NUMERICAL SIMULATIONS OF THE PLANAR BIAXIAL MECHANICAL BEHAVIOR OF BIOLOGICAL MATERIALS

Wei Sun(1), Michael S. Sacks(1), and Michael J. Scott(2)

(1)Engineered Tissue Mechanics Laboratory
McGowan Institute for Regenerative Medicine
Department of Bioengineering
University of Pittsburgh
Pittsburgh, PA 15219
Email: msacks@pitt.edu

(2)Edwards Lifesciences
Irvine, CA 92641

INTRODUCTION

Planar biaxial mechanical testing is becoming an increasingly utilized technique to characterize the mechanical properties of native and engineered soft tissues. In spite of its growing use, biaxial testing methods have not been standardized and methodologies vary widely. Due to the complex mechanical behavior of soft tissues and the general difficulty in performing biaxial tests [1], experimental artifacts can easily be introduced. A critical factor affecting biaxial mechanical behavior is the boundary conditions used in testing, such as the uniformity of the loading along the specimen edge and the method of attachment (e.g. grip, sutures). The strong influence of boundary conditions in biaxial tests were recently underscored by Waldman *et al* [2], wherein substantially different experimental results were obtained using different sample gripping methods on the *same* specimens. Moreover, the choice of attachment method may depend on the anticipated degree of mechanical anisotropy and compliance. As it is not possible to directly evaluate the effects of different boundary conditions on specimen internal stress distributions, we conducted the following numerical study to explore these effects. A general nonlinear anisotropic Fung type constitutive model, fully incorporating the effects of in-plane shear [3], was used. Biaxial mechanical test simulations were conducted with different material axes orientations under different attachment methods to examine the affects of tissue anisotropy and boundary conditions.

METHODS

Constitutive model. It is assumed that a planar biological material is hyperelastic in the pseudoelastic sense [4]. Thus the in-plane 2nd Piola-Kirchhoff stresses ($\mathbf{S}$) can be derived from a strain energy function W through:

$$\mathbf{S} = \frac{\partial W}{\partial \mathbf{E}} \tag{1}$$

Where $\mathbf{E}$ is Green-Lagrange strain tensor. We used a general Fung type model, with the ability to characterize in-plane shear response:

$$W = \frac{c}{2}\left[e^{Q} - 1\right],$$

$$Q = A_1E_{11}^2 + A_2E_{22}^2 + 2A_3E_{11}E_{22} + A_4E_{12}^2 + 2A_5E_{12}E_{11} + 2A_6E_{12}E_{22} \tag{2}$$

Where c and A are material constants.

Material constants. In this study, glutaraldehyde treated bovine pericardium was used as representative biomaterial. Multiple test protocols wherein the ratio of the axial stress were kept constant were collected for extensive coverage of the tissue stress-strain response. Further, two material axes orientations were used: I) 45 degree with respect to the specimen axes (x_1-x_2 in Fig. 1-a); II) 0 degree (x'_1-x'_2 in Fig. 1-a). The methods for obtaining the parameters c and A through non-linear regression and corresponding constraints have been previously presented [3]. It should also be noted that our approach is general for any biological tissue or biomaterial that follows a Fung model.

Finite element implementation. Eqn. 2 was incorporated into the commercial finite element software ABAQUS through the subroutine UMAT. 8-node biquadratic, reduced integration plane stress elements (ABAQUS element type CPS8R) were used for all simulations. Quasi-static simulations allowing for non-linearities arising from both the constitutive law and the large deformation were performed [3].

Biaxial testing simulations. FE models were developed for 25 mm x 25 mm x 0.4 mm test specimens using 400 CPS8R elements. Equibiaxial loading was simulated to examine and compare the effects of different boundary conditions. Two groups of simulations were conducted for each material axes orientation: one group explored the effects of different numbers of suture attachments. The specimen was attached with 4, 6 or 8 points of sutures on each edge. Point loads were applied at the boundary nodes (Fig. 1-a) to exert a net total 1 MPa Lagrangian stress along each side. The other group explored the effects of different gripping methods. The gripping methods are: a) suture attachment (SA) (Fig. 1-a); b) clamps on a square specimen (CSS) (Fig. 1-b); c) clamps on a cruciform specimen (CCS) (Fig. 1-c). Two leg lengths of CCS were chosen, one with 0.5-length of 25mm (CCS05) and the other with 1.5-length of 25 mm (CCS15). A 0.5 MPa Lagrangian stress was enforced on each side of specimen. Averaged outputs of the sixteen elements located in the center of FE models, delimiting ~5 mm^2 region, were used to compare with the experimental biaxial data.

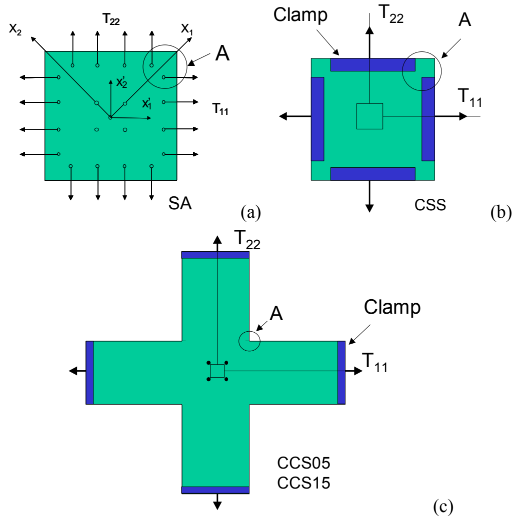

Fig. 1-Different gripping methods for FE simulations. a) suture attachment (SA); b) clamps on square sample(CSS); c) clamps on a cruciform sample with different length of legs(CCS05/CCS15).

RESULTS AND DISCUSSION

Effects of number of suture attachments The stress-strain data from the central sixteen elements from the simulation with 4 suture attachments were averaged and compared with the experimental data. An excellent match was obtained (Fig. 2). When we increased the number of sutures to 6 and 8, there was no difference between the suture loading conditions (Fig. 2). These results indicated that 4 suture attachments are sufficient for stress field uniformity in biaxial testing.

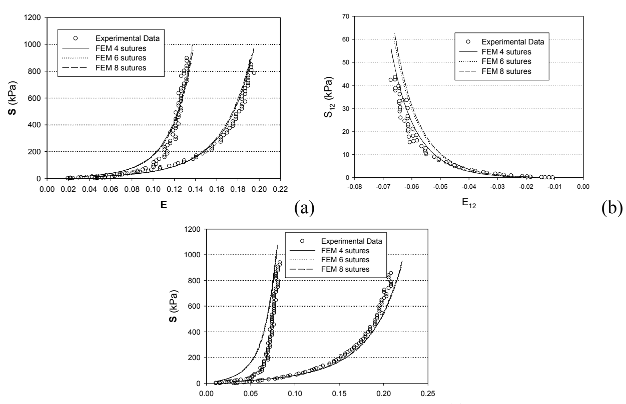

Fig. 2 - Stress-strain output of FE simulations with different numbers of suture attachments. a) 45-degree material axis, normal stresses, b) shear stress; c) stresses for 0-degree material axis.

Effects of different gripping methods. The deformation and von Mises stresses of FE models with different gripping methods for a 45-degree material axes are illustrated in Figs. 3 a-d. The SA configuration exhibited the most prominent anisotropic deformation, which is indicated by the material axis x_1 showing lesser extensibility than that of the orthogonal material axis x_2. Similar shear deformation could be observed for CCS05 and CCS15. However, for CSS, the specimen shape almost remained square. High von Mises stresses always occurred in region A (Figs. 1 and 3) due to the proximity of the corner and clamps (Fig. 3-b). The SA configuration exhibited the lowest stress in region A with von Mises stress of 8,382 kPa while the CCS configuration had the highest stress of 26,510 kPa. Further investigation of the inner stress distribution of the central sixteen elements indicated that for CSS the stress and strain were only about 30kPa and 0.045, respectively. For CCS05, CCS15 and SA, the stresses were 220kPa, 260kPa and 490kPa, respectively (Fig. 4-a). For 0-degree fiber orientation, there was a stress increase for the three clamp methods. The stress for CSS was 200kPa, and for CCS05 and CCS15 the stresses were 430kPa and 400kPa, respectively (Fig. 4-b). The results suggested that when collagen fiber orientation was aligned to the loading axes the clamped boundary artifacts are less severe.

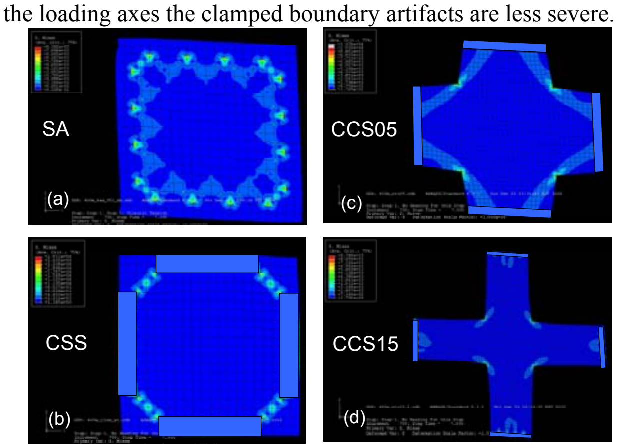

Fig. 3-FE simulation of effects of different gripping methods

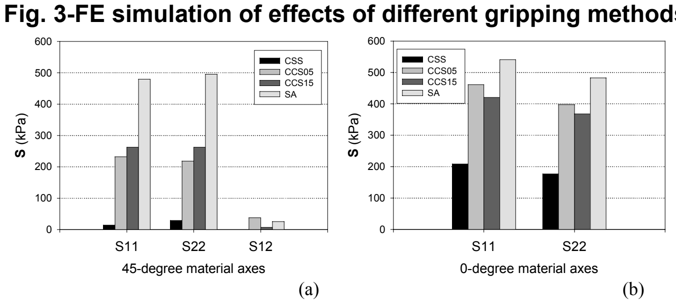

Fig. 4- Peak stress of inner central regions for a) 45-degree and b) 0-degree material axes.

Overall the three clamp methods had substantially lower stress in the central region compared to the suture attachment method. These results indicated that the clamp methods had strong boundary condition effects. Moreover, for the clamped methods the material in the inner region of the specimen was not fully loaded and therefore would not be fully stretched. The high stress concentrations for these methods makes the tissue appear to be stiffer, which is the scenario observed by Waldman *et al* [2]. This study presented the first FE simulations of biaxial tests under different gripping setups and different material axes orientations. These comprehensive, even though not exhaustive, biaxial test simulations would provide experimentalists more insights into interpreting experimental results.

ACKNOWLEDGEMENTS This work was funded by NSF grant BES-9978858 and Edwards Lifesciences Inc, Irvine, CA.

REFERENCES

1. Sacks, M.S., Journal of Elasticity, 2000. **61**: p. 199-246.
2. Waldman, S.D. *et al.* J. Materials Science: Materials in Medicine, 2002. **in-press**.
3. Sun, W., *et al.* Journal of Biomechanical Engineering, 2003. **in-press**.
4. Fung, Y.C., Biomechanics: Mechanical Properties of Living Tissues. 1993, Springer Verlag.