes of the paired virtual cameras (see dashed line in Figure 3b), thus improving the corresponding reconstruction accuracy. Yet, close to the stereo-plane, the viewing angle remains close to the actual angle ($\sim 1^\circ$) and the paired images have little pixel disparity, which leads to poor reconstruction accuracy. This problem was overcome in Genovese et al. (2011a,b) by merging, in a complementary way, the more accurate parts of the data (obtained as point clouds) obtained from two stereo pairs (e.g., 1-5 and 3-7 in Figure 3a) that were spaced evenly around the 360° imaging domain. This procedure was accomplished by selecting point clouds that avoided discontinuities between adjacent portions of the reconstructed

shape. Albeit effective (cf. Genovese et al., 2012), this interactive approach was subjective and time consuming.

With the new optical set-up, however, the increased number of stereo views (e.g., the 8 views in Figure 3) enabled an improved method of reconstruction that benefits from the redundancy in the available data. After each “camera” is calibrated via the DLT method, positions on the surface of the conical mirror for the i^{th} image point I_k^i (with $i=1,\dots,n$ the number of the measured points and $k=1,\dots,m$ the number of stereo views) and of the pin-hole H_k are known relative to a global reference system (please see Figure 2 in Genovese et al., 2011a). Hence, using a ray-tracing procedure, one can compute the line $\overrightarrow{K_k^i P^i}$ by mirroring the line $\overrightarrow{I_k^i H_k}$ with respect to the local normal to the conical surface at the intersection point K_k^i . Then, the i^{th} point P^i lying on the surface of the sample is automatically calculated as the intersection of the N lines $\overrightarrow{K_k^i P^i}$ by solving the over-determined system of equations using a least-squares approach. In this way, the effect of image point pairs with low disparity decreases as the number N of stereo views increases well beyond two. Figure 3c shows the significant decrease in error e_R in reconstructing radius along the full length of the 1-mm diameter phantom using this approach for $N=2, 4$, and 8. It should be noted that $N=8$ was found to be sufficient herein using this approach for the associated reconstruction accuracy was of the same order as the physical resolution of the system. Nevertheless, the reconstruction accuracy could be increased further by adopting a weighted least square scheme that uses as weights the Normalized Cross Correlation Coefficient.

2.2 Experimental set-up

Figure 2 also pictures the overall p-DIC system developed and tested in this work. The basic experimental arrangement followed closely that of the original p-DIC system (Genovese et al., 2011a) with the exception, as noted above, of the new conical mirror, the use of a rotating rather than a tilting stage for the flat reflecting mirror, and the use of a ring illuminator rather than four individual gooseneck illuminators. Because the accuracy of the geometric reconstruction and subsequent computation of full-field displacements and strains depends strongly on the alignment of the optical components, the camera is fixed rigidly to the optical bench using a 90° bracket. The deflecting flat mirror (initially at 45°) and the conical mirror are both mounted on triaxial translational stages for reciprocal adjustment of position whereas the conical mirror is mounted on a kinematic mount for fine alignment of its axis with the axis of the camera. The fluorescent ring light is positioned just above the conical mirror using a magnetic mount, which facilitates its easy removal during operations such as alignment of the conical mirror, positioning of the sample, filling the bath with physiologic solution, general cleaning, and so forth. When the fluorescent ring light was removed, a gooseneck fiber optic illuminator served as an auxiliary lighting system. Finally, a system of tubes connected a syringe pump and pressure transducer to the inlet to the sample via a three-way stopcock. The syringe pump, pressure transducer, and power supply for the fluorescent ring light were placed on a standalone shelf close to the optical bench to isolate the possible sources of vibrations.

2.3 Testing protocol

To illustrate the utility of this improved p-DIC system for studies in vascular mechanics, we quantified full-field surface strains for a suprarenal mouse aorta that was subjected to cyclic pressure-diameter testing both before and after intraluminal exposure to porcine pancreatic elastase; note that the cannulation procedure also allowed the vessel to elongate during pressurization and thus to deform biaxially. Moreover, we also resolved changes in strains,

at a fixed pressure, over the first 30 minutes of the intraluminal perfusion with elastase. All procedures involving mice were approved by the Yale University Institutional Animal Care and Use Committee (IACUC). Briefly, a 13 week old female C57BL/6 mouse was euthanized via an overdose of sodium pentobarbital and the suprarenal aorta was excised from the diaphragm to the left renal artery (Figure 4a). The aortic sample was then prepared by removing excess perivascular tissue and ligating all branches using single threads from a 7-0 silk suture (note: each suture consists of three wound threads). The proximal end of the aorta was secured on a blunt 23G needle (nominal diameter = 0.641 mm) and the distal end was secured on a blunt 30G needle (nominal diameter = 0.311 mm). Using two different sized needles accommodated the slight taper along the length of the sample, but more importantly allowed the smaller needle to be inserted through the larger one. In this way, the axial length of the sample could be adjusted and either fixed or allowed to change freely while maintaining the same long axis during pressurization; this procedure did artificially straighten the vessel, however, consistent with that during standard biaxial tests (cf. Collins et al., 2012). Note, too, that the upper end of the 30G needle was plugged with a small rubber cap (Figure 4b) either to render it pressure tight or, by removing the cap, to allow the lumen of the sample to be flushed when necessary. The aorta was then stained lightly with Evan's blue dye to attenuate wall transparency and create a dark background for the white speckle pattern (Figure 4c) that was applied using a fine-tipped airbrush and diluted white India ink. This (random) speckle pattern created a surface character sufficient for p-DIC while not affecting the mechanical properties of the arterial wall (Genovese et al., 2011b). Finally, the sample was placed coaxially within the conical mirror, connected to the pressurized tubing system, and immersed by gently filling the conical mirror and associated chamber with a Dulbecco's Phosphate Buffered Solution (PBS).

Mechanical testing consisted of three preconditioning cycles wherein the specimen was pressurized from ~0 to 80 mmHg while allowed to elongate axially under the action of the pressure. Following preconditioning, the sample was pressurized from ~0 to 140 mmHg in 10 mmHg increments and held at each increment in pressure for 1 minute. At the end of this brief hold, data were collected over ~3 seconds to enable data acquisition via a video frame sequence that was captured at 25 fps while manually rotating the flat mirror through at least 360°. Each set of 8-bit 2352×1728 pixel² images was saved for subsequent data processing as noted above (see, too, Genovese et al., 2011a).

Next, the rubber cap was carefully removed from the end of the 30G needle and a clear tube was inserted in its place. Using a second three-way stopcock placed just below the sample bath, the sample was then gently flushed with air (with no concern for loss of endothelial cells) to remove the PBS from within the sample. The sample was then collapsed gently by creating a small vacuum to remove air bubbles trapped in the vessel, and elastase (7.5 U/mg at 37°C) was delivered intraluminally until it appeared in the clear tube. The tube was then removed and the rubber cap replaced to render the sample pressure tight. Finally, the tubing was reconnected to the syringe pump and a constant 80 mmHg pressure was applied for 30 minutes, during which p-DIC measurements were collected nearly continuously via a sequence of ~3 s optical scans. Finally, the sample was flushed with and re-immersed in fresh PBS containing 2 U/ml of aprotinin for 20 min (Fonck et al., 2007); aprotinin is an inhibitor of elastase activity that minimized further intramural changes in wall structure. Finally, the aforementioned mechanical tests were repeated with pressurization over the range ~0 to 140 mmHg.

2.4 Strain Calculation

Full-surface, that is around the entire circumference and along most of the axial length of the sample, Green strains were calculated locally using established methods (Humphrey, 2002). Briefly, local triangular domains were defined in both the reference and subsequent current

configurations based on the reconstructed surface geometries, for which local mappings were computed by assuming separate homogeneous finite deformations within each “sub-domain”. Components of the 2-D deformation gradient tensor were then determined based on distances between the vertices of each triangular sub-domain (i.e., based on the mapping of finite, but short, position vectors between two neighboring points) in each configuration of interest. See Genovese et al. (2011a) for further details.

3. Results

Figure 5 shows calculated full-field circumferential, in-plane shear, and axial Green strains in the native aorta at pressures of 80 and 120 mmHg, with strains computed relative to a non-collapsed reference configuration maintained at 0 to 3 mmHg. All values of strain are superimposed on the actual geometry for ease of visualization. Note the regional variations in strain, with mean values $\pm$ standard deviation being 0.185 ± 0.088 , 0.003 ± 0.021 , and 0.176 ± 0.075 at 80 mmHg and 0.407 ± 0.141 , 0.015 ± 0.041 , and 0.368 ± 0.167 at 120 mmHg for the circumferential, in-plane shear, and axial values, respectively. Hence, the deformation field was, on average, nearly principal and equibiaxial. Of course, having computed the individual components of the surface strain, it is then straightforward to compute actual principal values (i.e., to determine the eigenvalues) as well as any of a number of possible scalar invariants, including the magnitude of the strain or the first or second principal invariants. Such metrics should be selected based on a specific theoretical motivation, typically in relation to the constitutive formulation of interest.

Figure 6 shows the evolution of Green strains during the intraluminal exposure of the aorta to elastase while maintaining the pressure at 80 mmHg; for clarity, results are only shown at 0, 10, 20, and 30 minutes of exposure with strains calculated relative to the configuration at 80 mmHg prior to exposure, which highlights the regional changes as a function of time of exposure. Mean values $\pm$ standard deviations were 0.111 ± 0.025 , 0.183 ± 0.074 , and 0.218 ± 0.078 in the circumferential direction, -0.002 ± 0.007 , $7.79\text{E-}05 \pm 0.007$, and $5.18\text{E-}04 \pm 0.010$ for the in-plane shear, and 0.046 ± 0.01 , 0.070 ± 0.025 , and 0.077 ± 0.021 in the axial direction.

Finally, Figure 7 shows results similar to those in Figure 5 except based on repeated pressure-diameter testing following the 30-minute exposure to elastase (cf. Figure 6). Note the significant dilatation and increased regional variations in strain, with mean values decreasing significantly, relative to the updated reference configuration at 0 to 3 mmHg, to 0.056 ± 0.026 , -0.0002 ± 0.0031 , and 0.021 ± 0.010 at 80 mmHg and 0.079 ± 0.028 , -0.0003 ± 0.0036 , and 0.023 ± 0.011 at 120 mmHg. Although the strains remained nearly principal on average, they became less equibiaxial. For easier comparison, we list in Table 1 the mean values of strain associated with Figures 5 to 7.

4. Discussion

Experimentalists typically seek to identify experiments that represent simple initial or boundary value problems for they simplify greatly the interpretation of the data. Hence, classical experiments in mechanics and biomechanics that are designed to elucidate material behavior are well represented by standard uniaxial and biaxial stretching tests wherein nearly homogeneous finite deformations can be inferred in central regions well away from end effects. Nevertheless, there are many situations in biomechanics, particularly in pathophysiologic cases, wherein the situation of interest demands that one perform experiments that are complicated by complex geometries or applied loads. In such cases, one must identify new methods for inferring material behavior.

Towards this end, we proposed a number of years ago a sub-domain inverse finite element method for characterizing locally the mechanical behavior of thin biological soft tissues (Seshaiyer and Humphrey, 2003). Since then, this method has proven useful in diverse applications, most recently in the quantification of local mechanical properties in the lens capsule of the eye (Pedrigi et al., 2007). Prior implementations of this approach have been based on experimental data that included local calculations of strain based on the tracking of a finite number of closely spaced “tracking markers” using bi-plane video camera systems (cf. Hsu et al., 1995; Everett et al., 2005). Although effective, such experiments are limited in practice to a small number of regions of interest and they often require painstaking placement of the tracking markers on the surface of the sample. In contrast, the present full-field approach provides comparable information in nearly limitless regions of interest following a simple preparation of the surface, as, for example, using an air-brush to spray the surface with India ink, which we and others have shown does not affect the mechanical properties (e.g., Genovese et al., 2011b).

In this paper, we showed that modest, but important, modifications of our prior experimental implementation of the p-DIC approach enabled a significant increase in the rate of data acquisition (from sampling periods of ~3 minutes to 3 seconds, with ~0.5 seconds possible with motorized stages) as well as in the accuracy of geometric reconstruction (using many stereo views rather than the minimum of two stereo-pairs) over a larger axial domain (from 4 to 8 mm given the present optics). We similarly presented an improved, objective method for data analysis that relies on the redundant data supplied by the new multi-view p-DIC system. These advances were demonstrated using standard pressure-diameter testing of a mouse artery, with the extra demonstration of monitoring evolving strain fields during exposure of the vessel to elastase. This protease degrades the elastin within the arterial wall, which is the primary constituent (90%) of the elastic fibers that comprise the elastic lamellae (Humphrey, 2002). Loss of elastin leads to marked dilatation, with a significant loss of distensibility (circumferential) and extensibility (axial) in response to applied loads (cf. Fonck et al., 2007; Collins et al., 2012). The present data confirm prior findings in this regard, but extend those findings by showing evolving regional changes as well. Indeed, because of the continued dramatic dilatations over the initial 20 to 30 minutes of exposure, nearly continuous data collection was needed to track the sequential configurations needed to compute strains (cf. Figure 6, which shows strains at only a few times). In particular, it was found that effects of the elastase tended to be greater in the lower regions of the vessel (relative to the direction of gravity during the experiment wherein the specimen was mounted vertically), which suggested a possible non-uniform intraluminal exposure of the aortic wall to the elastase despite perfusion with a well-mixed solution under pressure – here, note that Figure 7 shows strains computed relative to the updated nearly unloaded configuration, which revealed greater distensibility and extensibility in the upper regions that were less affected by the elastase, whereas Figure 6 shows strains computed relative to the native configuration at 80 mmHg, which reveals a consistent greater dilatatory effect of elastase in the lower region. Indeed, based on this observation, we revisited findings presented in Ferruzzi et al. (2011) and confirmed that regions of greater changes due to elastase similarly correlated with the direction of gravity (with the specimen mounted horizontally in a standard biaxial testing system). There is a need for caution, therefore, in the interpretation of data resulting from intraluminal exposures of arteries to any effector solution (e.g., proteolytic or vasoactive), and full-field measurements clearly will aid in this regard.

Finally, it is emphasized that, similar to that which is done in standard uniaxial and biaxial tests, the present experimental approach artificially straightened the aorta when fixing it within the device (cf. Figure 4). Whereas one typically assumes homogeneous deformations in associated uniaxial and biaxial tests within a central gauge length, the present results (cf.

Figure 5) clearly reveal that this is not the case – there were circumferential variations in strain at a particular axial location, even within the central region. Again, advantages of full-field strain measurements are clear. Nevertheless, it is interesting to note that the “end effects” due to securing the aorta to the cannulae, with suture, near its ends tended to be localized near the ends, thus supporting the common St. Venant type assumptions regarding the utility of collecting data within central gauge regions (note: the ends of the aorta were not visible in the p-DIC image following treatment with elastase because of the marked elongation as well as dilatation). Moreover, our use of a “double” cannula, formed by inserting a smaller diameter blunt needle within a larger one, allowed the vessel both to extend naturally due to pressurization and to tend back toward its natural curved geometry (cf. Figures 4 and 7) under certain conditions. Such a cannula appears to be useful experimentally when axial extensions are not controlled externally (cf. Genovese et al., 2011b).

In conclusion, consistent with our prior findings for aortic aneurysms excised from an angiotensin-II infusion mouse model (Genovese et al., 2012), regional variations in strain should be expected in many biomechanical experiments and full-field methods of measurement can thus prove to be very useful. Indeed, given advances in inverse methods for local characterization of material properties (e.g., Seshaiyer and Humphrey, 2003; Avril et al., 2010; Zhao et al., 2011), the improved panoramic – Digital Image Correlation method presented herein promises to contribute significantly both to material characterization and the correlation of local mechanical behavior with underlying microstructure (based on histological information that is now easily collected using both intravital and standard histological methods; cf. Ferruzzi et al., 2011; Schriebl et al., 2012).

Acknowledgments

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Figure 1.

Schema motivating a multi-view panoramic-Digital Image Correlation (p-DIC) method. A full rotation of the goniometer-mounted flat mirror about the r axis yields a full rotation of the viewpoint (i.e., virtual camera) around the axis s of the specimen that is placed co-axially within the conical mirror. The stereo angle θ_s is exaggerated for clarity of representation, but is typically about 1 degree.

Figure 2.

Photograph of the modified panoramic-Digital Image Correlation (p-DIC) system showing the overall set-up as well as the three new components: the conical mirror (A), the goniometer and rotational stage (B, C) that hold the flat mirror (G), and the ring illuminator (D). Other components include: the video-camera (E), 45-degree bracket (F), specimen bath (H), syringe pump (I), pressure transducer (L), tubing (M), and auxiliary lighting (N), if needed.

Figure 3.

Illustrative 3-D shape reconstruction of a 1-mm diameter cannula using the multi-view p-DIC system. Panel a: top-view photograph showing a p-DIC image of a centrally placed cannula as well as the fixed calibration pattern on the outer downward sloping portion of the conical mirror; note, too, the superimposed directions of the eight illustrative polar-symmetrical lateral views. Panel b: top view of typical raw data obtained using only two of the stereo views shown panel a. Note the reconstruction error corresponding with the stereo-plane (dashed line). Panel c: plots of the reconstruction error for radius e_R by using $N=2$, 4 (i.e., the original number of stereo views), or 8 (illustrative of increased capability with the multi-view system) stereo-views shown panel a.

Figure 4.

Pictures of the excised mouse suprarenal aorta tested in this work. Panel a: sample prior to mounting (note the natural curvature). Panel b: cannulated and secured native aorta mounted at $P \sim 0$ mmHg. Panel c: aorta after staining with Evans blue dye, air-brush spraying with white India ink, and intraluminal perfusion with elastase, again at $P \sim 0$ mmHg. Note that pictures in panels b and c are reported at the same length scale, which reveals the marked dilatation following exposure to elastase.

Figure 5.

Full-field measurements of circumferential (panel a), in-plane shear (panel b), and axial (panel c) Green strain before intraluminal exposure of the mouse suprarenal aorta to elastase; results are shown at three pressure levels (~0, 80, 120 mmHg), with the axial length free to expand with pressurization. Values of strain (different colors) are computed relative to the original unloaded configuration and superimposed on the actual geometry, with all geometric images shown at the same scale to facilitate comparisons. Figures 5, 6, and 7 also have the same length scales to facilitate visual comparisons, and the 0 mmHg image shows the reference image “mesh” to reveal the level of tessellation.

Figure 6.

Representative time course (i.e., shown at three times) of the distributions of circumferential (panel a), in-plane shear (panel b), and axial (panel c) Green strain during the intraluminal exposure of the mouse suprarenal aorta to elastase at 80 mmHg and 37°C. Strains are computed relative to the original configuration at 80 mmHg and plotted on the corresponding deformed shape. Note that all strain plots in Figures 5, 6, and 7 have the same length scale to facilitate visual comparisons.

Figure 7.

Distributions of circumferential (panel a), in-plane shear (panel b), and axial (panel c) Green strain after the intraluminal exposure of the mouse suprarenal aorta to elastase for 30 minutes; results shown at three different pressure levels with strains computed relative to the updated, unloaded configuration and plotted on the corresponding deformed shape. Note that all strain results in Figures 5, 6, and 7 have the same length scale to facilitate visual comparisons.

Table 1

Green Strains Before Intraluminal Exposure to Elastase (computed relative to the original configuration at ~0 mmHg)			
Pressure (mmHg)	E	E_z	E_{zz}
80	0.185 ± 0.088	0.003 ± 0.021	0.176 ± 0.075
120	0.407 ± 0.141	0.015 ± 0.041	0.368 ± 0.167
Green Strains During Intraluminal Exposure to Elastase at 80 mmHg (computed relative to the original configuration at 80 mmHg)			
Time (min)	E	E_z	E_{zz}
10	0.111 ± 0.025	-0.002 ± 0.007	0.046 ± 0.01
20	0.183 ± 0.074	7.79E-05 ± 0.007	0.070 ± 0.025
30	0.218 ± 0.078	5.18E-04 ± 0.010	0.077 ± 0.021
Green Strains After 30-min Intraluminal Exposure to Elastase at 80 mmHg (computed relative to the updated unloaded configuration at ~0 mmHg)			
Pressure (mmHg)	E	E_z	E_{zz}
80	0.056 ± 0.026	-0.0002 ± 0.0031	0.021 ± 0.010
120	0.079 ± 0.028	-0.0003 ± 0.0036	0.023 ± 0.011