

● *Original Contribution***ACOUSTIC RADIATION FORCE IMPULSE IMAGING ON EX VIVO ABDOMINAL AORTIC ANEURYSM MODEL**ÁINE P. TIERNEY,^{*†‡} DOUGLAS M. DUMONT,[‡] ANTHONY CALLANAN,^{*†} GREGG E. TRAHEY,[‡]
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(Received 6 November 2009; revised 25 February 2010; in final form 27 February 2010)

Abstract—A method for reliable, noninvasive estimation of abdominal aortic aneurysms (AAA) wall mechanics may be a useful clinical tool for rupture prediction. An *in vitro* AAA model was developed from an excised porcine aorta with elastase treatment. The AAA model behaviour was analysed using acoustic radiation force impulse (ARFI) imaging techniques to generate and measure wave propagation in both aneurysmal and normal aortic tissue. Opening angle measurement showed a fourfold decrease from healthy aorta to AAA model and pathologic analysis verified this elastin degradation. Maximum wave velocity at 180 mm Hg was 7 mm/ms for healthy tissue and 8.26 mm/ms for the aneurysmal tissue. The mechanical changes produced in the artificially induced aneurysm were found to be detectable using ARFI imaging. (E-mail: tim.mcgloughlin@ul.ie) © 2010 World Federation for Ultrasound in Medicine & Biology.

Key Words: Abdominal aortic aneurysms, Acoustic radiation force impulse (ARFI) imaging, Radiation force, *In vitro* AAA model.

INTRODUCTION

An abdominal aortic aneurysm (AAA) is a focal balloon like dilation of the terminal aortic segment that occurs gradually over a span of years (Vorp 2007). The usual definition of AAA is an infrarenal aorta of diameter greater than 30 mm (Van Damme et al. 2005). Each year, there are 200,000 (United States), 500,000 (world-wide) newly diagnosed (Vande Geest et al. 2004). These aneurysms are hazardous because of their propensity to rupture. About 30% to 50% of patients with a ruptured AAA die before they reach hospital. Even with surgery, there is 50 to 70% mortality rate associated with rupture. The majority of AAAs are asymptomatic until rupture; this has led them to become the 13th most common cause of death in the US (Vande Geest et al. 2004).

The pathogenesis of AAA formation is not well understood. They are characterised by a destruction of elastin and collagen in the arterial wall. The underlying

problem in aneurysmal disease is this weakening of the aortic wall resulting in progressive dilation leading to eventual rupture (Golledge et al. 2006). Previous researchers have been successful in generating *in vivo* aneurysms in small animals, *e.g.*, rabbits, rats, mice (Anidjar et al. 1990; Sinha et al. 2004) and *ex vivo* in porcine aortas (Kratzberg et al. 2009) using a method of elastase perfusion in the aorta. The *in vivo* elastase perfusion has been found to lead to subsequent aortic dilatation, collagen and elastin degradation, MMP upregulation and an extensive inflammatory cell infiltrate in the outer media and adventitia of the aortic wall, which is typical of AAA formation (Sinha et al. 2004). The elastin degradation in the wall will be characterised by the opening angle alterations due to elastase treatment, similar to Fan et al. 2005.

The maximum diameter of an AAA has long time been considered as the main determinant in predicting its risk of rupture, *i.e.*, when the AAA reaches 5.5 cm it is thought that the risk of rupture warrants repair (Van Damme et al. 2005). However, several studies have questioned the reliability of this criterion by showing that small aneurysms (<5 cm) can rupture and that larger aneurysms (>5 cm) can remain quiescent for years (Darling et al. 1977; Doyle et al. 2009; Limet et al. 1991). This

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coupled with a 4% to 5% mortality rate with interventional surgery indicates a critical need for improved noninvasive AAA rupture predictors (Lasheras 2007).

Therefore, there exists a need for a noninvasive, cost-effective, safe and accurate mechanism for detecting changes in abdominal pathology. Several ultrasonic methods have been previously investigated. Intraoperative ultrasound (IOUS) has been shown to be effective in detecting changes in liver pathology (Cervone et al. 2000) but the invasive nature of IOUS would restrict its use. Intravascular ultrasound (IVUS) elastography has recently shown promise in the characterization of focal plaques in coronary arteries (Schaar et al. 2003). Again a challenge for this method, however, is the introduction of the ultrasonic probe within the vessel lumen, which exposes the patient to the risk of dislodging a vulnerable plaque. Acoustic radiation force impulse (ARFI) imaging presents an attractive method as it involves remote interrogation with short acquisition times, which because is implemented on a diagnostic ultrasound machine is at relatively low cost.

ARFI Imaging is a relatively new imaging modality that has been developed in Duke University (Durham, NC, USA) over the last 10 years. Acoustic radiation force is a phenomenon associated with the propagation of acoustic waves through a dissipative medium (Fahey et al. 2008a). It is caused by a transfer of momentum from the wave to the medium, arising either from absorption or reflection of the wave (Torr 1984). This momentum transfer results in the application of a body force in the direction of wave propagation (Nightingale et al. 2002b).

ARFI imaging provides information about the local mechanical properties of bodily tissue. The acoustic radiation forces generate localised displacements in the tissue and these displacements can be tracked using ultrasonic methods (Trahey et al. 2004). The tissue response to these forces can be monitored both spatially and temporally. The tissue displacements are inversely related to tissue stiffness (Nightingale et al. 2001).

Radiation force has also been demonstrated to generate propagating waves within tissue (Sarvazyan et al. 1998; Zhang et al. 2005). Wave propagation speed is directly related to the mechanical properties of the tissue. Estimates of vascular stiffness can be derived by measuring the velocity of the propagating wave. Wave generation similar to Zhang et al. 2006 was employed in this study. A single transducer on a diagnostic scanner is used to both generate the radiation force and track the wave velocity and displacements (Trahey et al. 2004). ARFI Imaging has been shown to be effective in cardiac/liver ablation monitoring, breast mass imaging and monitoring cardiac myocardial stiffness (Fahey et al. 2008b; Hsu et al. 2005). Challenges exist in adapting the ARFI

imaging method so it can effectively displace and effectively monitor the dynamics of deep lying tissues, such as the abdominal aorta.

This article uses ARFI to examine the material responses in aortic tissue and in phantom AAA animal tissue models examining the effect of elastin reduction on mechanical parameters. This article hypothesises that ARFI could be implemented to provide additional information on the changing mechanical properties of an AAA that lead to rupture.

MATERIALS AND METHODS

Ultrasound measurement

All *ex vivo* imaging was implemented on a Siemens Antares platform (Siemens Medical Solutions USA, Inc., Ultrasound Division, Issaquah, WA, USA) with a VF10-5 handheld transducer at a frequency of 8 MHz. The scanner has been modified to allow user control of the acoustic beam sequences and intensities and access to raw radio-frequency (RF) data.

ARFI measurement

In soft tissues, where the majority of attenuation results from absorption and under plane wave assumptions, this radiation force magnitude can be represented by the following equation (Nightingale et al. 2000; Nyborg 1965; Starritt et al. 1991; Torr 1984, Trahey et al. 2004)

$$F = \frac{W_{\text{absorbed}}}{c} = \frac{2\alpha I}{c} \quad (1)$$

where F [dyn/(1000 cm)⁻³], or (kg s⁻² cm²), is acoustic radiation force, W_{absorbed} [W (100 cm)⁻³] is the power absorbed by the medium at a given spatial location, c [m s⁻¹] is the speed of sound in the medium, α [m⁻¹] is the absorption coefficient of the medium and I [W cm⁻²] is the temporal average intensity at a given point in space. For a focused acoustic beam, the radiation force is applied throughout the focal region of the acoustic beam.

During the ARFI sequences, an initial reference line is acquired using standard B-mode parameters. Reference lines are used to establish the initial tissue position. This is followed by a high-intensity focused “push” pulse, which mechanically excites the tissue. The excitation pulse is then followed by a series of tracking pulses, which are utilized to monitor the tissue displacement response.

Displacement measurements were calculated using cross-correlation between 0.5 mm kernels from a reference line and subsequent tracking lines. This was performed over an approximate 60 mm lateral field of view on the proximal wall. Each dataset was filtered to remove linear bulk motion artefacts. Displacements within the lumen were masked out. The mean of the displacements for the four acquisitions was calculated.

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