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Material characterization and modeling of head for dynamic simulations

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Abstract: The modeling of the response of the human head to blast like loading is of importance for many applications including the study of traumatic brain injury resulting from improvised explosive devices. One key issue in simulating the response of the head is to have models that are characteristic of the response of the head and its components under these conditions. We review different characterization efforts for evaluating the response of the skin, skull, and brain within this window of response and use these results to develop models appropriate for the characterization of each component. We discuss efforts made to construct an artificial head model capable of simulating similar results in a blast environment.

Keywords: bTBI, head, skin, skull, brain, high loading rate characterization, ultrasonic tests.

1 Introduction

Blast traumatic brain injury (bTBI) currently represents the main cause of military TBI, but it is a poorly understood form of TBI since little is known about the effects of blasts on the human head and the brain injury thresholds for blast like loadings have not been established yet [Agoston, Gyorgy, Eidelman and Pollard (2009); Nyein, Jerusalem, Radovitzky, Moore and Noels (2008)]. Some researchers have recently shown that mild TBI can be caused by the early time intracranial wave motion [Taylor and Ford (2008); Chafi, Karami and Ziejewski (2007)]. These researchers have shown that intracranial wave motion can generate significant intracranial pressure, negative pressure and shear stress in the brain causing TBI.

The interactions of a blast wave with the head (the interaction duration was shown as about 0.05 ms with sudden peak pressure of about 4 MPa using air skull interface simulation [Taylor and Ford (2008)]) results in the propagations of stress waves among and within different components of the head along with dissipations of energy due to the viscosity and heterogeneity of the head tissues. This wave

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propagation event in the head plays an important role within 2 ms to reaching maximum stresses [Ganpule (2009)]. This part of the blast response is characterized by the peak pressure, the duration acting on the head and the wave propagation properties or the high strain rates responses of the head.

A lot of effort has been made to simulate human head response under various blast conditions to get brain injury information, but there is a time scale mismatch between the blast loading which generates waves in the head and the current constitutive models used for human head (skin, skull and brain) which are based on low frequency dynamic or low strain rate tests which can not represent the wave propagation properties of the head. For example, the role of shock absorption (or energy dissipation) of the skin, skull and the brain under blast waves could be very different than under quasi-static conditions. Unfortunately, there are few references that take into account the energy absorption properties of skin and skull, and consider the high strain rate response of brain, which might be the main difference between blast TBI and normal impact TBI.

This paper is organized as follows: first a short introduction to blasts is presented, then the high loading rate characteristics of the main components of the head (skin, skull and brain) are reviewed from literatures, followed by a preliminary method to construct a constitutive model which can be used to capture the wave propagation properties of viscoelastic materials.

2 Literature review

Blast waves are pressure waves with finite amplitude that propagate from atmospheric explosions. The explosion releases a large amount of energy in a short duration of time. A blast creates a large discontinuous pressure wave in which density increases in the air and which can cause large loading transients on an object. A simple free field blast wave can be described by the idealized Friedlander waveform with a rapid rise to the peak pressure and then an exponential drop of the overpressure, together with a prolonged under-pressure, which results in a combination of compression and tension in the time scale of milliseconds as it impacts on a sample and propagates through it [Taylor (1950)]. This represents very rapid loading.

Since we are focused on bTBI, the tolerance of humans to this blast impact should control the magnitude of the wave considered in modeling and studies. Bowen curves, which provide an estimated tolerance to a single blast at sea level for a 70-kg individual with the human oriented perpendicular to the blast [Bowen, Fletcher, Richmond, Hirsch and White (1968)] are usually chosen as the references for characterizing and then simulating the blast loadings on the human head.

The Bowen curves show that the threshold for unarmored lung injury is 0.55 MPa (80 psi) (equivalent to a free air explosion of 0.0648 kg TNT at a 0.6 m standoff distance), and 1.7 MPa (250 psi) (equivalent to a free air explosion of 0.324 kg TNT at a 0.6 m standoff distance) is the lethal dose, with approximately 50% survival from lung injury, while 2.75 MPa (400 psi) results in 1% survival from lung injury [Bowen, Fletcher, Richmond, Hirsch and White (1968); Nyein, Jerusalem, Radovitzky, Moore and Noels (2008)].

2.1 Skin

Response to rapid loading of the skin has been studied through ultrasonic characterization. In addition, slower tests also exist that characterize the Dynamic Mechanical Analysis (DMA) with frequency less than 100 rad/s [Holt, Tripathi and Morgan (2008)], biaxail tension test [Aba (1994)], confined and unconfined compression tests [Wu (2003)], dynamic indentation with a frequency range from 10 to 60 Hz [Boyer (2007)]. Even though useful for understanding and modeling the general response, the time scale here is too different from the scale relating to blast.

From ultrasonic tests one can obtain the wave speed, attenuation coefficient (used to describe energy loss) and backscatter value (used to describe non-homogeneity of the material structure) at different frequencies. The acoustical speed, attenuation coefficient and backscatter value have been measured with frequencies between 20 to 30 MHz on excised human skin samples from the upper and lower back, chest and abdomen [Moran (1995)], the speed of sound in the epidermis is 1645 m/s and in the dermis is 1595 m/s with little change within this frequency range, and shows that attenuation increases with the increase of frequency following a power law relation. In vivo ultrasonic attenuation and backscatter coefficients of normal human forearm dermis using frequencies between 14 to 50 MHz and subcutaneous fat using frequencies from 14 to 34 MHz were determined in [Raju and Srinivasan (2001)], and fitted by a linear relation between attenuation and frequency. Ultrasound properties of rabbit and human skin tissues under various transverse stresses have been studied in vitro over the frequency range from 15 MHz to 40 MHz in [Pan, Zan and Foster (1998)] and have shown that the ultrasound attenuation coefficients decrease significantly with increasing strain, but the speed of sound and backscattering coefficients only exhibit little dependence on the pre-strain applied.

2.2 Skull

The skull bone is heterogeneous composed of compact bone and spongy like diploe filled with softer marrow. Because of the large differences in mechanical properties between the bone and marrow, the diploe is a strong scatterer of ultrasonic waves. The sound speed in compact bone is about 3000 m/s, the compact bone attenuates for MHz-range ultrasonic waves very strongly and much larger than that in soft tissues and biological fluids [Mobley, Kasili, Norton and Vo-Dinh (1999)].

Ultrasonic techniques have been used to explore the viscoelastic properties of bone via wave attenuation, where bone exhibits high viscoelastic damping at the ultrasonic frequency range [Lakes (2004)]. There are studies that consider the relative contribution of bone and soft tissue in the attenuation of peak dynamic forces as a function of frequency [Paul, Munro, Abernethy, Simon, Radinand and Rose (1978)], and have shown that soft tissue and bone attenuated peak forces most effectively at higher frequency ranges. The underlying physics of propagation of ultrasonic waves in the head has also been studied for various ultrasonic transducers by measuring the attenuation coefficient and phase velocity for ultrasonic propagation in samples of brain tissue and skull bone from sheep [Mobley, Kasili, Norton and Vo-Dinh (1998)]. They used these material properties to investigate the propagation of ultrasonic wave fields in the head.

2.3 Brain

The complete characterization of brain response to high loading rates is yet incomplete, even though much has been understood for this highly heterogeneous material consisting of gray and white matter and which includes large geometric features. There is a large body of results that can be drawn from and which indicate a large rate dependence in brain matter. For example, unconfined compression and relaxation tests (with strain rate < 0.01 1/s) on bovine brain white matter [Cheng and Bilston (2007)] is very different from high speed compression tests done by using Split Hopkinson pressure bar with strain rates 1000/s, 2000/s, and 3000/s [Pervin and Chen (2009)] from the same region as [Cheng and Bilston (2007)]. These tests indicate that for the same strain level, the stress was 3 orders of magnitude higher in 3000 1/s tests than that observed in 0.01 1/s tests.

There is both stress relaxation and oscillatory tests done on brain. The step relaxation tests include rise times that are outside the scale needed for blast impact. For example, results are reported for a rise time of 20 ms in [Bilston (2001)], and greater than 7 ms in [Tamura (2008)]. The frequency dependant response has also been evaluated up to 10Hz in [Bilston (2001)], up to 16 Hz in [Brands (1998)], and up to 6310 Hz in [Nicolle, Lounis, Willinger and Palierne (2005)]. The results reported in [Nicolle, Lounis, Willinger and Palierne (2005)] provide conditions that might be considered in the range of linear response in shear at up to 6310 Hz, but when converted into strain rates provide a maximum strain rate of only 25 1/s, which is still outside the range considered for blast loading.

Currently only the Split Hopkinson bar test and ultrasonic techniques can be used to estimate high loading rate response of brain tissue. There is measurements of absorption coefficient of mammalian brain in the frequency range of 0.5 to 7 MHz at 37˚C [Goss, Frizzell and Dunn (1979)], which shows a linear dependence of the absorption on the frequency. Also there is comparison of the ultrasonic absorption and attenuation at 1 MHz for brain tissue [Damianou, Sanghvi, Fry and Maass-Moreno (1997)]. This indicates that the absorption coefficient is 2.4 neper/m while attenuation coefficient is 7 neper/m at 1 MHz. Two ultrasonic techniques are used to separately measure the longitudinal ultrasonic velocity (1557 m/s), absorption coefficient (0.14 1/cm) and the bulk modulus (2.7 GPa) of the bovine brain gray matter at 1.7 MHz [Etoh, Mitaku, Yamamoto and Okano (1994)]. Also there are measurements of attenuation as a function of frequency for lamb brain white and gray matter for frequencies from 232 kHz to 584 kHz [Lin, Shieh and Grimm (1997)]. This shows a different dependence than reported in [Etoh, Mitaku, Yamamoto and Okano (1994)], and they have also studied anisotropic attenuation behavior in the white matter.

3 Viscoelasticity

As shown previously, in order to capture the responses of human head under blast like loadings, the experimental characterization has to be in the same time scale as the blast loadings. Once obtaining these results, it would be useful to fit the response to a simple model. In the linear range this can be done. For the simplest assumptions, one can take the response to be isotropic, which decouples the response into bulk and shear terms, each having a relaxation modulus. For a Maxwell model, each term can be represented by a relaxation function $G(t)$ in the form

$$G(t) = G_o \exp(-t/t_R) \tag{1}$$

where G_o is the instantaneous modulus and t_R is the relaxation time. We would like to select these material moduli in a way that will result in the observed responses at the loading rates of blast.

For a linear material, this can be studied, for example, by studying the application of a step pulse on a half space, which then can be extended to arbitrary pulse shapes by superposition. We will consider a shear response characterized by the lateral displacement function $u(x,t)$ in terms of the axial location x and time t. The step pulse can be characterized by $u(0,t) = H(t)$, where $H(t)$ is the Heaviside step function. This motion has a velocity field given by $v(x,t)$ subject to the impulse $v(0,t) = \delta(t)$, where $\delta(t)$ is the Dirac delta function.

For a general linear viscoelastic material we have the constitutive equation for stress $\sigma(t)$given by the convolution integral

$$\sigma(t) = \int_{-\infty}^{t} G(t-\tau)\frac{\partial v(x,\tau)}{\partial x}d\tau \tag{2}$$

In the absence of body forces, this must be introduced into the equation of motion given by

$$\frac{\partial \sigma}{\partial x} = \rho \frac{\partial v(x,t)}{\partial t} \tag{3}$$

where ρ is the density. The result of substituting Eq.2 into Eq.3 is an integral equation given by

$$\int_{-\infty}^{t} G(t-\tau)\frac{\partial^2 v(x,\tau)}{\partial x^2}d\tau = \rho \frac{\partial v(x,t)}{\partial t} \tag{4}$$

We can now seek the solution as the sum of solutions constructed from the response to the aforementioned step. This can be written as

$$v(x,t) = \int_{-\infty}^{t} F(t-\tau)v(0,\tau)d\tau \tag{5}$$

where $F(t)$ is the response to a single unit step displacement. The Laplace transform of Eq.4, after substitution of Eq.5, yields

$$\bar{G}(s)\frac{\partial^2 \bar{F}(x,s)}{\partial x^2} = \rho s \bar{F}(x,s) \tag{6}$$

Given the boundary condition of unity at the impulse and zero at infinity, one obtains

$$\bar{F}(x,s) = \exp(-\sqrt{\frac{\rho s}{\bar{G}(s)}}x) \tag{7}$$

The inversion of this for a general relaxation function is problematic, but we can seek to see the form of the relaxation function that results in an exponentially attenuating steady wave. For this we expect the wave to have the form of the step, but to be shifted through a wave speed c, associated with the motion of the step. For the exponentially attenuating wave of initially unit step we have $u(x,t) = \exp(-\alpha x)H(ct-x)$, where α is the attenuation defined by $\alpha = -[\ln u(x,t)]/x$. This results in a velocity of the form

$$F(x,t) = v(x,t) = c\exp(-\alpha x)\delta(ct-x) \tag{8}$$

The Laplace transform of this is

$$\bar{F}(x,s) = \exp(-\alpha x)\exp(-sx/c) = \exp[-(\alpha + s/c)x] \tag{9}$$

Comparing Eq.7 and Eq.9 provides an expression for the Laplace transform of the relaxation modulus given by

$$\bar{G}(s) = \frac{\rho s}{(\alpha + s/c)^2} \tag{10}$$

This can be inverted to give the relaxation function

$$G(t) = \rho c^2(1-\alpha ct)\exp(-\alpha ct) \tag{11}$$

This relaxation is only good for short times when $t < 1/\alpha c$, and can be problematic in numerical procedures, particularly noting that the relaxation characteristics are different for the different modes of deformation. This limiting time for compression waves moving into the brain would be approximately 0.01 ms, which would correspond to approximately 0.14 m of travel before the relaxation modulus would be zero. Even though this relaxation function is not that of the simple Maxwell model given in Eq.1, and is severely restricted in application time for simulations in the ms time range, the obtained relaxation modulus can be approximated by a Maxwell model by fitting the initial value and initial slope. The resulting model would not be restricted by this time limit, and would also not produce steady waves, but would be attractive for numerical simulation. In this case, we would get an equivalent Maxwell model of Eq.1 that would be given by material parameters

$$G_o = \rho c^2 \qquad t_R = \frac{1}{2\alpha c} \tag{12}$$

4 Summary

We have reviewed a number of results that provide an experimental characterization of the mammalian head (skin, skull and brain) which can be used to model the responses of the head under blast like loadings appropriate for study of bTBI. We also provide a description of how to provide an equivalent Maxwell model for the attenuation results, but obtained based on assuming steady wave motion that is exponentially attenuating. The Maxwell model would be more stable for numerical analysis and could be used, through superposition of elements, to obtain the response at multiple time scales. This also provides a pathway to extend the modeling to larger strains through standard methods.

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