Wave Propagation in the Human Brain and Skull Imaged in vivo by MR Elastography

E.H. Clayton¹, G.M. Genin¹, and P.V. Bayly¹,²

¹ Washington University in St. Louis/Department of Mechanical, Aerospace and Structural Engineering, Saint Louis, USA
² Washington University in St. Louis/Department of Biomedical Engineering, Saint Louis, USA

Abstract—Traumatic brain injuries (TBI) are common, and often lead to permanent physical, cognitive, and/or behavioral impairment. TBI arises in vehicle accidents, assaults, athletic competition, and in battle (due to both impact and blast). Despite the prevalence and severity of TBI, the condition remains poorly understood and difficult to diagnose. Computer simulations of injury mechanics offer enormous potential for the study of TBI; however, computer models require accurate descriptions of tissue constitutive behavior and brain-skull boundary conditions. Lacking such data, numerical predictions of brain deformation remain uncertain. Brain tissue is heterogeneous, anisotropic, nonlinear, and viscoelastic. The viscoelastic properties are particularly important for TBI, which usually involves rapid deformation due to impact.

Magnetic resonance elastography (MRE) is a non-invasive imaging modality that provides quantitative spatial maps of biologic tissue stiffness in vivo. MRE is performed by inducing micron-amplitude propagating shear waves into tissue with a surface actuator at steady state while images of the wave motion are acquired using a standard clinical MRI scanner. A custom synchronized MRI pulse sequence, with “motion-sensitizing gradients”, is used to encode wave displacements, and at various time points. Elastograms, or images with contrast corresponding to complex shear modulus (storage and loss modulus), can be computed from the raw spatial-temporal displacement data by inverting the governing equations of motion. Wave images and elastograms can provide fundamental insight into the dynamics of human brain and skull under rapidly time-varying loads.

In this study, we aim to understand in vivo brain motion as the cranium is exposed to acoustic frequency pressure waves (45 Hz). This loading approximates some physical features of blast, albeit at very low levels.

Keywords—MR-Elastography, TBI, brain response, non-invasive measurement, blast loading.

I. INTRODUCTION

The US Centers for Disease Control and Prevention (CDC) estimates that each year 1.4 million Americans suffer a traumatic brain injury [1]. The prevalence of TBI is surely related to the variety of ways one can subject their brain to insult. Most TBI cases are caused by one or more of the following events: direct physical contact with an external rigid body, inertial loading due to linear or angular acceleration of the head, or external pressure loads (i.e. blast wave). Despite its importance, little is known with confidence about the mechanics of TBI.

Diffuse Axonal Injury (DAI) is a prominent manifestation of TBI. It has been suggested that DAI is a micro-structural process in which neural axons are stressed beyond a limit, initiating a biochemical cascade that leads to axon destruction [2]. While uncertainty remains as to the biochemistry of DAI, understanding in vivo brain motion and brain-skull interaction as the cranium is subjected to external loading is fundamental to understanding TBI and may help explain why micro-structural changes appear where they do.

Experimental data is urgently needed to provide estimates of mechanical properties for use in computer simulations. Characterization of brain-skull interactions is equally important; the attachments that transmit force from skull to brain are critical in determining brain deformation [3]. In addition, computer simulations need to be validated before they can be relied upon to illuminate TBI or design countermeasures.

This study uses magnetic resonance elastography (MRE), an imaging modality designed specifically to measure oscillatory motion in soft tissue [4]. The objective is to illuminate the mechanical behavior of brain tissue while the skull is subjected to acoustic pressure waves. Results obtained will be applicable for both characterization of mechanical features of the brain and skull, and validation of computer models.

II. METHODS

Four human subjects, aged 18-29 years-old, were imaged in this study. Experiments were conducted at 1.5T on a MAGNETOM Avanto (Siemens) whole-body clinical scanner in Washington University’s Center for Clinical Imaging Research (CCIR) facility. The MRE imaging sequence implemented for this study consists of a specialized gradient-recalled echo (GRE), phase-contrast magnetic resonance imaging sequence. The sequence differs from a traditional
GRE imaging sequence in three ways. First, phase-contrast MR images are acquired, instead of magnitude images. Second, the sequence is triggered to acquire data in synchronization with an external actuator and repeated with time delays to acquire data at several time points. Third, a single cycle, trapezoidal magnetic gradient is added to the imaging routine. The "motion encoding gradient" (MEG), is responsible for spatially recording tissue motion at the MEG frequency. All procedures were approved by the institutional Human Research Protection Office Internal Review Board to ensure that the rights and welfare of the human research participants were protected.

The brain was imaged using a commercial phased-array MRI head coil with the subject in the supine position. Motion was transmitted to each subject’s skull by affixing two acoustic paddle actuators (Resoundant™, Resoundant Inc., Rochester, MN) near the left/right temples with Coban™ bandaging (Fig. 1). The paddle actuator geometry is similar to that of a timpani (kettle) drum. It consists of a rigid plastic hemispherical frame with a flexible membrane stretched across the bowl. Flexible plastic tubing acts as a pressure waveguide, connecting each passive paddle actuator to an active driver outside of the MRI scanner magnetic field. The active driver consists of a voice coil actuator enclosed by cabinet. Pressure waves generated by the active driver are transmitted through the flexible tubing to the passive paddles, causing the paddle’s flexible membrane to vibrate on the skull.

The paddle actuators were configured to transmit a four cycle pressure wave at 45 Hz in synchronization with the

MRE imaging sequence. The input excitation amplitude and waveform were quantified by accelerometers attached to the exterior bowl (opposite the paddle-subject contact surface) of each paddle. A typical MRE image acquisition acceleration time history is shown in Fig. 2.

The custom gradient-recalled echo MRI pulse sequence contained 45 Hz oscillating magnetic field gradients to encode tissue displacements in the direction of the gradient at the same frequency. A single 3 mm thick trans-axial image slice was acquired with TR: 133.3 ms, TE: 27.5 ms, flip angle: 25˚, and in-plane resolution of 3 mm². The imaging sequence was repeated three times to record all displacement components (u, v, and w) relative to a laboratory coordinate system.

III. Results

Experimental data obtained from this study is presented in Figure 3. Propagating displacement waves were observed at wavelengths of 4-5 cm and propagation speeds of 2-3 m/s. These wavelengths and speeds are consistent with shear motion in a viscoelastic medium with storage modulus 3-5 kPa. Dissipation is apparent by the decay in amplitude as the waves propagate into the interior of the brain. Reflection from interior boundaries (for example, at the falx cerebri between the brain hemispheres) is also apparent. Interestingly, while the excitation is almost perfectly symmetric, the brain’s response at this frequency is anti-symmetric (Fig. 3). These observations are especially interesting because they are consistent among all of the four subjects studied to date.