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COMPARISON OF MATERIAL PROPERTIES BETWEEN THE MAIN AND LEFT PULMONARY ARTERIES OF CONGENITAL HEART DISEASE SUBJECTS USING CARDIAC MAGNETIC RESONANCE: A FEASIBILITY STUDY

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INTRODUCTION

Although surgical outcomes for congenital heart disease (CHD) subjects with complex right heart disease are excellent, subjects are left with residual lesions including pulmonary artery regurgitation and obstruction that result in right ventricular-pulmonary artery (RV-PA) dysfunction [1]. The appropriate timing of re-intervention to correct these residual lesions is difficult to determine due to lack of suitable quantifiable diagnostic techniques.

The PA wall (material) properties contribute significantly to the function of the RV-PA unit as they characterize the mechanical behavior of the arterial wall which has a subsequent effect on the overall RV-PA hemodynamics. Further, these properties are expected to vary between the main PA and branched pulmonary arteries. Therefore, quantification of the properties of each of the pulmonary arteries may provide valuable information on disease status and progression and thus aid in the longitudinal assessment of CHD subjects.

Constitutive modeling of the artery wall using *in vivo* and *in vitro* test data has been previously used to characterize the complex arterial wall properties [2, 3, 4]. However, application of such modeling techniques to the pulmonary arteries of CHD subjects has been limited due to the lack of *in vitro* test data. A previous study proposed a methodology of evaluating the wall properties using only *in vivo* pressure-diameter data coupled with a set of constitutive equations that define a material model [5]. Further, cardiac MRI could be a useful tool for assessing *in vivo* diameter in pediatric subjects. As a first step, we aimed to apply this methodology to characterize the PA wall properties of normal pediatric subjects. Therefore, in this feasibility study, using pressure-diameter data as input to a material model, we quantified and compared the properties of the main PA (MPA) and left PA (LPA) of two CHD subjects with normal RV-PA function and anatomy.

METHODS

Study population. Two CHD subjects (subject N1: age = 13 years, body surface area = 1.57 m^2, subject N2: age = 19 years, body surface area = 2.06 m^2) having a diseased aorta were included for this study. However, both subjects had a normal RV-MPA physiology and normal pulmonary valve function with no PA stenosis. Thus, the two subjects were considered "normal" for this study.

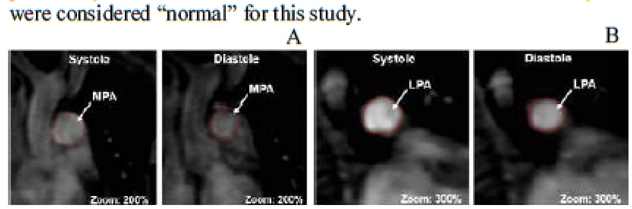

Figure 1: A representative MR image at systole (left pane) and diastole (right pane). A) MPA contour B) LPA contour

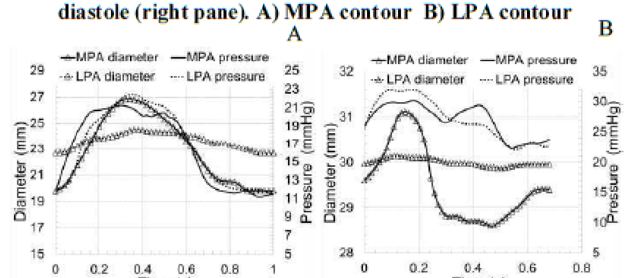

Figure 2: *In vivo* pressure and diameter pulses. A) Subject N1 B) Subject N2

Data acquisition. Cardiac MRI data was acquired using a 3.0 Tesla MRI scanner (Philips Healthcare, Best, The Netherlands). Contours of the MPA and LPA anatomy were obtained on the MRI slice using auto-thresholding (Figure 1) and the diameter (hydraulic) of the vessels was computed. In addition to MRI, cardiac catheterization was conducted and pressure was recorded at 30 time-points over a single cardiac pulse. CMR and cardiac catheterization were both performed in Cincinnati Children's Hospital Medical Center (CCHMC). The pressure and diameter data points, which were measured non-simultaneously, were synchronized using ECG gating. The gated pressure and diameter pulses are shown in Figure 2. The LPA showed smaller variations in diameter over a cardiac cycle in comparison to the MPA. However, a similar pressure pulse was observed in both the vessels.

Material model. A 2D nonlinear and anisotropic hyperelastic model [5] was used to quantify the PA wall properties and predict its mechanical behavior when subjected to bi-axial loading. The principal stresses were modeled using Fung's 2D strain energy function (SEF), $\psi = C(e^Q - 1)/2$ where, $Q = c_{\theta\theta}E_{\theta\theta}{}^2 + 2c_{\theta z}E_{\theta\theta}E_{zz} + c_{zz}E_{zz}{}^2$. The principal stretches in the circumferential (θ) and axial (z) directions were defined as $\lambda_\theta = D(A + \pi d^2)/d(A + \pi D^2)$ and $\lambda_z = z/Z$, respectively, where; A is the cross-sectional area of the wall, D and Z represent the inner diameter and length of the vessel, respectively, in the unloaded condition, and d and z represent the inner diameter and length of the vessel, respectively, in the loaded condition. The principal Cauchy stresses (in kPa), predicted by the model, in the circumferential and axial direction were defined as $\sigma_{\theta\theta}^{mod} = \lambda_\theta(\partial\psi/\partial\lambda_\theta)$ and $\sigma_{zz}^{mod} = \lambda_z(\partial\psi/\partial\lambda_z)$, respectively. In order to define the material model for the PA wall and predict the model constants, an equilibrium condition was imposed on the artery wall and the principal theoretical stresses (in kPa) were defined as $\sigma_{\theta\theta} = pd^2\pi/2A$ and $\sigma_{zz} = d^2\pi(4F + pd^2\pi)/4A(A + d^2\pi)$ where p is the pressure and F is the external axial force. The four constitutive parameters, C, $c_{\theta\theta}$, $c_{\theta z}$, and c_{zz} were then evaluated using a least-square error minimization algorithm where the error was defined as the difference between the model and theoretical stresses. An in-house MATLAB® code (MATLAB Inc., Waltham, MA, USA) was developed and used for computing the model constants.

RESULTS

Figure 3 shows the pressure and corresponding diameter recorded *in vivo* (marked by squares and triangles) during the systolic (loading) and diastolic (unloading) phases of the cardiac cycle. Distinct and non-overlapping characteristics were observed for the MPA and LPA.

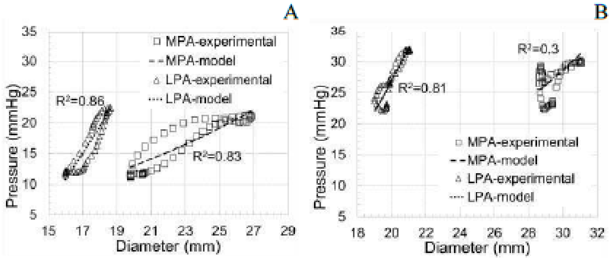

Figure 3: Pressure-diameter characteristics. A) Subject N1 B) Subject N2

Using these characteristics as input to the material model described above, the values of the material model constants which represent the arterial wall properties were obtained for the MPA and LPA of two normal subjects, N1 and N2, and are reported in Table 1. The properties varied between the MPA and LPA for both subjects. This is an expected outcome due to the distinct pressure-diameter characteristics. The C value, which defines the stress level in the artery wall, was observed to be higher for the LPA (16.8% for N1; 45.3% for N2) in comparison to the MPA. A similar trend was observed for the other constants ($c_{\theta\theta}$, $c_{\theta z}$, and c_{zz}) which govern the stress distribution in each loading direction. Further, $c_{\theta\theta}$ values were observed to be greater (absolute difference within 0.092) than c_{zz} (except for Subject N1-MPA) thus indicating a greater stress distribution in the circumferential direction.

Table 1: Values of the material model constants.

	C (kPa)	$c_{\theta\theta}$	$c_{\theta z}$	c_{zz}
Subject N1 - MPA	656	0.025	0.028	0.037
Subject N1 - LPA	766.5	0.175	0.091	0.083
Subject N2 - MPA	400	0.1	0.056	0.054
Subject N2 - LPA	581	0.219	0.159	0.214

In order to determine the accuracy of the material model, the model-predicted pressure-diameter curve (marked by dashed and dotted lines in Figure 3) was computed using the constants in Table 1 and compared against the experimental (*in vivo*) data (Figure 3). A close fit was observed between the model and experiments as can be seen from the R^2 values (ranging between 0.3 to 0.86) reported in Figure 3.

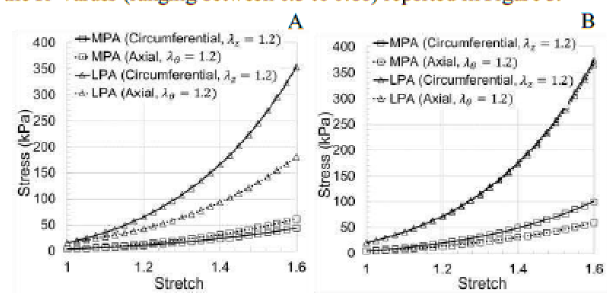

Figure 4: Stress-stretch characteristics. A) Subject N1 B) Subject N2

DISCUSSION

In vivo pressure-diameter characteristics in conjunction with constitutive equations serves as a useful method in quantifying the PA wall properties of CHD subjects. As observed from the results, these properties can quantitatively characterize and differentiate between the MPA and LPA. Further, the PA wall properties may be used to predict the mechanical behavior of the artery under a range of loading conditions, as shown in Figure 4. Circumferential and axial stresses in the MPA and LPA wall were computed at different stretch values. Higher stresses were observed for the LPA in comparison to the MPA for both subjects due to higher C values obtained for the LPA. For example, at a physiologic stretch of 1.2, $\sigma_{\theta\theta}^{mod}$ in the LPA and MPA of subject N1 were 65.9 kPa and 11.2 kPa, respectively. At the same stretch, σ_{zz}^{mod} in the LPA and MPA of subject N1 were 43 kPa and 13.8 kPa, respectively. Also, the LPA and MPA stress values were distinct and non-overlapping over the entire range of stretch values.

Thus, PA wall properties, obtained using cardiac MRI in conjunction with constitutive modeling, may provide valuable information for the longitudinal assessment of the RV-PA dysfunction in CHD patients. Subsequently, this information may aid in determining an appropriate time for re-intervention.

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