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● *Original Contribution*

# PRINCIPAL STRAIN VASCULAR ELASTOGRAPHY: SIMULATION AND PRELIMINARY CLINICAL EVALUATION

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**Abstract—It is difficult to produce reliable polar strain elastograms (radial and circumferential) because the center of the carotid artery is typically unknown. Principal strain imaging can overcome this limitation, but suboptimal lateral displacement estimates make this an impractical approach for visualizing mechanical properties within the carotid artery. We hypothesized that compounded plane wave imaging can minimize this problem. To test this hypothesis, we performed (i) simulations with vessels of varying morphology and mechanical behavior (*i.e.*, isotropic and transversely isotropic), and (ii) a pilot study with 10 healthy volunteers. The accuracy of principal and polar strain (computed using knowledge of the precise vessel center) elastograms varied between 7% and 17%. In both types of elastograms, strain concentrated at the junction between the fibrous cap and the vessel wall, and the strain magnitude decreased with increasing fibrous cap thickness. Elastograms of healthy volunteers were consistent with those of transversely isotropic homogeneous vessels; they were spatially asymmetric, a trend that was common to both principal and polar strains. No significant differences were observed in the mean strain recovered from principal and polar strains ($p > 0.05$). This investigation indicates that principal strain elastograms measured with compounding plane wave imaging overcome the problems incurred when polar strain elastograms are computed with imprecise estimates of the vessel center. (E-mail: m.doyley@rochester.edu) © 2016 World Federation for Ultrasound in Medicine & Biology.**

*Key Words:* **Anisotropy, Atherosclerosis, Plane wave imaging, Vascular elastography, Principal strain.**

## INTRODUCTION

Vascular elastography visualizes normal strains (radial and circumferential) within the carotid artery. However, several investigators are now developing methods to visualize shear strains because (i) peak shear strains and the American Heart Association classification of atherosclerotic plaques (Stary 2000) are correlated, and (ii) cyclic shear strains in the adventitia could trigger neovascular proliferation and destabilize plaques (Keshavarz-Motamed et al. 2014; Majdouline et al. 2014; Wan et al. 2014). Unfortunately, coordinate dependency limits the performance of vascular elastography. We can remove this dependency by diagonalizing the 2-D strain tensor because doing so will transform the measured values to a new coordinate system where the principal axes are the only directions along which strain is significant. Several investigators have investigated principal strain imaging (*i.e.*, diagonalizing the 2-D strain tensor). For example, Zervantonakis et al. (2007) used principal strain to reduce the dependence of strain on the transducer angle and the ventricular centroid. Jia et al. (2009) used principal strain vectors to highlight areas of ischemia. Lee et al. (2008) found that clinicians could use principal strains to differentiate abnormal from normal myocardium. However, suboptimal lateral displacement estimates have hindered the development of a practical ultrasonic approach for visualizing principal strains.

Von Mises coefficients (Maurice et al. 2002) and model-based elastography (Floc'h et al. 2010; Hansen

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et al. 2013; Huntzicker et al. 2014) also provide coordinate-independent mechanical parameters. Von Mises strain represents radial strain and, thus, does not provide any information about strains in the circumferential direction. Model-based elastography provides information about plaque composition (Baldewsing et al. 2005; Richards and Doyley 2011) that can identify life-threatening plaques. However, most modulus estimation techniques are not suitable for real-time applications, a problem that shear wave imaging techniques should overcome (Ramnarine et al. 2014). Our long-term goal is to develop practical methods for estimating coordinate-independent mechanical parameters. Therefore, the focus of the work described here was to illustrate that compounded plane wave imaging can provide accurate high-precision lateral displacement estimates, a prerequisite for principal strain imaging.

Synthetic aperture ultrasound imaging systems produce high-precision axial and lateral displacement estimates required to produce useful principal strains (Korukonda et al. 2013). Researchers have used the incompressibility condition to improve the quality of lateral displacements (Lubinski et al. 1996; Poree et al. 2015). However, incompressibility processing may not produce useful lateral displacements in transversely isotropic tissues such as the carotid artery because the Poisson's ratios in the longitudinal and the circumferential directions are not equivalent and are typically less than 0.5. Researchers have also reported that introducing lateral oscillation during beamforming (Anderson 1998; Basarab et al. 2009; Gueth et al. 2007; Jensen and Munk 1998) reduces the effective lateral beamwidth of radiofrequency (RF) echo frames and improves the quality of lateral displacement estimates. Another approach is to use an advanced beamforming method to improve the point spread function (PSF) of synthetic aperture ultrasound imaging systems (Hansen et al. 2010a, 2014; Korukonda and Doyley 2012; Korukonda et al. 2013; Poree et al. 2015). Synthetic aperture imaging systems (sparse array [SA] and plane wave [PW] imaging) acquire images with higher lateral resolution than those obtained with standard ultrasound systems. SA imaging systems transmit spherical waves sequentially from each element in the array, whereas PW imaging systems use all elements to transmit plane waves (*i.e.*, a single flash). Additionally, SA imaging systems apply focusing during both transmission and reception (two-way beamforming), whereas PW imaging applies focusing only on receive (one-way beamforming). Consequently, PW imaging produces images with high side-lobe levels and lower depth penetration, which reduces the precision of lateral strain estimates (Korukonda and Doyley 2012). However, the low transmission power of SA imaging systems restricts clinical applications. Plane wave imaging does not suffer from this limitation because all elements are active during transmission (it has higher transmission power). Furthermore, researchers have reported that spatial compounding reduces the side-lobe level incurred in PW imaging (Montaldo et al. 2009) because the compounding produces a synthetic transmit focus. We have observed that with appropriate beamforming, compounded plane wave (CPW) imaging can estimate lateral displacements precisely (Korukonda and Doyley 2012) and, thus, is a viable approach for estimating principal strains.

We hypothesized that CPW can visualize principal strains within the carotid artery. To test this hypothesis, we performed three groups of studies: (i) To assess the performance of principal and polar strain elastograms computed with CPW imaging, we simulated stable and unstable plaques with varying geometries. (ii) To assess how complex mechanical behavior manifests in principal strain elastograms, we performed simulation studies with transversely isotropic vessels with fibrous cap thickness ranging from 112 to 480 $\mu$m. (iii) To assess the performance of a prototype system of compounded plane wave vascular elastography, we performed a pilot study with 10 healthy volunteers.

## METHODS

The following subsections describe methods used in both the simulations and the patient study. These include beamforming, displacement and strain estimation and data analysis.

### *Beamforming*

We used the delay-and-sum method to reconstruct RF echo frames (Korukonda and Doyley 2012) on a 40 × 40-mm grid with lateral and axial sampling frequency of 52 lines/mm and 40 MHz. Several good references describe the principle of the delay-and-sum beamforming method (Van Trees 2002); therefore, this subsection provides only a brief description of the approach. We computed the backscatter intensity ($S$) at a given point ($x_0$, $z_0$) in the beamformed image as

$$S(x_0, z_0) = \sum_{i=1}^{N_{\mathrm{tx}}} \sum_{j=1}^{N_{\mathrm{rx}}} w_{ij} \mathrm{RF}_{ij}(t - \tau(x_0, z_0)) \qquad (1)$$

where $N_{\mathrm{tx}}$ and $N_{\mathrm{rx}}$ represent the total number of active transmission and reception elements, respectively; $\mathrm{RF}_{ij}$ represents the RF echo obtained when the $i$th element transmits and $j$th element receives; $t$ represents the time of flight of the echo; $\tau(x_0, z_0)$ represents the time of flight of the acoustic wave traveling from the transmit element to point ($x_0$, $z_0$) and back to the receive element; and $w_{ij}$